GREGG VANDERHEIDEN

RESNA'86

Employing Technology

Proceedings of the Ninth Annual Conference on Rehabilitation Technology

Minneapolis

City of Lakes

June 23-26, 1986 • Minneapolis, Minnesota

RESNA — Association for the Advancement of Rehabilitation Technology

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Max Donath
Harry Friedman
Marty Carlson
Editors

Ray Fulford
Tony Langton
Conference Co-Chairs

June 23-26, 1986 • Minneapolis, Minnesota

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These Proceedings of the Ninth Annual Conference on Rehabilitation Technology reflect the theme of the Conference, "Employing Technology."

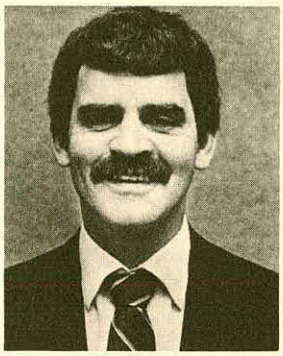
As an organization, RESNA has undergone some of the same changes associated with the growth and expansion of technology, and that growth and expansion exemplifies the spirit and character of an organization that will soon celebrate its 10th anniversary.

RESNA has a deep concern for the application of science and technology to the rehabilitation process. Its members include rehabilitation professionals from diverse backgrounds and disciplines: engineers; occupational, physical, speech, and other therapists; physicians; special education personnel; other health care providers; and not least, consumers.

As these proceedings demonstrate, the conference addressed not only the practical applications of technology but included reports from laboratories engaged in research and development related to every conceivable aspect of rehabilitation.

The annual RESNA conferences and the proceedings that are an integral part of them are among the major resources for all those who today devote their professional lives to the service of handicapped people, as well as for those yet to come.

Sheldon R. Simon, M.D.
President



Raymond Fulford
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Researchers continue to be challenged by the demands associated with the development of effective rehabilitation modalities. An incredible diversity of technology and innovative clinical approaches are being brought to bear on solving some of the problems associated with sensory, neuromuscular or skeletal dysfunction.

This volume contains the papers presented at the 1986 RESNA Annual Conference sponsored by the Association for the Advancement of Rehabilitation Technology and held in Minneapolis. A wide variety of topics are covered, and thanks to the assistance of the respective SIG chairpeople and their assocates, have been able to develop an exciting scientific program which reflects the diversity of the membership, and of the diverse professionals that make up the rehabilitation team.

The conference this year reflects several new developments. Two new Special Interest Groups were formed--one on robotics chaired by Larry Leifer of the VA Rehabilitation R&D Center in Palo Alto, CA and the other on biomechanics chaired by Peter Walker of the VA Medical Center in West Roxbury, MA. Research in these two areas has attracted considerable attention in the rehabilitation community as reflected by a continued increase in both the number and quality of the presented papers.

This year we also have been fortunate in developing three special sessions which have attracted participation by a group of highly respected professionals from both here and abroad. The special session on robotics was organized by Rick Foulds of the Tufts University REC with the support of the World Rehabilitation Fund. A special session on voice technology was organized by Robert Mills of the Electronic Industries Foundation REC. The technology transfer special session was organized by Seldon Todd of the VA Office of Technology Transfer.

We would like to extend our sincere thanks to the above organizers, the authors, and the anonymous reviewers whose time and efforts have made the Conference a success. Without the organizational capabilities of the RESNA staff none of this would be possible. The results you see are a reflection of their attention to detail and their efforts above and beyond the call of duty.

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Scientists and Engineers with Disabilities Evaluate Assistive Devices

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American Association for the Advancement of Science

The social-psychological aspects of devices intended to assist disabled persons have not yet been investigated although assistive devices such as wheelchairs and hearing aids have the potential to become an extension of the user's self and to influence social interaction . In an effort to initiate that research, the current study undertakes an exploratory investigation of disabled persons' perceptions of their own assistive devices. A national mailed survey of disabled scientists and engineers was conducted because this group would be particularly likely to utilize devices and to use them in a variety of social settings. This paper reports the first phase of the research findings frin tge syrvet, The high response rate (47.4%, N=595) indicates the subjects' interest in the topic. Associations among demographic, attitudinal, and social responses to assistive devices are presented here. In addition, there will be a discussion of the AAAS Resource Group of Scientists and Engineers with Disabilities as consultants in the design and evaluation of adaptive devices and service delivery systems.

The current study undertakes an exploratory investigation of disabled adults' perceptions of assistive devices. A national mailed survey was conducted using the Resource Group of Scientists and Engineers with Disabilities which has been identified by the American Association for the Advancement of Science. Although this resource group is a very specialized segment of the disabled population, it was selected for study because the members would be particularly likely to utilize devices and to use them in a variety of social settings. The total number of questionnaires returned is 595, representing 47.4 percent of the 1254 persons on the mailing list. The study gathered data on the following topics; subjects' demographic and disability-related characteristics; in which social settings they utilize devices; and attitudes toward their assistive devices. When patterns of device utilization and attitudes toward devices are observed in association with other dimensions, first-stage analysis shows that: (1) demographic variables generally are not associated with device use or attitudes about devices, (2) that disability characteristics frequently are associated with utilization and sometimes with attitudes, and (3) that social settings are frequently associated with attitudes. Findings generated in this exploratory study will guide selection of variables which should be used in further research.

The research population is 78% male, highly educated (38% with a graduate degree), and likely to be employed (66%). Eightly percent who are employed work in professional or technical occupations. Eight percent who are employed work part time, four percent are self-employed, and 6% are retired. The mean age is 47.2 with a range from 19 to 88 years. Respondents reported having various disabling conditions: vision impairment, 13% of the population; hearing, 17%; speech, 17%; limb loss, 27%; neuromuscular, 35%; and multiple or other disabilities, 4%. Assistive devices are more likely to be used for mobility, transportation, and employment. The devices least likely to be used are those which assist housekeeping and childcare. There was considerable agreement on the type of device considered most significant for full independence. Forty-four percent listed mobility aids as most significant. Communication aids hold the second position with 10 percent reporting these as most significant to independence.

As expected, the type of device utilized varies with disability type. Respondents with sensory impairments are likely to report using devices for employment, education, and communication activities whereas persons with nonsensory disabilities are associated with more general device uses such as personal care, transportation, and mobility. Variation also appears in the social context of devices identified as significant to full independence. While those with neuro-muscular impairments use significant devices in employment, education, friendship, family and solitary situations, visually and hearing impaired persons are least likely to use significant devices in those settings or in public. Persons with visual impairment are also associated with a unique pattern of device use in socially informal settings; visual impairment is the only disability type associated with "never" using significant devices in friendship and family situations. Perhaps devices deemed significant to full independence are less likely to be applied in less formal social roles. These findings suggest distinctions in the way devices are perceived as important to independence.

Survey respondents were also asked to indicate feelings elicited by assistive devices. Public response to assistive devices was perceived as "about average" with more respondents observing positive rather than negative responses from the public. A large proportion (66%) do use devices to avoid the sometimes uncomfortable event of receiving help from others. Individual attitudes toward devices chiefly are positive. Fifty-seven percent say that using devices is always socially acceptable. Other responses show the population sees assistive devices as mostly beneficial, necessary, and a key to normality. But thirty-two percent say devices are often restricting, thirty-two percent feel devices are often inconvenient and only eleven percent say they are always a source of pride.

Respondents attitudes toward devices were observed in association with other dimensions such as demographic characteristics, disability-related characteristics, utilization of specific device types, and social settings. The questionnaire inquired how frequently devices elicit specific attitudes. Attitudes when measured in association with other research dimensions do show that negative attitudes have different associations from positive attitudes toward devices. Negative attitudes tend to show weak associations with respondents' use of specific types of devices and their choice of devices considered significant to full independence. In contrast, positive attitudes tend to show moderate associations with social variables such as device use in public, employment, friendship or family settings. Possible interpretations suggest that positive attitudes toward devices may be more frequently expressed by people who are engaged in social activities, or that social settings encourage positive attitudes about devices, or that an external element influences both social settings and attitudes about devices. Further research is needed to clarify the relationship between attitudes about devices and the other major dimensions of disabled persons' lives.

The AAAS Resource Group of Scientists and Engineers with Disabilities serves as a consultant pool for researchers, designers, evaluators, manufacturers, marketers and perscribers of technologies . Individuals, universities, companies and rehabilitation facilities interested in using this resource can contact AAAS for appropriate lists from the data base. Ways to use the consultants in the Resource Group will be discussed, especially in regard to design and evaluation of adaptive devices and service delivery systems.

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ESTABLISHMENT OF A COMPETITIVE INDUSTRY TO EMPLOY DISABLED PERSONS: INITIAL BUSINESS OBJECTIVES

1.2

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ABSTRACT

This paper presents some of the primary objectives for getting an industry, new or existing, involved in the vocational rehabilitation, or first time employment situation (habilitation), of disabled persons by providing competitive employment. The classical sheltered workshop alternative can be a useful interim approach but is not the most cost effective. It violates the economic principle that total production costs, including wages and all supplemental support expenses, must not exceed revenue generated. The optimum solution appears to be to establish and operate a competitive industry to employ disabled persons working alongside those considered "able-bodied". The competitive industry may be considered the ultimate service delivery system for vocational rehabilitation and placement.

INTRODUCTION

Reliable statistics on the number of potentially employable disabled persons are difficult to assemble since there is significant inconsistency as to the definition of a "disabled person". Published estimates range from as low as 13 million to as high as 50 million (1)(2), depending on the census and public reporting methods, and on answers to the question "disabled for what?". Regardless of the unemployment figures used, a major gap exists between the obvious "demand" to gainfully employ disabled persons, especially the "severely disabled", and the apparent short "supply" of industries fully equipped, both technologically and philosophically, to do so. Research indicates that current attempts to fill the gap may be grouped into four major categories:

1. Organizations Planning Competitive Industries to replicate Center Industries Corporation (CIC), Wichita, Kansas.
2. Existing Competitive Industries, Large and Small, employing varying ratios of disabled workers.
3. Sheltered Workshops with government wage exemptions.
4. Special Placement Services such as Projects With Industry (PWI), etc.

The histories, current experiences, and future plans of selected operations from each of the above four categories have been studied. It is from the information received that most of the objectives for competitive employment have been written.

The organization from which the majority of material has been gathered is CIC. It was the primary resource for three important reasons. First, it is the industry with which this author is most familiar. Secondly, the organizations referred to in category 1. above have been consulting with the originators of CIC regarding endeavors to establish similar industries; and thirdly, CIC has been markedly successful at competitively employing disabled persons, some quite severe, since 1975.

BACKGROUND

CIC is a not-for-profit corporation with approximately 70,000 square feet of manufacturing space currently employing 112 persons; more than 80 of whom are disabled. Its annual gross sales to more than 30 major customers exceeds $2 million and the average hourly wage of its employees is about $4.30. CIC continues to grow as a competitive manufacturing facility. In fact, employment has nearly doubled since 1983; gross income has risen more than 30 percent in the past two years, and a physical plant expansion of over 30 percent has taken place in the past year. CIC serves also as a laboratory where rehabilitation engineers attempt innovative schemes to augment the physical capabilities of the handicapped workers. The continued growth at CIC is evidence that the establishment and operation of a competitive industry to employ disabled persons is not only achievable but can offer ever increasing employment opportunities. Consequently, it has been determined appropriate to investigate why and how CIC came to be and continues to grow and to determine how similar systems might be established and operated in other locations.

INITIAL BUSINESS OBJECTIVES

Notice in the heading of this section that the word "BUSINESS" is used as an adjective to describe the type of initial objectives to be established. It serves the same purpose as the term "COMPETITIVE" in the main title of this paper. If the primary intention of establishing an industry to employ disabled persons is social or humanitarian in nature, then the most critical specification for the industry has been overlooked. It is imperative that the industry become self-sustaining within a scheduled period of time and that all its employees receive their major subsistence through the revenue generated. The industry must produce a successfully marketed product or service which meets the delivery schedules and quality standards of its customers. The product or service must generate adequate revenue to cover all production costs. A list of 14 initial business objectives for such a competitive industry are as follows:

1. It has been determined that both rehabilitation and habilitation are most effective if they occur within the general area in which a person will function vocationally (3). Therefore, the most logical objective to rehabilitation in gereral is normal, competitive employment.

2. The primary business objective is to acquire a long-term production contract. By long-term it is meant that the product or service must have a guaranteed life of at least three years and require full-time employment of at least 10 persons.

3. The facilities layout or production planning team must include a good cross-section of persons representing those who will be working in the industry (3).

4. A very important preliminary step is to identify surrounding existing rehabilitation programs and neighborhood competitive businesses in order to eliminate potential competition and to establish cooperative working relationships (3).

5. The location of the industrial facility should be in the normal mainstream of society (3), such as in an industrial park. It must be reasonably close to a public or specialized transportation route, unless transportation will be provided by the industry.

6. If building a new facility, it will be necessary to investigate the typical items of concern such as site zoning, code compliance requirements, terrain irregularities for drainage, etc. If moving into a previously owned facility, compatibility with specific layout requirements must be ensured (3).

7. The general steps to the start-up of an industry should be taken in the following order:

a. Locate a long-term production contract.
b. Determine the physical requirements for the production facility.
c. Establish an estimate of the production workforce, including both management or supervisory staff and direct labor personnel.
d. Determine all the necessary support functions and estimate adequate space. The support functions are sales (marketing), accounting, personnel, engineering (including a machine/fabrication shop), maintenance, quality control, aide and attendant care, etc.
e. Determine the amount and sources of initial capital for the facilities, equipment, staffing, and daily operation. Capital can be acquired via investments, grants, and loans and must be sufficient to meet all expenses for the industry for at least 18 months.
f. Do not forget to include in your financial planning all requirements for expansion for the first 18 months. At least three production contracts must be planned for.
g. Before "groundbreaking" takes place, 75 percent of the funding should be available and plans should be made for the industry to become totally self-sufficient within five years.
h. An influential advisory council of successful business persons should be established. Once the industry is incorporated, this advisory council can become the board of directors.

8. Once the facility has been built and hiring is underway, consider the following hints regarding the future of the industry (3):

a. Look to Vocational Rehabilitation, PWI, and local sheltered workshops for qualified disabled job applicants.
b. Keep the name and production capabilities of the industry before the business public through direct mailouts of brochures, media, trade and civic organizations, etc. This is a relatively simple and inexpensive marketing technique to improve the probability of procuring contracts, but be careful not to exploit the handicapped workers.
c. Hire persons only when jobs are available. Do not create non-productive work for the sole purpose of occupying persons who would otherwise be unemployed.
d. Keep posted on current legislation regarding employment incentives and disincentives.

9. Some of the special adaptations that might be needed to assist disabled persons are (4):

a. Widening access, e.g. doors, aisleways.
b. Labelling in braille and removing obstacles for visually impaired employees.
c. Raising or lowering equipment and tables.
d. Moving or altering equipment controls.
e. Providing jigs and holding fixtures.
f. Eliminating architectural barriers via ramps, special parking, automatic doors, non-skid surfaces, accessible drinking fountains, restrooms, etc.
g. Converting complex paperwork to simplified standard checklists.
h. Insuring that hearing impaired personnel be given clearer visability of facilities, equipment, etc.

10. There are a few generally quotable facts which should help supply useful information as well as positive motivation to competitive employment of handicapped persons:

a. "...automation and technology have greatly altered requirements for many jobs heretofore closed to the handicapped (5)."
b. "It has been carefully calculated that for each dollar spent by the Federal Government in its joint programs with the states, the rehabilitated man or woman of today will pay back at least five dollars in Federal income taxes alone during the remainder of his (or her) work life; this is, of course, in addition to state, municipal, or other taxes. There is no charity in vocational rehabilitation (5)!"
c. The Lockheed Missiles and Space Company, Education and Training Department, discovered a "high level of manual dexterity among deaf employees (5)."
d. The U.S. Department of Labor says, "Impaired persons have fewer disabling injuries than unimpaired ones when exposed to the same work hazards (5)."
e. "No employer, unless subsidized, could stay in business if hired handicapped just because they were disabled (5)."
f. "Handicapped people can fill jobs which will release some more mobile person for a post the disabled can't fill (5)."

11. Financial and management assistance can be supplied by the Small Business Administration in the form of Handicapped Assistance Loans, financial counseling, government contracts, publications,

management assistance in the form of large firms being asked to assist small businesses, Service Corps of Retired Executives (SCORE), etc.(5)(6)(7)

12. Some of the disadvantages of disability are compensated for by special government support which actually can create significant disincentives to gainful employment (8).

a. A determination was made that work became a definite disincentive if the take home pay was more than $420.60/month. At that level SSI and Medicaid benefits were terminated.

b. There were definite set backs observed in total disposable income at annual pay levels of $3300 and $9300.

c. There are over 80 public programs for disabled persons in the U.S. falling within three basic categories: 1) state and local service programs, 2) income support programs, and 3) income maintenance programs.

13. Each person hired should be provided a minimum of three productive workstations. The industry remains at the mercy of the customer, therefore, if a contract is terminated or delayed for any reason, the employee must be capable of working alternate jobs. Productivity is not complete without versatility.

14. Last but certainly not least are 6 items to consider which have been taken from literature generally describing how to succeed in business, regardless of the abilities or disabilities of the workforce (9)(10).

a. A competitive manager knows how to arrive at "a fair day's pay" and how to provide non-financial incentives. The employees must be paid minimum wage and above, and provided a full fringe benefits package with vacations and paid holidays, etc.

b. Management at all levels should expect, and even demand, improvement, but not perfection.

c. Marketing must be a key function in any competitive business since "marketing grows in importance as an industry matures (11)." Both "under-marketing" and "over-marketing" must be avoided.

d. Most businesses fail because of either management errors, slumping economy or aggressive competition. Plans must be made regarding how the industry will avoid or be able to overcome all three.

e. Strategic allocation of duties and responsibilities is critical. If the chain of command is unclear, the industry will have the kinds of personnel problems most difficult to solve (9).

f. A final quote in this section emphasizes the primary importance of identifying and securing a successful production contract. "First, of course, there must be a genuine business opportunity, for if there is none, no amount of managerial talent (or capital) can overcome this lack (6)."

CONCLUSIONS

The ideal solution to unemployment of disabled persons is equal employment opportunities in existing competitive jobs. Unfortunately, there is a very short supply of occupations fully equipped to employ disabled persons, especially those who are considered "severely disabled". Sheltered workshops, in most cases, are legitimate attempts to provide vocational training but are often non cost effective. The optimum solution appears to be the establishment of a competitive industry where disabled persons work alongside those considered able-bodied. The competitive industry must be able to produce a successfully marketed product or service which meets the delivery schedules and quality standards of its customers. The product or service must generate adequate revenue to cover all production costs.

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AN INVESTIGATION OF THE VALUE OF VOCATIONAL REHABILITATION ENGINEERING IN THE SHELTERED EMPLOYMENT ENVIRONMENT

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ABSTRACT

The Rehabilitation Engineering Center in Wichita, Kansas, has, as its emphasis, research in the area of finding means to employ disabled persons using technology. A project to investigate the value of vocational rehabilitation engineering assistance to sheltered workshops has been underway for three years. Three sheltered workshops in the state of Kansas were chosen for demonstration projects. The work that was done at one of these workshops is presented here as a case study. Work station modifications and economic benefits are discussed

INTRODUCTION

The Rehabilitation Engineering Center (R.E.C.) in Wichita is federally funded by the National Institute of Handicapped Research, is under the sponsorship of the Cerebral Palsy Research Foundation of Kansas, Inc., and has a cooperative relationship with the College of Engineering at Wichita State University. One of the 12 research projects that the R.E.C. is currently investigating is the value of vocational rehabilitation engineering support in the sheltered employment environment. The R.E.C. in Wichita has a 10-years' history of being successful in the modification of workstations and employment of severely physically disabled persons in non-sheltered employment (principally at Center Industries Corporation in Wichita) and has had occasion to do work on a fee-for-service basis for various sheltered workshops in the state of Kansas. This paper presents a case study of one of three workshops that were chosen as demonstration projects for the federally funded research.

BACKGROUND

The workshop chosen as a case study for this paper is the Big Lakes Developmental Center, Inc., in Manhattan, Kansas. Three workstations were modified.

METHODS

Workstations to be modified were chosen on the basis of the legitimacy of the application of industrial modification techniques to enhance the disabled worker's ability to operate the machines, and, to improve productivity through specialized tooling (adaptations).

The three workstations that were modified are: 1. a large drill press which has a gang drill head for drilling several holes at once (see Figure 1); 2. a band saw that is used to cut plywood in circular disks (see Figure 3); and, 3. a radial arm saw. Following are descriptions of the modifications.

Drill Press

A large drill press was used to drill six holes simultaneously in pieces of plywood that are either 16 inches, or 21 inches, square and is shown in Figure 1. This is accomplished through the use of a device called a gang drill head which holds the six drill bits and rotates all of them for drilling. Two obstacles kept the desired worker from operating this machine. First, he could not reach the "start" and "stop" controls on the machine. Second, good judgement was required during the drilling process to insure good clean holes were the result.

The first obstacle was overcome by moving the control push buttons to the bench top for easier access (good for all operators). See lower right hand corner of Figure 1. To insure a good pass of the drills through the plywood yielding clean holes was a greater problem to overcome. Due to the differing sizes of the holes to be drilled, one drilling rotation speed could not be found that would accommodate all of the drill bits. A different rotating speed is required for small holes (faster) than for larger ones (slower). Staff members and high functioning clients could operate the machine

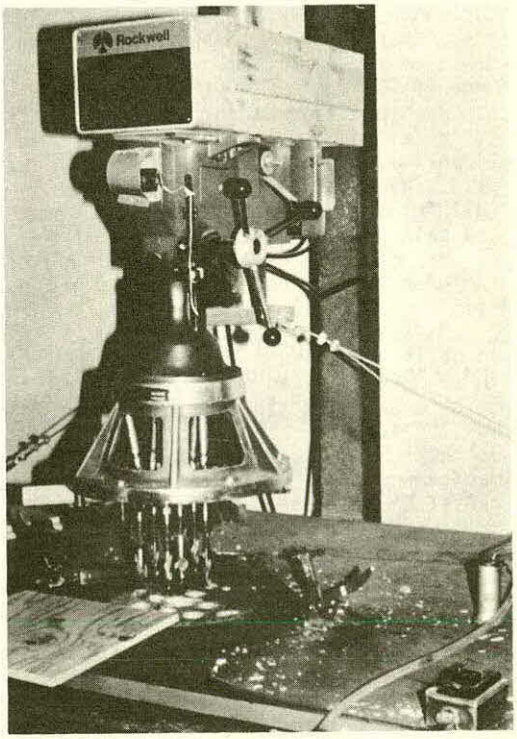

FIGURE 1: DRILL PRESS

by using proper caution and "feel" as the drill bits passed through the material. The solution to this problem was to design and fabricate a special cutting tool to drill the largest hole which was the most difficult to drill consistently. This tool was not constructed like a drill bit, but more like a circle cutting tool. See Figure 2, second

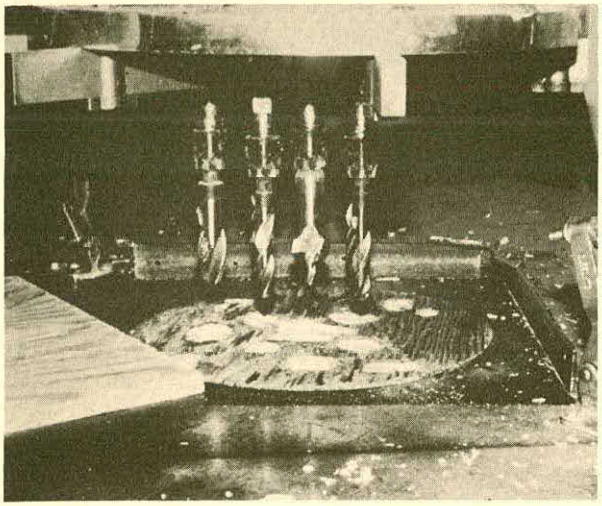

FIGURE 2: GANG DRILL HEAD

"drill" from the right, in the group of drills.

Band Saw

The band saw shown in Figure 3 is used to cut the square pieces of plywood mentioned above in circular disks. Shop personnel had installed a device on the table of the saw to act as a spindle around which to rotate the workpiece in the sawing process. That device did not permit changing the saw blade without removal of the device from the saw. That prohibited clients from changing the blades. The spindle was on a slide mechanism that did not have accurate locating stops. This caused frequent breakage of saw blades due to misalignment. A new fixture was designed, fabricated, and installed which provided for accurate locating of the workpiece and the changing of blades without removal of the device from the machine. Figure 3 shows the moveable spindle retracted, ready to receive the next workpiece. Figure 4 shows the work piece as it completes the cutting process.

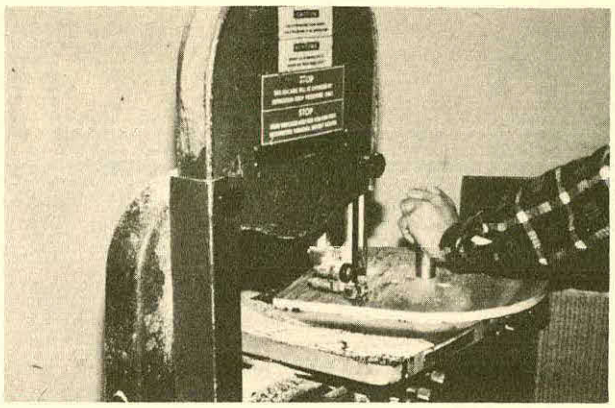

FIGURE 4: WORKPIECE AT END OF CUT

Radial Arm Saw

Various pieces of lumber are cut to length on the radial arm saw shown in Figure 5. A client having the use only one arm had difficulty operating the saw, while, at the same time, holding the workpiece in place. A supply of air was close at hand, so a pneumatic clamping system was designed, fabricated, and installed on the bed of the saw. Figure 6 shows the two clamping cylinders activated, holding a typical workpiece. See the lower right hand corner of Figure 5 for the handle on the air valve which activates the clamping cylinders. The cylinders can be operated both at a time, or individually for shorter workpieces.

ECONOMICS

Costs

The costs that are presented here are based on the assumption that the individual workshop would be paying nominal rates for engineering time, fabrication (shop) time, materials, and travel. Engineering is calculated at $50.00/hour, fabrication at $25.00/hour, and travel at 21¢/mile.

Travel. Three trips were made to Manhattan to ac-

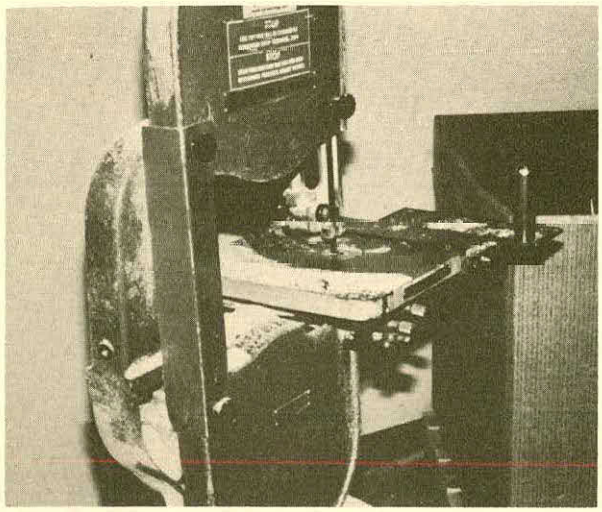

FIGURE 3: BAND SAW WITH SLIDING SPINDLE

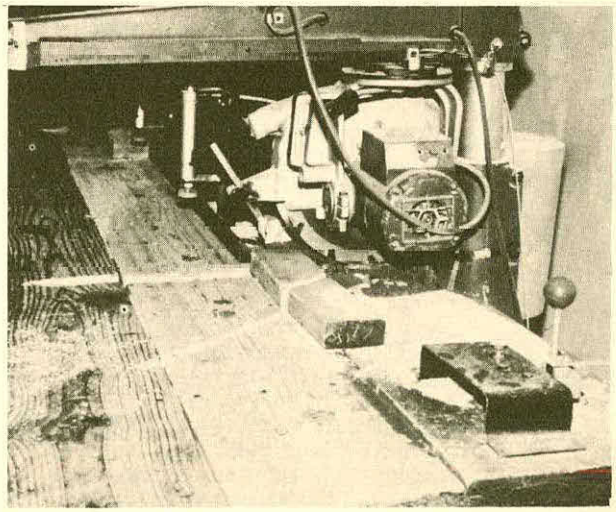

FIGURE 5: RADIAL ARM SAW

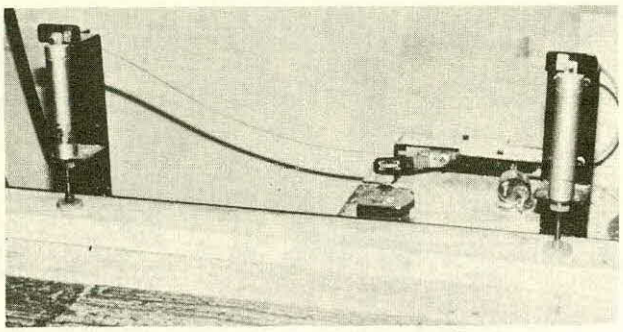

FIGURE 6: PNEUMATIC CLAMPS

complish the modifications. A total of 840 miles were driven at a cost of $176.40.

Drill Press. Eight hours of engineering design time was spent ($400.00). Twenty hours of technician time was required for fabrication, and four hours were spent in the installation ($600.00). Total cost for the drill press modification was $1,050.00 including material costs of approximately $50.00 (excluding travel).

Band Saw. Sixteen hours of engineering ($800.00), sixteen hours of shop fabrication ($400.00), and four hours of installation time ($100.00), were required excluding travel. Material costs were less than $25.00.

Radial Arm Saw. Twelve hours of engineering ($600.00), twenty-four hours of shop fabrication ($600.00), and four hours of installation time ($100.00) were required excluding travel. Material costs were approximately $200.00.

Total cost, then, to the workshop for all of the work that was done would have been $4,051.40.

Benefits

Drill Press. The subject client was not able to operate this machine prior to the modification. His production rate was an immediate 64% of normal (acceptable rate for the shop is 50% of normal). Therefore, the client's income from operating this machine went from "zero" to $2.18/hour. The improved large drill bit allowed the shop to go from a change to a new bit every three days to not having to change the bit one time in the two years that it has been in use. Assuming that there would have been over 150 changes in the two years at a cost of $15.00/change, $2,250.00 has been saved.

Band Saw. Clients are now able to change the saw blade (once every three days). Time required for blade change is five minutes (compared to 60 minutes before, done by staff, once each day, with 7-10 client man-hours lost during each blade change). Accepted rate of production is 31% of normal ($1.05/hour). Saving is estimated to be $50.00/week client earnings not counting staff involvement saved.

Radial Arm Saw. Subject client's production rate increased from 26% of normal to 39% of normal and operation of the machine is now done in a safe manner. Safety is a difficult quantity to measure economically. The increased income for the client was $.44/hour. A considerable reduction of staff involvement with this client was noted, but economic benefits were not charted.

DISCUSSION

Monetary benefits speak for themselves. Savings in staff effort and concern are difficult to measure. The author has learned that perhaps the greatest benefit to this workshop has been that staff are free to do supervisory work in relation to the clients while the clients are doing the production work (some of which had to be done by staff, themselves).

Space limitations prevent discussion in depth. A full report of his research project may be obtained from the author approximately May 1, 1986.

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A MICROCOMPUTER WORKSTATION FOR ENHANCING FUNCTIONAL RECOVERY

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ABSTRACT

A prototype microprocessor-based rehabilitation system that can be used to enhance functional recovery has been developed. The system is a comprehensive workstation in which the patient exercises a variety of motor and cognitive skills to help restore the basic processes that underlie purposeful behavior.

The system has three major components. First, a number of input sensors serve to enlarge the range of skills that can be addressed for training. Second, a set of training tasks based on modern theories of recovery is implemented. Third, a special software structure allows the rehabilitation professional, with no knowledge of programming, to generate new training tasks involving the available sensors. The resulting Rehabilitation Workstation is within the range of individual affordability.

INTRODUCTION

The patient who has suffered an impairment of the central nervous system is a unique individual, characterized by his own particular set of deficits and strengths. Any combination of motoric and cognitive skills may have been affected, and hence a great variety of strategies and techniques is required to facilitate functional recovery. To regain the ability to interact with the environment, especially in complex or novel situations, the impaired person must reestablish fundamental motor programs which underlie activities such as eye-hand coordination and oculo-motor skills, as well as basic cognitive-perceptual processes such as attending, discriminating, and planning.

Current microcomputer technology offers the potential for power, flexibility, and universality in dealing with basic motor and cognitive skills. The microcomputer can present systematic and modifiable rehabilitation tasks, and can configure these tasks for varying degrees of complexity; it can recognize and interact with widely divergent forms of human communication; it can collect, record, and analyze sophisticated temporal and spatial responses -- and it can accomplish all of this with a level of consistency and precision not otherwise possible.

We have designed and implemented a microcomputer-based rehabilitation system, called the Rehabilitation Workstation, for enhancing functional recovery. The Workstation consists of a standard microcomputer equipped with a variety of input modalities. These sensors serve the dual purpose of enlarging the set of motor and cognitive skills that can be addressed for training, as well as making the system accessible to patients requiring specific communication tools. Software has been implemented to present therapy tasks involving these input modalities independently and in combination. Thus, skills can be selectively exercised, and different levels of complexity can be chosen for different functions. In addition to the existing package of therapy tasks, the Workstation contains a software application generator to allow the rehabilitation professional to create specialized sets of training programs to meet the specific requirements of each patient. With this capability, the clinician can maintain his own individuality of style, and the tool can be customized to each patient's particular needs.

BACKGROUND

Initial uses of microcomputers for enhancing functional recovery have been very encouraging. Numerous systems exist which contain original software for addressing cognitive processes from simple foundational skills to more complicated tasks (1,2). In most cases, clinical testing has demonstrated that patients adapt well to these computerized tools. The Rehabilitation Workstation has built upon the research underlying the existing systems with the goal of extending such systems to emphasize the motor aspects of rehabilitation.

MATERIALS AND METHODS

Input sensors

The computer selected for this work is the Radio Shack Color Computer. The sensors have been designed and built with two goals in mind. First, each must be low-cost; second, each must be easy for the non-technical person to use.

The analog-to-digital conversion (ADC) on the Color Computer is limited to 6 bits of resolution. Since most sensors require 8 bit resolution, we built a new ADC and mounted it inside the standard disk interface case, thus developing a complete package that can easily be plugged into the cartridge port. In addition, the analog input lines can be brought out of the other side of the case allowing for the modular interconnection of sensors to the device. Our goal in reconstructing the ADC was to provide a "universal" interface to all of the sensors, and thus eliminate a significant amount of additional circuitry.

The eye sensors we developed are modified safety glasses without lenses. A printed circuit board is cut to fit in place of the lower portion of the lens. A modulated infrared light-emitting diode (LED) is mounted between two infrared phototransistors. The LED is horizontally in

line with the pupil and is slightly lower vertically. In addition to the modulator chip, a differential amplifier chip includes external controls for adjusting both the gain and the zero point. Sufficient gain exists so that full scale deflection is possible when viewing a 19 inch screen at a distance of 18 inches. No electrical connection to the subject occurs. The result is a smooth change in voltage output as the eyes move from left to right.

The pressure transducers that we have built use a driven shield approach. By driving the shield, we detect only changes in capacitance due to the pressure applied, and eliminate extraneous effects. The transducers are made out of copper clad mylar with separating soft sponge material. They range in size from that of a quarter to the size of a chair seat. The shape can be flat, or the pads can be wrapped around objects for grasp. In all cases, the transducers have been found to be linear over a range of up to 150 pounds.

Our touch plate electronics is battery powered and optically isolated from the circuit being controlled. This approach requires a pair of touch plates so that skin contact is used to "bridge" the plates and close the circuit. The touch plate surfaces are formed from single-sided printed circuit board that can be easily machined into any desired geometrical shape.

We evaluated several commercial touch screens, and developed a number of touch screen electronics packages. Initially, we were not satisfied with the repeatability of the results. The problem was determined to be the initial finger bounce as the sheets come in contact, resulting in an erroneous initial reading of the X,Y pair. By sampling the screen resistance value, we are now able to take accurate and repeatable measurements. The unit mounts to a standard 19 inch television screen with velcro pads or can be positioned inside the cabinet.

We developed a voice sensor consisting of a two-chip device with a low cut-off at 275 Hertz and a high cut-off at 4 kHz. Two sounds activate the voice unit - a "huh" and a "siss". These sounds were chosen to maximize the distinction between high and low utterances, and because they are easy to form, even for subjects unable to enunciate clearly. To add another dimension to the sounds, the software can discriminate between volume levels.

Therapy tasks

The therapy tasks that have been developed for the Workstation are rooted in a model of rehabilitation based on modern theories of recovery and goal-oriented behavior (3). The fundamental rehabilitation issue is the manner in which the therapeutic process can exploit the ability of the central nervous system to elicit long-term readjustments or modifications in functional performance. The model proposes a reeducational program which embodies two basic elements: (1) an organized sequential motor plan; and (2) a sensory information component.

The motor organization concept suggests that sub-functions should be trained in serial order, commencing with midline stability including both ocular and postural stabilization, proceeding to interactions responsible for translation between stability and initiation of movement, and concluding with guided limb movements. Several Workstation tasks have been developed for each of these categories.

The sensory information concept is based on the observation that the central nervous system exhibits remarkable plasticity of function. There is little doubt that functional recovery may be substantially aided by non-specific factors, i.e., sensory experiences encountered during daily life may lead to the employment of tactics which further augment naturally evolving plastic changes. Yet, to profit from the potential of the central nervous system to repair a dysfunctional motor system substantially, specific factors most likely need to be employed, i.e., regulated feedback including both knowledge of results and knowledge of performance (4). The Workstation is ideally suited to monitoring the characteristics of the behavior and feeding back the error and quality of the performance. In each of the therapy tasks, one or more of the input sensors is used to measure the actual performance of the patient; the system then translates this into meaningful information for the patient, and uses visual, auditory, or tactile means to attempt to induce improvement in ongoing or future performance.

Application generator

An application generator is a software structure that converts problem specifications into computer programs in a particular application area. This enables professionals outside of computer science to develop their own software without learning to program (5). The inclusion of an application generator in the Workstation was motivated primarily by the perceived need for therapists to customize or change tasks.

The formal nature of the application generator required a precise and rigorous definition of therapy task. This resulted in identifying each task by means of five components: (1) the attributes of the stimulus, e.g., visual, stationary; (2) the attributes of the response, e.g., the sensor involved, a description of a correct response; (3) the characteristics of the feedback, e.g., auditory, given during or at the end of the session; (4) automated session controls, e.g., the means by which the difficulty of the task is changed, the criteria for modifying the difficulty of an ongoing task or for proceeding to a new level of difficulty; (5) statistics, e.g., which data to collect, what analysis routines to use. The user specifies this information to the system, and it in turn generates a program which is an original therapy task. The application generator can

recognize inconsistencies, ambiguities, and incomplete specifications and prompts the user for further information, and can also make assumptions by consulting its knowledge base.

The application generator has proved to be the most difficult aspect of the Workstation, and to date has been only a partial success.

RESULTS

The Rehabilitation Workstation is currently being evaluated under several experimental paradigms to assess its effectiveness as a therapeutic instrument. Each of the paradigms addresses three fundamental questions: (1) does systematic training with the Workstation, using its fundamental features of practice, feedback, and motivation, lead to improved performance on the associated therapy tasks; (2) does such training lead to improvement in functional performance indices, e.g., Activities of Daily Living Skills; (3) does such training bring about adaptation of enhanced performance on either the therapy tasks or the functional tasks. The experiments employ two modes of operation. Training consists of practice on specific therapy tasks implemented on the Workstation, with feedback and motivational elements. Testing consists of execution of the therapy tasks, with no indication given of the quality of performance and no motivational encouragement, together with a measurement of functional performance indices.

Some examples of the goals of ongoing protocols are: (1) improved guided limb motion in the hemiplegic arm of stroke patients; (2) improved pointing and gaze control in a severely involved geriatric population; (3) improved postural stability in a spinal cord injured population; (4) motion perception and control manipulation in cerebral palsy children who are wheelchair candidates; (5) improvement of grasp, isolated control, and finger extension in patients with hand injuries. Quantitative results are not yet available from these protocols, but preliminary anecdotal comments are encouraging. Therapists' acceptance of the Workstation is very positive, and many patients have reported a sense of improvement.

DISCUSSION

The Rehabilitation Workstation enables standardization and objectivity in the presentation and evaluation of therapy, and can detect and respond to significant occurrences in the patient's performance. Moreover, it is a motivational device which provokes attention and carries a certain fascination and allurement.

At each step in the research, the prevailing philosophy was to seek low-cost alternatives. This has resulted in the following design strategy. The base system consists of a low-cost microcomputer that contains all of the existing tasks and all of the capabilities required to create new tasks. Further, all feedback associated with each task initiates from the base system. From an engineering viewpoint, then, the sensors can all be considered optional and interchangeable peripherals. This means that the same therapy task, which exists autonomously in the base system, can be executed by using any one of a number of different sensors -- the tasks are generic and sensor-independent. For example, a task that involves moving an object on the screen to coincide with a second object on the screen can be performed with eye movements, by placing a finger directly on the first object and pulling it across the screen, or by squeezing a pressure pad with the torso. This sensor substitution can be made with no modification to the task software and with no overt action required by the user.

An operable system can be configured for an individual patient, consisting of precisely what is required by his particular needs, and need not include expensive features that he may never use. This makes it possible for the Workstation to be used as a home-based system under the direction of a rehabilitation professional. Thus, a limited supply of therapists may be able to serve a wider group of patients by allowing a normal patient/therapist session to be expanded into many times that amount of self-directed or family-directed therapy.

ACKNOWLEDGMENTS

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A TRACHEOSTOMY FILTER DEVICE

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ABSTRACT

A filtering device has been developed to enable individuals with tracheostomies to return to work in unclean air environments. Contaminated air which would directly enter the lungs and cause irritation and disease can now be filtered of dusts, mists and fumes, warmed to near body temperature and humidified.

INTRODUCTION

Vocational Rehabilitation serves many clients who have had laryngectomies. Without the larynx, which acts as an airway between the pharynx and trachea, an individual cannot breathe through his nose or mouth. To restore breathing, a tracheostomy is performed in which the trachea is sutured to the skin of the neck to provide a permanent opening to the trachea through the neck. This opening is called the stoma.

As air passes through the nose, three distinct functions are performed by the nasal cavities:

First, the air is warmed close to body temperature by the extensive surfaces of the turbinates and septum.

Second, the air is almost completely humidified even before it passes beyond the nose.

Third, the air is filtered so that almost no particles larger than 4 or 6 microns in diameter enter the lung through the nose.

When a person breathes air through a tracheostomy, the cooling and especially the drying effect in the lower lung can lead to serious lung crusting and infection (1).

Tracheostomy patients generally wear a piece of gauze over their stoma to prevent large particles from entering their lungs. Many vocational rehabilitation clients cannot return to work in their previous occupations because the gauze does not offer sufficient protection from lung irritants encountered in their work environments such as welding and paint fumes, and construction dusts. For years vocational rehabilitation has retrained these individuals for jobs that would place them in clean air environments.

The filter device described in this paper is designed for a laryngectomy client who wants to return to work as a construction site foreman. There are four major design criteria for his system:

One, it has to filter construction dusts, mists and metal fumes.

Two, the device must warm the air entering his lungs, especially in the winter months.

Three, the system has to humidify the air to ensure comfort and reduce the coughing fits common in dry lung conditions.

Four, the device must pull apart quickly to allow for easy cleaning and to allow the client to cough and clear his lungs of fluid and foreign matter.

MATERIALS AND METHODS

The filter pack design (figure 1) is patterned after OSHA approved gas masks which protect nose/mouth breathing workers from toxic industrial hazards. Such gas masks typically consist of a rubber supportive and sealing structure which holds filtering materials in two retainers, and an exhaust valve. The use of two retainers maximizes the surface area of the filter material available for air intake, thereby minimizing breathing resistance. The retainers each have inhalation valves to prevent exhausted air from entering the filter material.

Fig. 1. Tracheostomy Filter Device

FILTER DEVICE

The supportive structure of the tracheostomy filter device is of thermoformed 0.093" ABS plastic. Filtering materials from Willson's 1200 series dust/mist/fume respirators, consist of 1/2" fiberglass and 5/32" felt. The filter material retainer is a polypropylene mesh canvas (commonly used in needlepoint) which holds three layers of the fiberglass and one layer of the felt. The air passes first through the fiberglass and then through the felt. The surface area of the filter material exposed to the air is 20 square inches. The retainer holds the filtering materials in place over a plenum or air space.

Filtered air enters the plenum and then travels up to the humidification chamber through fourteen slots, providing 0.66 square inches of access. A 0.030" gum rubber inhalation flap valve, glued above the slots inside the humidification chamber (figure 2), allows passage of air in one direction only. The humidification system and the filtration system are in two separate compartments, preventing moisture from getting into the filter material.

Polypropylene fibers[1] soaked in distilled water are used as a wicking material in the humidification chamber. Polypropylene fibers were chosen for their chemical inertness, excellent mildew resistance, and low moisture regain properties. A swivel L-tube connector bridges the humidification chamber and the 15/16" I.D. respiratory tubing which leads to the stoma. A tube screen (figure 3) separates the polypropylene fibers from the L-tube connector, preventing the fibers from entering the air passageway. The same plastic mesh used for the filter retainer, surrounded by nylon silkscreen material, comprises the tube screen.

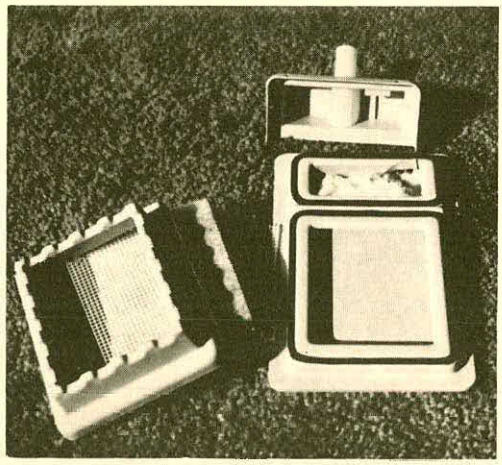

Fig. 2. Filter Device Components (Inhalation flap valve is taped open, exposing three of the fourteen air entry slots)

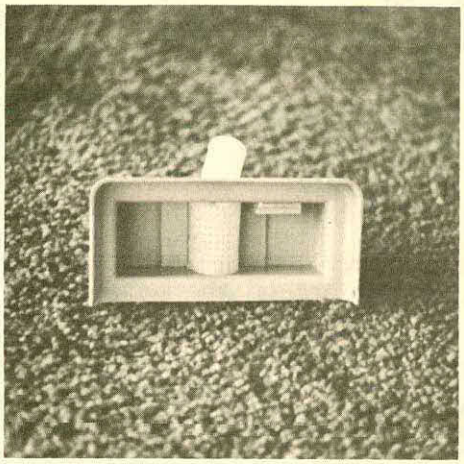

Fig. 3. Tube Screen

A T-tube connects the respiratory tubing to the stoma connector and to an exhaust valve commonly used in external respiratory systems[2] (figure 4). The stoma connector is an adhesive-backed flanged plastic connector obtained from commercially available tracheostomy shower guard products[3].

Expanded 3/16" neoprene seals glued in place to the outer surfaces of the filter plenum and the humidification chamber allow the system to seal yet be easily serviced. The ABS plastic covers for the filter and humidification sections are secured to the back of the device by eight #6-32 x 3/8" socket pan head screws. A 1/4" NPT brass plug allows quick access to the humidification chamber for addition of distilled water.

Fig. 4. Connection to Stoma

FILTER DEVICE

The respiratory tubing disconnects easily from the swivel L-tube connector and from the T-tube connector. The T-tube easily pulls away from the stoma connector allowing prompt clearing of lung congestion.

The simplest way to warm the air entering the lungs is with body heat. The 10"L x 7"W x 2"H filter device is fastened comfortably to the chest with 3/4" cotton strapping. The eighteen rectangular plastic projections surrounding the front edges of the filter chamber allow air to flow into the system from the sides when clothing is worn over the device. In wintertime a coat worn over the tracheostomy filter pack is sufficient to warm air entering the lungs to near body temperature. Breathing resistance is only slightly increased.

DISCUSSION

The tracheostomy filter pack is versatile. Other types of filtering material can be used including those that filter radionuclides, organic vapors and pesticides. The humidity in the humidification chamber can be controlled by the amount of fiber and distilled water in the system. The device offers minimal breathing resistance and is comfortable to wear.

The filter pack allows a tracheostomy client to return to his field of work with no need for expensive retraining and years lost from the work force. The device may lead to an overall improvement of the client's health and well-being.

FOOTNOTES

1. 3 den 2" Merge G14H78 polypropylene staple, Phillips Fibers Corporation.
2. Exhalation Mouthpiece, Calculair.
3. Stomaguard I Shower Shield, Medmart.

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AUDIOBIOFEEDBACK FOR TASKS

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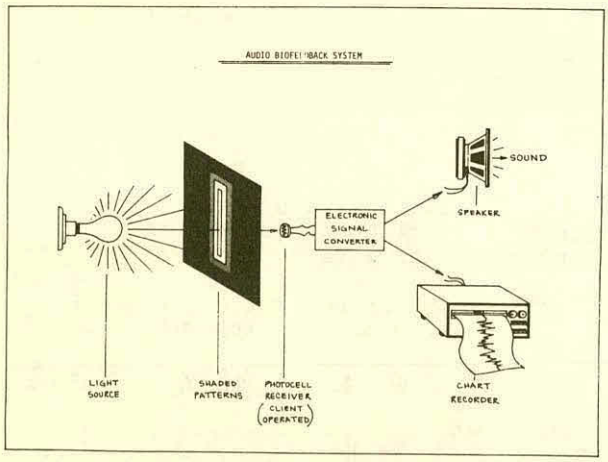
ABSTRACT

The purpose of this project is to explore the use of audio-biofeedback (ABF) as an augmentative sensory system to assist handicapped persons in motions and positions associated with tasks performed in independent living and vocational pursuits. This paper reports the results of tests using a simulated drill press equipped with an ABF system. Twenty (20) able-bodied and 20 handicapped persons were tested. The results show that performance was substantially improved using ABF.

INTRODUCTION

The purpose of this project is to explore the use of audio-biofeedback (ABF) as an augmentative sensory system to assist handicapped persons in motions and positions associated with tasks performed in independent living and vocational pursuits. A schematic view of the ABF system is shown in Figure 1. This figure shows that the amount of light transmitted through the shaded patterns is transformed into an audio signal. The pitch of the audio signal indicates whether or not the operator is on target and, if off target, gives an indication of the distance away. The signal is also relayed to an automatic chart recorder which produces a permanent record of performance.

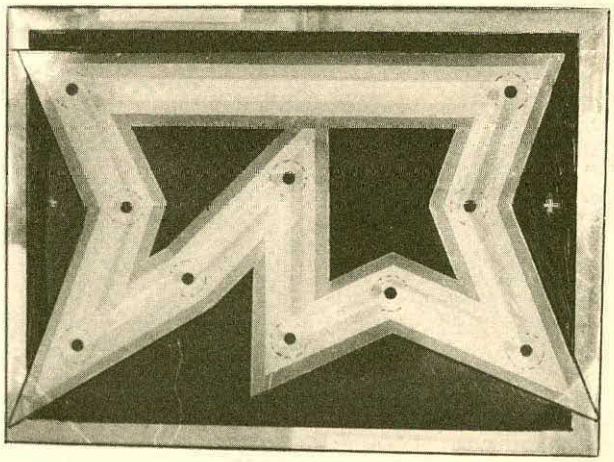


Schematic View of ABF System
Figure 1

This project is similar to conventional biofeedback procedures (1,2), but it differs in that the purpose is to improve task performance. This work is an extension of work using a similar ABF system with a light table (3).

EQUIPMENT AND METHODOLOGY

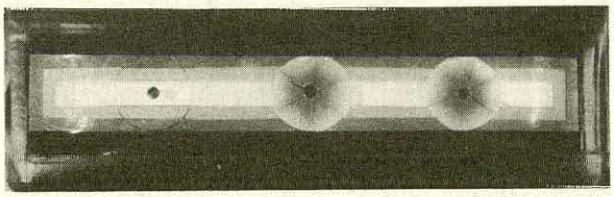
The test equipment consisted of a simulated drill press and shaded workpiece patterns which the operator moved from one hole location to another. The light source is located beneath the work surface and the light is directed through the shaded patterns to a photocell which replaces the drill bit. The two shaded patterns tested are shown in Figures 2 and 3. The first workpiece pattern (Figure 2) was shaded to indicate 10 hole locations and the most direct path between holes.



Drill Press Workpiece
Figure 2

Additional shading indicated the distance from the optimal path. The second workpiece (Figure 3) concentrated on three methods of guidance (by tonal feedback) to the hole location. The first hole location was identical to those used in the drill workpiece pattern and consisted of a black dot indicating the hole location. The second hole location was a black dot, surrounded by a circular pattern, which was lighter at the circumference and gradually darkened toward the center giving tonal guidance to the

hole location. The third hole location was shaded like the second hole but, additionally, utilized a separate sound activated by an electrical contact at the precise drill location.



Hole Location Workpiece
Figure 3

Two groups of subjects were tested, each consisting of 20 adults. The first group consisted of handicapped persons in the age range of 21-51 years with a median age of 29. The second group was comprised of able-bodied adults with a median age of 42. Education levels of the handicapped group consisted of 14 individuals with high school diplomas, while the education of the remaining six individuals was less than high school. The able-bodied group education levels consisted of five with high school diplomas, four college graduates and 11 with postgraduate degrees. The handicapping conditions were as follows: spina bifida (1), cerebral palsy (11), head trauma (2), CVA (2), spinal cord injury (1), and others (3).

Data was collected on accuracy of hole location and on time of performance. In general, time was a secondary consideration for these tests. However, one set of data was taken where the operator was instructed that time was an important factor. Test data was gathered with and without ABF for initial baseline data and after a short familiarization period. A final test was performed with an emphasis on time.

RESULTS

Table 1 presents a ratio comparison of accuracy from the simulated drill press pattern (Figure 2) tests with and without ABF. The values are derived by dividing the total number of hits with ABF by the number of hits without ABF. Thus, a ratio value greater than one shows the improvement achieved with ABF. The scores show substantial improvement with ABF, however, there is a difference in the improvement ratios between the two groups. The able-bodied group showed greater improvement in the baseline tests while the handicapped group improved more after familiarization. A possible explanation for this is that the able-bodied group showed quicker adaptability. These results show promise of this system's use as a training device as well as its installation as a permanent assist.

Ratio Comparison of Accuracy With ABF
vs. Accuracy Without ABF
(Group Totals)

GROUP	BASELINE	AFTER FAMILIARIZATION	TIME EMPHASIS
Handicapped Subjects	1.20	2.13	1.35
Able-Bodied Subjects	1.96	1.42	1.54

Drill Press Tests
Table 1

Table 2 shows accuracy comparisons made by dividing scores of handicapped performance with ABF by able-bodied performance without ABF. A ratio of one indicates equal performance. Since the ratios are near the value of one, they indicate ABF enables the handicapped group to achieve scores closely approximating the accuracy performance of the able-bodied group.

Ratio Comparison of Accuracy With ABF
Handicapped Subjects with ABF (HC w/ABF) vs.
Able-Bodied Subjects Without ABF (AB w/o ABF)

COMPARISON	BASELINE	AFTER FAMILIARIZATION	TIME EMPHASIS
HC w/ABF / AB w/o ABF	1.01	0.95	0.88

Drill Press Tests
Table 2

Table 3 shows the accuracy results of the hole location tests (Figure 3) as ratios of hits with to hits without ABF.

Ratio Comparison of Accuracy With ABF
vs. Accuracy Without ABF

Group	HOLE #1	HOLE #2	HOLE #3
Handicapped Subjects	1.00	1.89	1.88
Able-Bodied Subjects	1.44	1.33	1.43

Hole Location Tests
Table 3

The handicapped persons showed a very significant improvement at holes 2 and 3, where a gradation guidance to the hole was provided. This improvement suggests that ABF may be especially helpful to handicapped individuals when there is guidance to a precise point. As in the previous drill press results, the handicapped person's accuracy scores with ABF were approximately equal to the able-bodied scores without ABF. For both drill press workpieces, the performance times were about 20% longer for the able-bodied group and 30% longer for the handicapped group when using ABF. On all tests, performance times for the handicapped group were 2 to 3 times longer than that for the able-bodied group.

For the tests where the emphasis was placed on time, the performance times with and without ABF were approximately equal. The accuracy on these time emphasis tests was in the general range of accuracies for the other tests.

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A COMMUNICATOR FOR CONSTRUCTION AND USE IN A STATE SCHOOL

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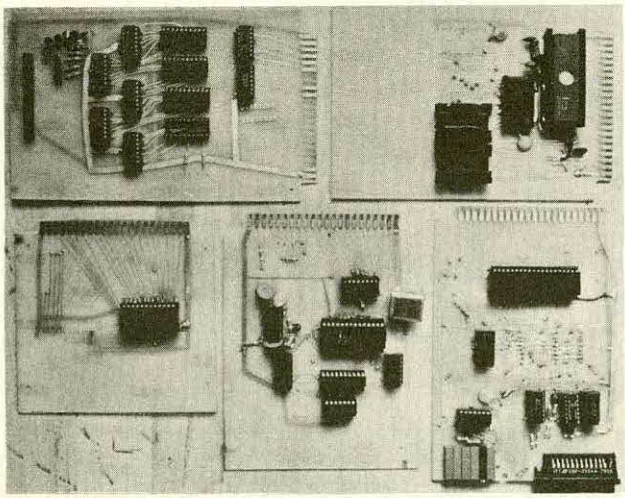
ABSTRACT

An electronic communicator built with two integrated circuits and having nine selection techniques is described. Assembly of these communicators by students at a state school for the physically handicapped and mentally retarded is also described.

BACKGROUND

The first author has built several electronic communicators. While on temporary duty at the Moss Rehabilitation Engineering Center, work was started on a row-column scanner using discrete integrated circuits. This project was not completed due to the authors return to the Department of Health Education and Welfare. The second communicator was completed at the University of Virginia as a masters project (1,2). The five circuit boards used in this prototype are shown in figure 1. Much of the complexity is due to the display driver and serial interface boards, as well as a design goal of general purpose usage.

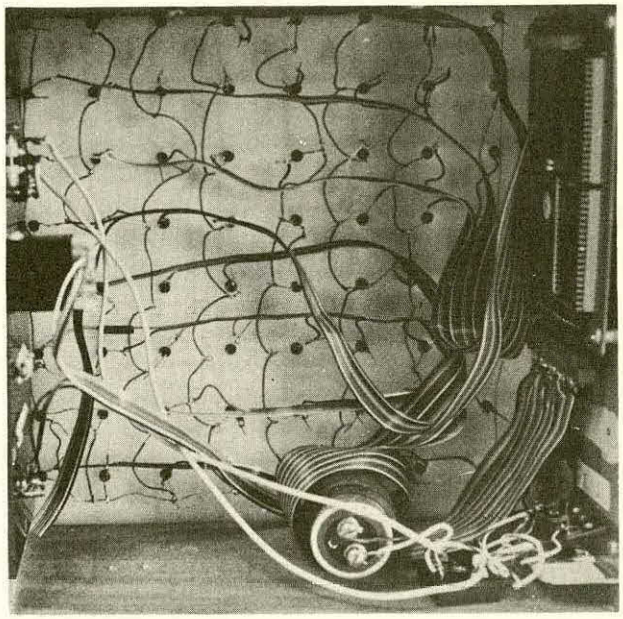


Masters project circuit boards, Figure 1

This and all subsequent communicators use an Intel 8748 microcontroller. These are single chip devices with built in memory.

This communicator was shown to a speech pathologist and occupational therapist at the Pauls Valley State School (PVSS). During this meeting arrangments were made for a visit to PVSS.

Several months later a three integrated circuit scanning communicator was developed for a client at the O'Donoghue Rehabilitation Institute. This communicator used the 8 X 8 light emmitting diode (LED) display and power supply from the communicator started at Moss. This was shown to the PVSS staff with the intent of having a state vocational school construct several units. Instead the superintendent proposed that some residents who in the past repaired bicycles and constructed leather items could assemble the communicators.

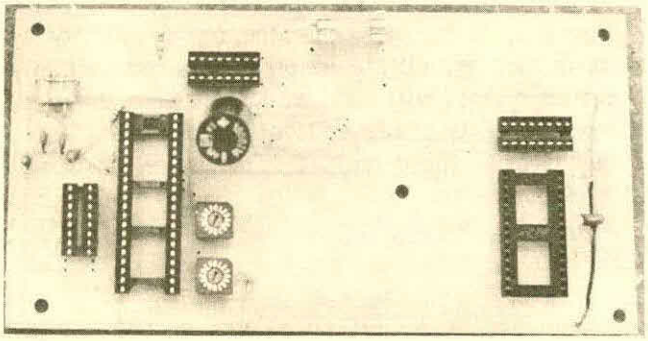


Moss/ O'Donoghue communicator, Figure 2

Two National Semiconductor 5450 display drivers were used in this design, each having 32 nonmultiplexed outputs, requiring a total of 65 wires between the circuit board and display. A rear view with removed back is shown as figure 2.

For simpler construction a fourth communicator was designed using a single printed circuit board and two ICs. The Intersil 7218 display driver used in the design multiplexes the display and was used with trepidation due to its low average drive current per LED. Use of high efficiency LEDs alleviated this concern, and allowed construction with only 16 wires between the circuit board and display.

A rechargeable battery is used for portable operation, providing an estimated 8.5 to 30 hours of continuous operation depending upon the the selection mode used. A partially populated circuit board is illustrated as figure 3.



PVSS Scanner board, Figure 3

Only the first communicator has connections for a printer or computer but this capacity could be added to the other communicators by the addition of another IC and appropiate programming.

INSTRUCTION IN ASSEMBLY AND SOLDERING

Many therapists have a fear of soldering, but the occupational therapist had previous knowledge of soldering from an adaptive switch workshop, and the speech pathologist quickly learned by soldering practice boards.

TASK ANALYSIS

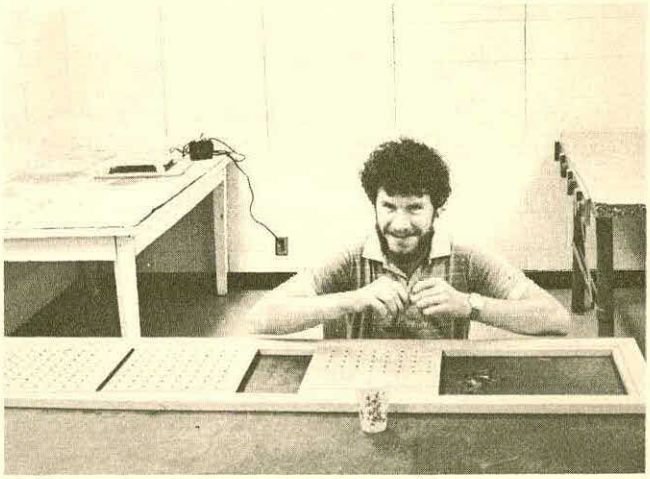
A task analysis, was performed and assembly divided into four subgroups, (1) circuit board stuffing and soldering, (2) matrix assembly, (3) case construction, (4) battery and switch cable assembly. Pictorial diagrams were developed showing how each component was to be inserted into the printed circuit board, and wiring of the matrix cable.

WORKER SELECTION

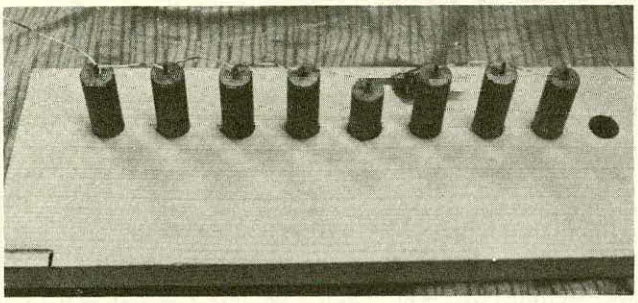
The occupational therapy area used to prepare the project has a large volume of drop in client visitors. Many of the higher level clients would observe what we were doing and tell us they had a soldering gun and had worked on their stereos. In addition they knew of friends with experience and tools. Each of the four clients selected to assemble the first ten machines had previous workshop experience. Clients chosen to complete the second ten communicators were selected on the basis of occupational therapy evaluations. Once soldering irons arrived, practice soldering sessions were initiated and the project moved to a workshop. At the same time other clients were chosen to assemble and sand the cases.

ASSEMBLY AND TEST JIGS

Many workshops are using old production technology. This has hindered productivity and caused sheltered workshops to be less competitive (3). For this project mounting the selection scheme and speed switches with the integrated circuits on a single printed circuit board greatly simplified the assembly process. Several assembly and test jigs were developed to insure proper and easy construction. The prevocational director constructed a jig holding eight display panels, easing LED insertion. Another jig was developed to form the buss wire used to connect the LEDs. These are shown as figures 4 and 5.

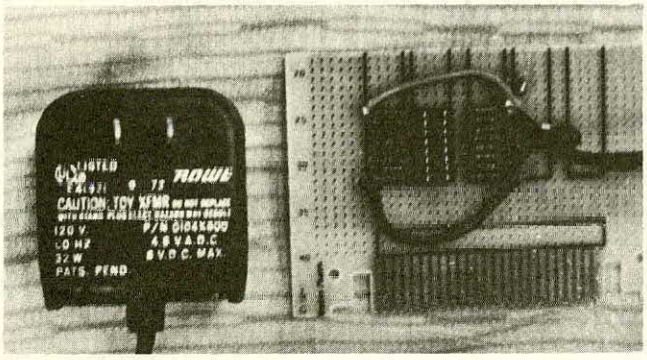


Display panel holder, Figure 4



Wire forming jig, Figure 5

Completed display panels were tested in the display tester. The tester should light all LEDs, an unlit row or column suggests an open in the drive line connection, an unlit LED suggests an incorrectly inserted LED or bad solder joint, and an intersecting dark row and column indicates a shorted connection at the intersection. The display tester is shown as figure 6.



Display panel tester, Figure 6

The circuit board was tested in stages. The voltage regulator and power connections were tested using a ten LED bar gragh display. Once proper operation was confirmed or assembly errors corrected, the ICs were inserted and the circuit board connected to a light display. The display allows conformation of proper soldering by connecting to each row and column. If an LED fails to light, a display driver pin is bent or a solder connection is bad.

RECORDS

A fair amount of paperwork is involved in this project. Parts must be selected, suppliers chosen. Worker time and quality control sheets must be kept. In the future it is hoped a client will assume these duties as a paid position.

OPERATING MODES

The communicator's several selection schemes allow matching to the users needs and abilities. Six scanning modes with 2,4,16, and 64 selections, two directed scanning modes, and quadrant subselection are available All selection schemes except for quadrant selection are speed dependent and use a sixteen position control to vary speed between .32 and 5.12 seconds per selection. Another feature of the communicator is its ability to accept two jacks. This allows communication between two system users using the same selection scheme but not necessarily the same symbol set.

SUMMARY

This project has demonstrated the feasibility of communicators with fewer parts than commercially available units with similar or lesser capabilities, and the ability of higher level state school clients to assemble these units for the more physically involved clients.

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ACKNOWLEGEMENTS

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ISSUES REGARDING FEEDBACK IN LIMB PROSTHESES

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ABSTRACT

The concept of feedback in the design of artificial limbs is discussed. Feedback schemes are categorized and examples described. Proper integration of the human with machine (human with prosthesis) remains a key problem in prosthetics and appropriate feedback schemes are probably necessary to bring about effective integration.

INTRODUCTION

Everything affects everything. To the extent that this is true, everything that happens has an influence, however slight, upon its cause; a condition where we might say there is feedback from every output to every input. Of course, under these conditions the concept of input, output, and feedback becomes blurred. We regard many actions as being substantially open loop; that is, where there is essentially no influence of output upon input or in the human-operator case where there is no knowledge of system output by the person producing the system input. Control of an output variable under these conditions is extremely difficult because an input must be pre-learned or pre-programmed in order to produce a desired output and there is no basis for adjustment of the output when perturbations occur in the input-output relationship or when external conditions alter the desired output.

The field of feedback control (control theory) has developed around the concept of taking output variables and feeding them back to the input in order to influence the system's output behavior; perhaps in such a way as to help it meet some performance criterion. Output information at the input may permit a system to compensate for changes that occur in input-output relationships or to compensate for the influence of external disturbances on the system. However, feedback is not without disadvantages, a primary one of which may be the development of instability when input and output interact to produce oscillations or to force the output to some extreme value. These kinds of influences have been extensively studied in technical systems, particularly in linear systems. They have been less studied in human-machine systems, partially because it is difficult to mathematically model the human operator.

The field of Human Factors has studied, to some extent, how to present displays to human operators so that the operators can efficiently and effectively control the mechanisms they are operating. Feedback of output variables so that the operator has knowledge of the results of his action, is common in these systems. Feedback to the operator is often in the form of visual displays (e.g. dials, lights, numerals, etc.) and in some cases in the form of auditory signals. There has been interest in other forms of signals to the operator, particularly tactile inputs, vibrational inputs to the skin, and electrical stimulation of the skin. Nevertheless, human operators of today's machines work primarily on the basis of visual input and upon the physical interactions between the operator and the machine and between the operator and the machine's control interface (e.g. steering wheel).

Although control theory and human factors have made substantial contributions to technological progress, neither have had much influence on the problem an amputee has in controlling his prosthesis whereby the machine is essentially a part of the human body and is not controlled by typical input modalities (i.e. through the hands or the feet).

FEEDBACK DESCRIPTIONS

Powered upper-limb prostheses are currently controlled primarily through visual feedback with some assistance through what has been called incidental feedback. This latter feedback results not by plan but occurs incidentally. Examples are motor noise, prosthesis vibration, socket forces, and the like. Sensory feedback can be supplemented artificially by vibration of the skin, by electrical stimulation of the skin, by auditory and visual signals, and by other means. Such feedback has been called supplemental sensory feedback (1), and abbreviated as SSF. Feedback information may also be received through the control interface, which has been called control interface feedback (2). An example of interface feedback occurs in the power steering systems of automobiles where substantial information about the front wheels is available to the operator through the steering wheel. Another possibility is to construct artificial reflexes that bypass the human-operator. An example of such a reflex arc is illustrated by a prehensor which through vibration caused by an object slipping from its grasp causes the mechanism to increase its gripping force until slippage stops (3). Finally, many well-designed limb actuator systems use velocity, force feedback, or some other feedback within the mechanisms (closed loops) in order to improve performance of the mechanisms.

In Figure 1 an attempt has been made to diagram the feedback pathways described in the last paragraph. This issue has been treated in more detail elsewhere (2) and that paper references many of the papers that describe the results of experiments with various forms of SSF. SSF has been examined extensively in research laboratories in attempts to incorporate it into prostheses or to use it as a communication link to persons with severe visual or auditory deficits. It is interesting that although there have been a number of attempts to use SSF to help amputees control their artificial arms there are few such systems in day-to-day use. Amputees frequently report that they appreciate SSF in a prosthesis (e.g. electrical stimulation of the skin with frequency modulation proportional to prehension force). However, SSF does not appear to greatly improve performance, where that performance has been measured objectively.

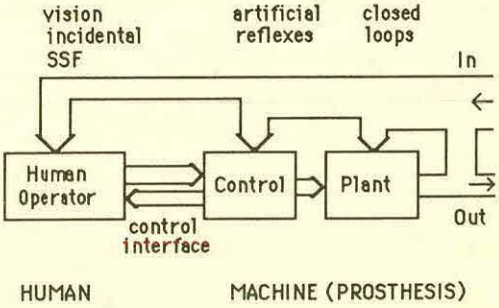

Figure 1. General diagram showing signal interaction.

COMMENTS

Some people believe control of artificial arms is poor because of the lack of appropriate feedback in the prosthesis design. This may be true but the key word is appropriate. It does not appear that the addition of any kind of SSF or other form of feedback automatically improves prosthesis performance. The information fed back needs to be in forms that are easily interpreted by the human and preferably in forms that can be interpreted at a subconscious level.

In an unpublished paper (4) the author has discussed a control interface feedback method that he believes can be effective in control of upper-limb prostheses; partially because it incorporates feedback through the control interface. The method involves the principle of extended physiological proprioception (e.p.p.), which was originally put forward by Simpson (5). In this system, control is such that output position corresponds to input position at all times, output velocity corresponds to input velocity and output acceleration corresponds to input acceleration. The body's own proprioceptive senses are employed to inform the body of the state of the prosthesis joint being controlled. This method seems natural. It is employed in traditional body-powered, cable-actuated prostheses and is one of the reasons these prostheses integrate well with their users.

SUMMARY

The human body uses multiple output channels and multiple input channels to effectively control its limbs. In artificial limbs, present technology usually limits the number of output and input channels to ten or fewer. Consequently, we cannot hope at this time to duplicate the sensory-motor performance of the human limb. However, as powered limb components increase in number and in performance and as the number of clinical fittings of these prostheses expands, more and more is being learned about how to improve their function. Expanded use of feedback in powered limb prosthetics can surely be expected as experience with these prostheses continues to grow.

ACKNOWLEDGEMENT

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ABSTRACT

In this paper the present status of sensory feedback in upper-extremity prosthetics is discussed in terms of the results of a small scale clinical evaluation of a system developed by the authors. Suggestions are made concerning priorities for future research.

INTRODUCTION

Among other topics which enjoyed popularity in the 60's and 70's was the provision of sensory feedback in upper-extremity prostheses. At best, clinical success was marginal, and most projects seem to have been abandoned. It is timely to review this topic now because, while the need is unchanged, technological advances may make a solution more feasible.

BACKGROUND

A cursory review of the literature in the period 1965-1980 discloses published research in sensory feedback for prostheses from Australia [1], Canada [2], Germany [3], Italy [4], Japan [5], the Netherlands [6], and Sweden [7] as well as the USA [8-10]. Clearly there is some concensus that sensory feedback is needed. However, priorities among the various purposes of feedback, described with particular clarity by Korner, [7], vary considerably. There is the possibility that feedback will improve the mechanical function of the prosthesis, in the sense of making it a true closed-loop system. This aspect of the topic is reviewed well by Childress [11]. Alternatively there is the recognition of the normal hand as a sensory organ, and the possibility of restoration of some portion of that function through sensory feedback.

Whatever the specific objectives, it is fair to say that clinical success with sensory feedback in prostheses has been marginal. Mann concluded a thoughtful 1981 review [12] by noting that "... despite the early and persistent awareness of the importance of sensory feedback ... this remains among the most refractory of problems...". That position is equally valid today.

THE UNB EXPERIENCE

Research on this topic in the UNB Bio-Engineering Institute led to the development in 1979 of a system using electrocutaneous stimulation to indicate prehension force as measured by a strain gage on the prosthetic hand [2]. Ten amputees were fitted with the system for clinical evaluation, and the results reviewed in 1984. Four clients had been lost to followup, and six were included in the study. Data on the study are as follows:

Age at Feedback Fitting: Min 7 yr
Max 38 yr

Followup Period: 10 yr 1
5 yr 2
2 yr 3

Sex: Male 5
Female 1

Etilogy: Congenital 3
Acquired 3

Amputation level: B/E 4
W/D 1
Partial Hand 1

Prosthesis: Myoelectric 5
Switch-Controlled (Partial Hand) 1

Previous Prosthesis Use: Min 1 yr
Max 16 yr

Backup Prosthesis: None 3
Conventional (one only) 2
Conventional (home and work) 1

Results:

Status at review: Discontinued 5
Still using 1

Reasons for discontinuing: To change to removable battery pack 2
High repair rate 1
No advantage 1
Error in prescription of new prosthesis 1

Overall Rating by Users: Advantageous 3
Questionable Adv. 1
No Advantage 1
Disadvantageous 1

Future: Would recommend feedback to others 5/6
Discontinued but would like to try next system developed 4/5

Specific Advantages Cited: Ability to hold an object without visual attention 4
Prosthesis "more like part of body" 2
Ease of handling delicate objects 2

Specific Disadvantages: Frequent repairs 4
System unreliable 2
Stimulus irritating 2

From objective data and interviews it is evident that both our technological skills and our insight into clinical needs are inadequate.

PRESENT STATUS

Both from the UNB study and from the published literature it is apparent that the mechanical functioning of a prosthesis is neither dependent on or greatly influenced by provision of sensory

feedback - at least with present systems. It is thought that a myoelectric hand is at a great disadvantage relative to a conventional terminal device because of the loss of feedback furnished by the harness. But users of myoelectric prostheses show skills which are inconsistent with dependence upon visual feedback alone. This seems to be particularly true when congenital amputees are fitted very early in life [13].

Further, comments of users of the crude systems which have been available for evaluation indicate an appreciation of the feedback unrelated to improved mechanical function. The comment in the UNB study that the prosthesis was "more like part of the body" is more likely an indication of some beneficial somatosensory replacement than of improved mechanical performance.

Thus we have something of a dilemma. Feedback is appreciated, even in a crude system, despite ample evidence that "it isn't worth the trouble" in any practical, measurable, sense. The question is what to do next.

THE FUTURE

Present efforts at UNB are concerned with improving the man-machine interface through the use of a surgically-implanted telemetry system [14]. It is felt that overcoming the unreliability and the possibility of stimulus discomfort due to insecure electrode-skin contact may lead to improved understanding of other aspects of the problem.

Another aspect of feedback which deserves a thorough review is the choice of input parameter and the transducer system used to measure it. The ideal finger-tip transducer, small in size, very stable and rugged, continues to elude researchers. There may be grounds also to suspect that the choice of input parameter - almost always force or pressure - is based on what is feasible more than on what is needed.

This leads to the most important objective which should guide research in the next decade. That is, to work for a better understanding of the clinical need. Only as this is accomplished will it be feasible to focus sufficient technological effort on the specific problems to permit a reasonable solution.

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SENSORY FEEDBACK in UPPER LIMB PROSTHESES: a ROBOTICS PERSPECTIVE

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ABSTRACT

In upper-limb rehabilitation there has been a tendency to focus attention on the "missing limb" to the relative exclusion of "missing function". We have responded technically to the "I cannot move" imperative rather than "I cannot feel". In general, biological systems never attempt one without the other. Why then have technical efforts to restore upper-limb function neglected sensory feedback? There are at least two main reasons for this situation: 1) amputees have "compensated" for the loss of tactile and kinesthetic sensation by using alternative sensory channels (primarily vision and hearing); and 2), the engineering of closed-loop motion control systems is relatively difficult. Based on results from human factors, psychophysics and robotics research, this paper defines a rationale and strategy for renewed emphasis on sensory-motor integration in upper-limb prostheses.

INTRODUCTION

It is hypothesized that full expression of the potential of upper-limb prostheses can not be realized without a balanced emphasis on sensory and motor function. An examination of "first principles" in neural control of movement physiology (Talbot and Gessner, 1973), manual control (Sheridan and Ferrell, 1974), robotics (Brady et al, 1982) and tele-manipulation (Thring, 1983) will provide the key ingredients for a strategic approach to the next generation of upper-limb prostheses. The following review is based on the assumption that "manual-control" tasks are a valid experimental model for the reach-pick-reach-place tasks so typical of upper-limb prosthesis usage.

SENSORY COMPENSATION

An examination of **Figure.1** dramatizes the richness of sensory feedback available to the human operator in a manual control system. Importantly, the position, rate of change and acceleration of the output variable are directly observed through multiple channels. The operator benefits from this redundancy because the sensory information channels have different properties and reference frames. Through reference to **Table.1** you will be reminded that some channels are biased towards the extraction of relative, rather than absolute, measures. Vision and touch adapt rapidly whereas hearing and kinesthesis have rather long adaptation time constants. Vision and hearing are telesensory modalities requiring selective attention for manual control tasks. In contrast, touch and kinesthesis, directly sense internal body states related to the controlled variable and make this information available to the operator at a non-conscious level. The importance of this difference will be examined in the next section.

Externally powered upper-limb prosthesis usage relies almost exclusively on visual feedback. Vision requires selective attention. Selective attention is a primary determinant of fatigue (work load and information processing demand) in manual control and instrument monitoring tasks (Sheridan and Ferrel, 1974). While effective, some investigators (e.g., Mann and Reimers, 1970) have suggested that the relatively low acceptance rate for externally powered prostheses may be a result of over dependance on visual compensation and under-utilization of kinesthesis.

Human powered upper-limb prostheses retain access to touch and kinesthesis information channels, in varying degrees, as a function of specific design features. While this would appear to directly alleviate the selective attention workload, these sensory channels are typically compromised because they retain two meanings: 1) they continue to be used as primary touch and kinesthesis channels for their original anatomical location and function; and 2), they are assigned a new "secondary" function in connection with prosthesis usage. Hence, while using the "correct" sensory channels, we have again added a secondary information processing task that will cause confusion and fatigue.

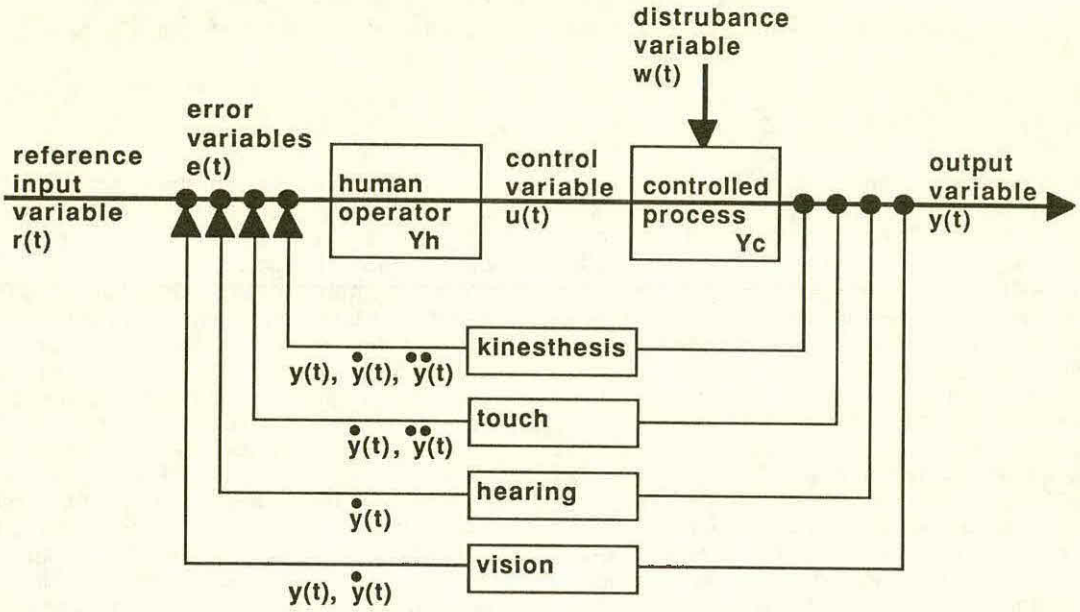
An alternative approach may be found In surgical reconstruction of upper-limb and hand function. Chase (1986). In this context, muscles (kinesthesis information) and skin (touch information) demonstrate marked plasticity following surgical re-allocation of their function and placement. Relatively complete perceptual remapping can be demonstrated within 4-6 months. The new function becomes the primary and "only" representation of these sensors and muscles such that confusion is eliminated and the workload minimumized.

INFORMATION PROCESSING

One may view all human motor and sensory activity as information rate limited. Visual monitoring of instruments and/or manual control targets has been studied extensively (Rouse, 1980). It may be safely concluded from these studies that dependence on constant visual attention for upper-limb usage does make an appreciable time varying demand on one's conscious information handling capacity. As a "secondary task" to primary "activity of daily living" manipulation objectives, visual monitoring of the prosthesis can be expected to increase the workload (decrease the time to fatigue) proportionately.

While measures of "non-consious" information processing channel capacity are not readily available, there is ample intuitive evidence that new manual

Figure.1.Simple Manual-control System
(after Sheridan and Ferrell p.177)



differential sensory feedback

tasks can be very demanding (e.g., learning to touch type) and subsequently become "automatic" (non-conscious). For the moment, suffice it to say that non-consious information processing capacity is very much greater than that associated with consious selective attention. There is a parallel to this situation in robotic motion control.

ROBOTIC MOTION CONTROL

Robot motion control is roughly equivalent to manual control, with a computer replacing the human operator in **Figure.1**. To quote Harmon (1985), "touch-sensing technology for robotics and prosthetics is presently very primitive". In fact, an examination of Brady et al (1982) will reveal no mention of tactile sensation for robot motion planning and control. How can this be the case? As we know them today, robots are largely blind, deaf and have no sense of touch. They do, however, have the equivalent of a highly refined sense of "kinesthesis". Using a combination of joint and motor motion sensors, the control computer knows precisely where the manipulator is at any given moment. The control loop is closed by direct internal measurement of joint position, rate of change and, less frequently, acceleration.

There is considerable robotics research concerned with machine vision. But machine vision is used primarily for motion planning, not for direct motion control. This is comparable to human use of vision during manual tasks. Technically, the complexity of touch sensing is similar to vision. As with vision, machine hearing is used (though rarely) for task planning. It may be concluded that kinesthesis is a rather important sensory modality when it comes to motion control by man and machine.

TELE-ROBOTIC MOTION CONTROL

As reported by Leifer (1982), it has been demonstrated that functional restoration of upper-limb motor function can be accomplished by the use of industrial manipulators as "tele-robots". An examination of the information processing requirements for remote manipulation revealed that two thirds of the information required for a "reach-and-touch" task was allocated to the terminal third of the task (i.e., the relatively demanding position-to-touch phase). It was concluded that a robotic aid should have proximity and touch sensing capability to perform this task automatically, to reduce the operator's workload. For this class of "prostheses", human vision and hearing play their normal behavioral roles in support of motion planning and supervision. The human operator's senses complement those of the robot.

In his review of industrial tele-robots, Thring (1983) also looks for symbiotic task performance by man and machine. He stresses the importance of force feedback in tele-manipulation by able bodied operators. Force information can be conveyed by finger-tip touch stimulators and/or as joint forces applied to the operator. Importantly, force feedback increases task performance accuracy and reduces fatigue.

CONCLUSIONS

This review of the role of sensors in both human and machine motor control systems strongly suggests that future attempts to develop upper-limb protheses should explicitely balance the allocation of resources

Table.1.Transducer Characteristics for Human Senses

(an extention of Sheridan & Ferrell, 1974, p.261)
(● from Cratty, 1967, p.72))

BASIC ATTRIBUTE	Kinesthesis● mechanical tension and displacement	Touch mechanical pressure and displacement	Hearing mechanical vibration	Vision electromagnetic radiation
primary level of conscious interaction	passive non-conscious	selective non-consious	passive consciousness	selective consciouness
adaptation time constant	20 - 40 msec for rapid increase 20 - 30 min steady-state	5 - 10 seconds for nonvibratory loads	30 - 90 minutes essentially non-adapting	20 - 40 ms for rapid increase 20 - 30 min for rapid decrease
dynamic range for stimulus magnitude	1000 detection to pain	100 detection to pain	100,000,000,000 detection to pain	1,000,000,000 detection to pain
absolute identification of magnitude	5 to 7 steps	5 to 7 steps at 200 -300 Hz on finger tip	5 to 7 steps at 1,000 - 2,000 Hz	5 to 7 steps on gray scale
relative discrimination of magnitude differences	360 just noticeable differences	15 just noticeable differences	325 just noticeable differences at 2000 Hz	570 just noticeable differences
spatial resolution	0.3 degrees shoulder rotation	1 mm two-point threshold on finger	3 - 6 degrees median plane binaural localization	1 min of arc at fovea with bright high contrast target
temporal resolution	5 - 10 msec depending on intensity	10 - 50 msec depending on intensity	2 - 3 msec depending on intensity	20 - 50 msec depending on intensity

between sensing and actuating. Inner "non-consious" and outer "selective attention" control loops should both be used to minimize the user's conscious information processing and transmission load. It would appear that better utilization of kinesthetic sensation would parallel the evolution of robotic motion control technology while making effective use of residual body resources.

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THE ROLE OF SHAPE SENSING IN REHABILITATION ENGINEERING

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ABSTRACT

Computer aided design and manufacturing methods are being applied in attempts to improve the quality and reduce the cost of current labour-intensive methods of custom fitting appliances in rehabilitation engineering. The first step in these systems involves taking some measurements of the body part to be fitted. Shape sensing instruments have been developed for this purpose that generate detailed numerical images. Such instruments have an important role to play in the development of fitting methods by analysing manual fitting methods and by developing "libraries" of standard shapes for use in the computer aided fitting process.

INTRODUCTION

Many categories of devices require to be custom-fitted to closely match body shape. These categories include prosthetic limbs, functional and cosmetic orthoses, special seating and orthopaedic footwear. Shape sensing technologies offer alternatives to labour intensive casting and measurement methods. The adaptation of computer aided design and manufacturing methods to rehabilitation engineering has provided much of the impetus for the development of shape sensing for this field, see Saunders et. al. (1985). This paper outlines the method of shape sensing used in our laboratory and then discusses the role of shape sensing in rehabilitation engineering. A fuller description of one of the shape sensors has been provided elsewhere by Fernie et. al. (1985).

INSTRUMENTATION

The instruments developed in our laboratory are each based upon the same principle. A line of light (usually generated by a cylindrical lens attached to a low powered helium-neon laser) is projected onto the object measured. This line is viewed from an angle by a video camera. A counter begins with each horizontal scan line of the video and is arrested when the image of the projected line is encountered. Thus 242 numbers corresponding to the count value at each horizontal scan line on the video frame are retrieved.

In order to measure the total surface it is necessary to cause the projected line to scan that surface. Three of several possible strategies have been used in our laboratory to date. The simplest involved rotating the object slowly in front of a stationary line projector and camera and capturing video frames of data at fixed angular intervals. This was found to be appropriate for certain inanimate objects such as prosthetic casts. Where greater speed was required the line projector has been rotated about the object at a rate synchronized to the collection of data from stationary cameras placed circumferentially. A third method employed linear motion of the camera and projector past a stationary object and was selected for foot and trunk measurement. The numbers of cameras and projectors have varied depending on the typical shape of the object to be measured and the measurement strategy chosen.

DISCUSSION

Jalkio et. al. (1985) have examined many of the theoretical issues related to the design of shape sensors employing triangulation methods. Our instruments, however, do not employ the geometrical calculations described by these authors. In order to reduce the cost of components and the precision required in manufacture, the instruments are calibrated using reference shapes to generate "look-up" reference tables. The three dimensional coordinates of the measured object are calculated by reference to these tables.

The technology that we have developed is only one of several possible ways of sensing shape. The selection of the shape sensing method is simpler than the determination of the most appropriate ways of applying that technology.

The need for precise shape measurement is not obvious when one considers the fact that the shape of an amputation prosthetic socket does not match the shape of the residual limb and, similarly, shoes do not match identically the shape of feet. The difference is, of course, a consequence of the need to provide a shape that supports the body part and distributes the applied forces appropriately. In theory, such a shape could be developed

during the computer aided design process if not only the surface shape but also information on tissue mechanical properties and on the desired patterns of stress distribution were input. In practice the limitations in the state of the art of in-vivo tissue property measurement and mathematical modelling of complex composite biological tissues will probably preclude this approach for the present.

Skilled prosthetists, orthotists and footwear specialists have developed relationships between the surface shape and the modified shape empirically through many years of patient fitting. It seems important to take account of this knowledge and apply it in the computer aided process.

A recently developed measurement system in our laboratory obtains approximately 10,000 measured points on the surface of the foot. In contrast, when one buys a ordinary shoe in the shoe shop the assistant will probably make, at most, two or three manual measurements of the foot. So few measurements are needed since a range of standard shapes of shoes has already been manufactured and it is only necessary to select the one with the closest fit.

Someone with a foot deformity would not be fitted so easily but the difference is one of degree. A larger range of standard shapes would be required to include most deformities. This would be impractical if the shapes were stored as finished shoes or even as shoe lasts. However, storage of a large range of standard shapes in numerical form is practical. The actual number of standard shapes to be stored can be reduced by two means. Firstly the shapes can be mathematically enlarged or reduced in order to vary "sizes" and secondly the shapes can be constructed from different shape elements that can be selected and assembled to match different deformities.

The challenge now is not so much to develop shape sensing instruments since several exist; but rather, to determine how best to classify shapes and to select and adjust them.

Shape sensors are needed for two research purposes a) to measure large numbers of successfully fitted devices in order to develop libraries of standard shapes. b) to study the relationships between measurements of the body surface shape and the modified shapes of the fitted devices in order to develop selection strategies for modification based upon unloaded shape characteristics.

The shape sensors required for the two research tasks will be relatively sophisticated instruments capable of measuring with reasonably high precision whole surfaces. Whereas, the shape sensors required for general clinical use will probably be much simpler devices that will provide only the reduced data needed to make appropriate selections from libraries of standard shapes or shape modules and to adjust them to the individual.

ACKNOWLEDGEMENTS

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SHAPING TECHNOLOGY IN THE FIELD OF PROSTHETICS

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ABSTRACT

A set of design criteria for sensing the shape of an amputee's stump is presented and used as the basis for evaluating various shape sensing technologies. A mechanical probe type shape sensing system is described and its use in quantifying the external shape of the AK stump is discussed as it relates to generating a grid for finite element analysis in CAD/CAM studies and comparing the segmental volumes of the loaded and unloaded stump.

Quantifying the surface topology of an object is the first step in numerous processes ranging from topographical map making to making coins and artificial parts for the body. Each application has its own requirements for accuracy, precision, and surface smoothness. To satisfy the diversity of requirements, a number of techniques have been developed over the years (2,4,5,6,8,9) and have met with varying degrees of acceptance when new applications are tried. Thus before the most effective technology can be selected, the investigator must carefully delineate the required results. In this regard, sensing the shape of an amputee's stump requires a measuring system that can collect the necessary data before the amputee tires and provides data that are at least as precise as the data currently collected by a prosthetist and can be manipulated to produce the same sorts of information that are currently used by the prosthetist. Moreover, the measuring system must be easy to operate and not have large space requirements if it is to be clinically usable.

The technologies which have been applied to the task of quantifying the shape of an amputee's stump can be classified into two basic groups-contacting and noncontacting sensors. In the first category are the laser based measuring systems and stereo photogrammetry (1,3,6,7,) while the second includes mechanical probe devices (9) and ultrasonic systems which require the leg to be immersed in water. The devices in the first category generally provide very fast data collection (2-5 seconds); but, because of the short distances involved in this application and the speed of light, it is generally not practical to resolve distances less than ±0.2 inches (5mm). On the other hand, the devices in the second category are generally capable of greater precision (less than ±0.05 inches) but they require substantial time to acquire the data (3-10 minutes) and they require more user input.

By working with several prosthetists, it has been possible to determine a set of criteria that the shape sensing system must satisfy if it is to emulate the stump measuring technique used in current prosthetics practice. The criteria are:

1. The shape sensing process should take no more than 7 minutes to collect the required data.
2. The measurements should be precise enough to match the current circumferential measurement error of ±0.25 inches. The requirement means that radial distances must be accurate to ±0.04 inches.
3. Vertical measurements should be accurate to ±0.1 inches.
4. Measurements should be reproducible within 2 percent.

Since the performance of measuring systems which fall into the first category is well documented in other publications (5,7,8) and the Rehabilitation Engineering program at Baylor College of Medicine and The Institute for Rehabilitation and Research has concentrated on developing contact type shape sensing devices, this paper will concentrate on shape sensing systems that use mechanical probes.

In designing a contact type shape sensing apparatus, it is imperative to consider a number of traditional electronic design parameters. The linearity of the potentiometers or Linear Voltage Displacement Transformers (LVDT) used have to be examined and assessed in the design. When the voltage output from these devices is to be buffered and/or amplified, the offset error, maximum non-linearity, and gain error must be considered in a worse-case-analysis to insure that the final design remains within the constraints of the original specifications. When the system logs the data digitally, the analog to digital (A to D) converter needs to be selected by considering: 1) the least significant bit uncertainty error (assuming the utilization of the full dynamic range of the A-to-D); 2) offset error; and 3) the speed of conversion to assure the required data acquisition rate can be achieved. Position encoders may be used instead of LVDT's or potentiometers provided that the maximum resolution of the encoder is considered and that the maximum rate of change in position does not cause counts to be missed; this is particularly important if these counts are logged by the microprocessor. Stepper motors can also be implemented in the design assuming that the system can accurately remember the actual position of the probe. In this regard mechanical linkages need to be minimized to reduce slop in the system which could lead to significant uncorrectable errors.

Two unique shape sensing systems have been developed at TIRR and have been used to quantify the shape of sockets for use in lower extremity prostheses.

The Range of Motion System is the first device that was developed. It is a generalized instrument that can record the coordinates of points in space or on the surface of an object referenced to the axes of the instrument. The instrument is capable of resolving linear distances to ±0.1mm and arcs to ±0.01 minutes. Unfortunately, the system is rather unwieldy for use in characterizing stump shape and was found to be better used for other applications where the surface is less geometric than a stump, e.g. the foot or the contours of the scoliotic back.

The contourgraph is the second measuring system that has been developed and it appears to be able to meet or exceed the criteria previously cited. The original prototype consisted of a fixed disc and a concentric rotating feeler which by contact determines the contour of a cross section of an amputee's stump and also the prosthetic socket at a chosen level. The contour is then traced out by a plotting pen. The instrument is capable of carrying more than one feeler enabling it to trace more than one contour at different levels at a single turn. The pressure of the feeler against the stump is adjustable. During the tracing, the pressure of the feeler on the stump is constant for all levels. That is, the pressure is set at the start of the tracings and kept constant throughout.

The prototype system was evaluated and an automated, computer compatible model has been developed. This new version uses capacitance switches to sense the surface being measured and the pressure required to activate the switches is less than 45 dynes/cm^2 so that the tissue deformation is not detectable (less than 0.01mm). The instrument uses radial measurements at intervals of 20 degrees around each cross section to characterize the shape of an above knee stump and the cross sections are one inch apart. These variables can be adjusted if other more complex shapes are being characterized. The instrument simultaneously measures diametrically opposed points to reduce the measurement errors and reduce topographic data acquisition time so that it is convenient to characterize a 13 inch above knee stump (234 points) in 6 minutes.

While this system works quite well for generating the points required to fabricate a finite element grid or to calculate the volume of the stump and/or socket, it does not provide the level of detail which is required to effectively characterize a foot or hand nor does it work well in replicating shapes that have small concave discontinuities in the surface.

In order to effectively sense the shape of a structure the purpose of the measurements must be clearly stated or it is very probable that the shape sensing technology will be misused.

In prosthetics it is imperative that engineers understand how the shape information fits into the intended use so that efficient, cost effective aids can be developed that solve the problems rather than confound the users with confusing data and overly complex machines.

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AN OPTICAL METHOD FOR NON-CONTACT 3-D MEASUREMENT

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ABSTRACT

A non-contact 3-D measurement system suitable for use in fitting prosthetics is described. Optical and electronic aspects of its construction are detailed. Problems common to such systems are discussed and solutions presented. Experimental results from a prototype system are included.

INTRODUCTION

There are many reasons that a non-contact measurement system is more suitable than contacting techniques for obtaining the data required for proper fitting of prosthetics. Since non-contact systems exert no force on the surface being measured the surface will not be deformed, hence the measurement can be more accurate. Furthermore, a non-contact method will cause the patient no discomfort. The technique we describe here was originally developed for automated inspection of mass produced articles but it is clearly applicable in this field as well.

A number of methods have been demonstrated for 3-D coordinate measurements of an image. These include time of flight, stereo matching, and structured light. Most commercially available systems employ structured light, as this technique affords the resolution and speed required for many practical tasks. In this paper we discuss a structured light 3-D measurement system with the potential for high speed and no moving parts.

BACKGROUND

Common structured light systems project a point or stripe of light onto an object and use the position of the stripe's image on a detector to determine the distance between the detector and the object. The light must be scanned over the object to obtain a 3-D picture. By replacing the single point or stripe by a set of parallel stripes we can make distance measurements along several lines of the object simultaneously. Fitting several light stripes on a camera scan line does reduce the range over which each stripe is free to travel, hence reducing the depth resolution. However, trading off depth resolution for speed is often necessary in real time applications.

Structured light systems usually base their distance measurements on the absolute position of a light stripe in the detector plane, comparing this with the stripe's position for some known reference distance. This technique requires that the source and detector be rigidly mounted to one another, lest variations in their relative position and orientation affect the measurements. This also precludes the combination of diffractive optics (which can conveniently distribute laser beams) with laser diodes since the wavelength shift of a laser diode with temperature would lead to a shift in the position of the stripes which would be interpreted as a change in distance.

To overcome these problems, we considered using the relative positions of stripes in two different detectors as our depth cue. Such a system is shown in Fig. 1 with the resulting formula relating distance to image separation (1). In reality, cost, weight, and size restrictions rule out the use of dual cameras and instead the optical system of Fig. 2 was constructed, using a system of mirrors in front of the camera lens to produce a double image of each stripe. By properly positioning the mirrors, L (as defined in Fig. 1) vanishes, leaving a simple expression for the distance.

MATERIALS AND METHODS

We produce multiple illumination stripes in our system (Fig. 2) via a multifaceted hologram (2), H, which spatially divides an expanded incident laser beam into 16 beams. The 16 beams collectively overlap at the position of lens L_3 (to form a small effective source) but individually come to focus at the object plane, O. A

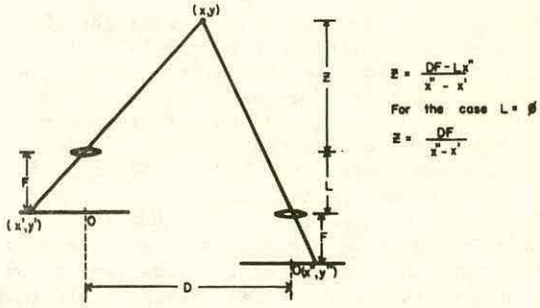

Figure 1

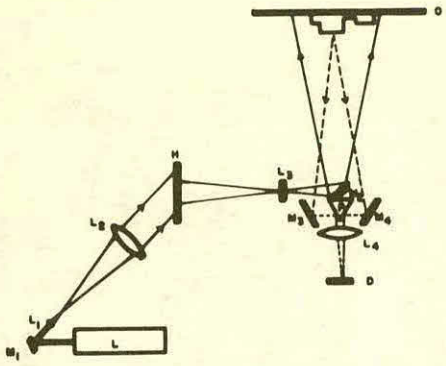

Figure 2

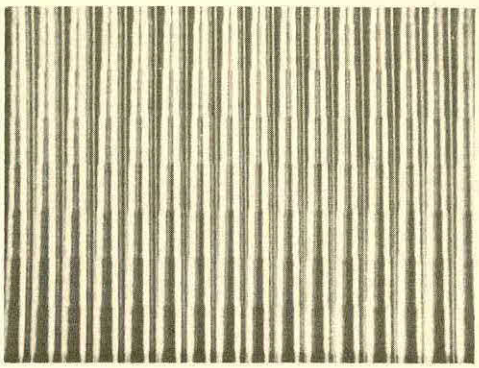

Figure 3

cylindrical lens, L_3, spreads each beam into a vertical stripe so that a set of 16 focused vertical stripes illuminate the object. Transversely translating lens L_2 in our prototype system, scans the stripes across the surface of the object. A television camera (lens L_4 and detector D) directed at this pattern can generate the data required for a 128 by 128 by 50 element 3-D picture in 8 frames. To improve the speed and reliability by eliminating moving parts, the translation of the collimating lens can be replaced by the sequential pulsing of multiple light sources (e.g. laser diodes). Again only 8 diodes would be required to produce a picture with the resolution mentioned above. In Fig. 2, the light source appears to be located at the lens pupil so that the center position of any pair of stripes does not change as z is varied, which further simplifies data interpretation and also allows one to treat the system as two single triangulation systems in the event of hidden part problems.

The dual virtual detector architecture shown in Fig. 2 does have some limitations. Focus and effective aperture size vary across the field of view, and mirror walk off limits the operating range. The separation, D, of the virtual detectors cannot be smaller than the lens aperture which precludes large operating ranges. However even with these limitations we have found that this system performs quite well. Figure 3 shows the image of stripes corresponding to a staircase object. Adjacent steps differ by 2.5 mm in distance from the detector, while the average range is 60 cm. The straight lines in Fig. 3 are bin markers added by our computer to indicate the zone in which the stripes should be found.

Although real time processing of the acquired image requires special purpose hardware, we have tested all of the necessary algorithms by software simulation. The main processing task is to determine the position of each stripe as accurately as possible. The Nyquist sampling theorem tells us that a sufficiently sampled, band limited signal can be reconstructed from its samples with arbitrarily good accuracy. Hence, in the absence of broadband noise we should be able to determine the position of our stripes with sub-pixel accuracy. Several good reconstruction algorithms have been tried. Smoothing by convolution with a Gaussian or Sinc function is very accurate but computationally expensive. Linear interpolation of threshold crossing points is much simpler and yields very good results for stripes with Gaussian profiles. Once the positions, and hence the separations of the stripe images have been determined, the x, y, and z coordinates of the object point can be found via the formula in Fig. 1 or lookup tables. A 3-D picture of the object can then be displayed with distance encoded as intensity or pseudocolor.

RESULTS and DISCUSSION

The accurate determination of stripe position is hindered by several ubiquitous optical problems. Surface roughness and variations in object reflectivity can alter the profile of a light stripe, and since these effects are angle dependent, the two images of a single illumination stripe can look quite different. In particular the diffuse reflectivity of skin can degrade stripe images rather severely, and additional work must be done to determine ways of improving the signal to noise ratio of such images. Lens aberrations can also distort the stripe profile, and can distort the distance - displacement relationship. Laser speckle is a problem since the source cannot be in focus at all distances and since the image of the stripe must be several pixels wide if we are to determine its position with sub-pixel accuracy. We have tried to

minimize these problems by lowpass filtering the video signal to reduce the effects of surface roughness, choosing lenses with minimum abberations, and by maximizing the system's depth of field to reduce speckle effects.

Even with software processing of the data, the system is fairly fast (2 minutes for a 128 by 128 picture) with most of the time used for data transfer. Hardware processing should reduce the time to well under 1 second. It is also fairly accurate (0.4 mm resolution over a range of 2 cm). Of course depth resolution, spatial resolution and speed can all be traded off for a given application. Some of our experimental results can be seen in the following photographs. Figure 4 shows a 3-D image of a printed circuit board when height is encoded as grey level. Notice that the orientation depressions in the integrated circuits are clearly visible. Figure 5 shows an image of a pile of metal washers, their relative distances readily measurable. Figure 6 is a side view of a nose. The object scanned to produce this data was a manikin rather than a human subject.

ACKNOWLEDGMENTS

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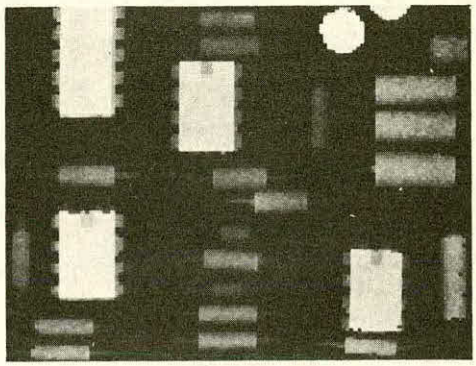

Figure 4

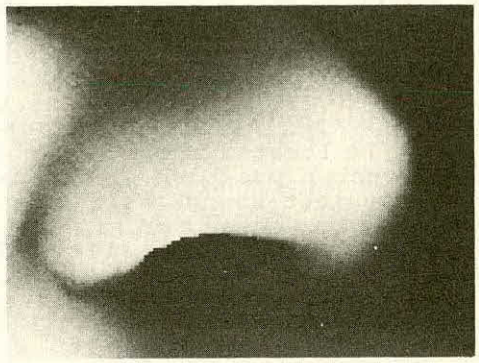

Figure 6

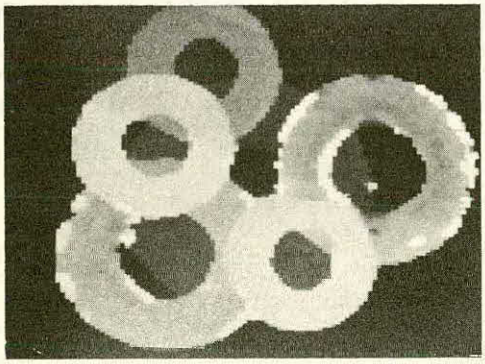

Figure 5

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A NEW DEVICE FOR MEASUREMENT OF VISUAL ADAPTATION

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ABSTRACT

A new computer controlled device for testing kinetics of visual adaptation is described. The reliability of the device was evaluated and found to be high (r=0.80). Patients with known risk factors for retinal disease were tested and found to have significantly slowed adaptation even though standard clinical tests of vision showed no abnormalities. This non-invasive measure of retinal function appears to be more sensitive to early alterations in function than existing clinical tests.

INTRODUCTION

The normal human visual system can respond to changes in light intensity covering about 12 log units; the photopic range (daytime levels) easily extends over a 6 log unit range (light differing by a factor of a million in intensity). This remarkable capacity to adjust to differences in intensity depends on rapid neural adjustments commonly called adaptation, a process which is retinal in origin. Defects in this process, caused by disease, can alter the speed and range of adaptation, making adjustments to changing illumination difficult or impossible. Slow adaptation can force unpleasant changes in lifestyle; walking outside for 15 minutes may effectively "blind" the individual for several hours.

In order to detect subtle changes in adaptation capability, it is possible to stress the retina in much the same way that treadmill tests are used to stress the cardiovascular system in order to detect early abnormalities. By measuring the recovery of visual sensitivity following exposure to a moderately bright light, it is possible to evaluate the integrity of the retinal adaptational mechanisms. Currently, clinicians have no standardized way of assessing the speed of these rapid adaptational processes which would allow identification of those patients who suffer from slow adaptation for the purpose of early diagnosis of anomalies as well as for protective filter treatment.

We have developed an instrument which allows measurement of retinal adaptation while separating out confounding factors such as alteration in the clarity of cornea or the ocular lens or uncorrected refractive errors. Using this device, we have demonstrated significant alteration in time constant of recovery in patients with hypertension and diabetic retinopathy who showed no abnormalities on standard clinical vision tests.

DESCRIPTION OF DEVICE

The test stimulus consists of a circular 5 min of arc flashing target presented on a moderate photopic background of 23 cd/m^2 at optical infinity. The test stimulus is generated by a yellow light-emitting diode. The target is presented in the center of a bold black Maltese cross to aid fixation (see Fig.1). The intensity of the test stimulus is controlled by an Apple IIe through a combined D/A and Digital I/O controller card in the Apple. The 20° glare field is produced by a standard photoflood lamp (see Fig.1). The intensity of the glare is fixed at 5.6x10^4 cd/m^2. (Calibration of the glare source is accomplished through a rheostat on the side of the device). The onset and duration of the glare source is under control of the Apple II using the controller card and a relay. The timing of the glare as well as timing of patients responses requires the use of a clock card in the Apple.

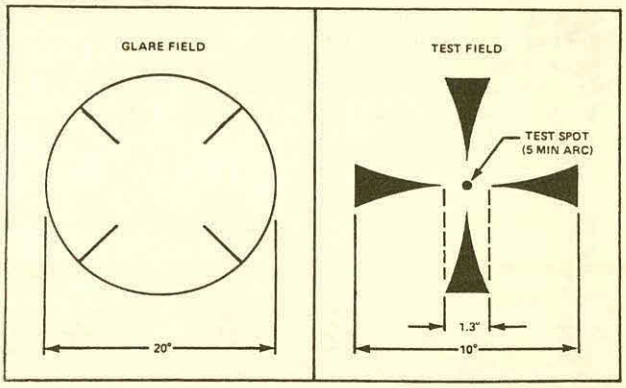

Figure 1

Target Configuration

Procedure

Prior to glare exposure, each patient's contrast threshold is measured. The contrast of the test target is gradually increased until the patient indicates that he sees the stimulus by pressing a button. Three measurements of this contrast threshold are taken before glare exposure. The patient then fixates the center of the glare field. Immediately after a 10 second exposure to the bright field, the patient's gaze is directed to the center of the Maltese cross, where the target, set at a preset contrast level, is intermittently presented (125 ms at 4 Hz). When the patient recovers sufficient sensitivity to detect the flashing spot, he pushes a button, causing the time to be recorded by the Apple and the contrast of the target to be further reduced below his threshold. When sensitivity has recovered further and the spot has been detected again, the patient pushes the button again. The time taken to recover to five preset contrast levels are recorded in this manner. The contrast levels are chosen to give approximately equal time periods between recorded points.

Analysis of Results

A number of factors can appear to slow down glare recovery. For example, if the patient is not wearing his glasses during the test, the "raw" measured recovery times will be prolonged. In addition, corneal haze and cloudy ocular media, such as early cataracts will significantly increase the measured recovery times. To avoid these confounding factors, we analyze the data collected in such a way that we derive a time constant of recovery, which is independent of pre-retinal changes. For each patient and for each glare recovery measure, we determine the time constant of the exponential recovery curve by plotting recovery time in seconds against log target contrast using a procedure that takes each person's contrast threshold into account. The procedure involves fitting an equation of the kind:

$$y=ae^{bx} + c$$

to the data where the x-values are the recorded recovery times and the y-values represent the contrast levels of the target. Least squares linear regression analysis is then performed by taking the natural logarithm of both sides of the equation and then determining the slope and intercept. The inverse of the slope "b" is the time constant, representing retinal kinetics, and the constant "c" represents the contrast threshold prior to glare exposure. The time constant is the time taken to recover 68 % of the full amplitude. Figure 2 shows the resulting fitted curves superimposed on the data points for two patients. The time constant of recovery is quite different for the two as indicated by the summary statistics below the graphs. The finely dotted line represents the contrast threshold prior to glare.

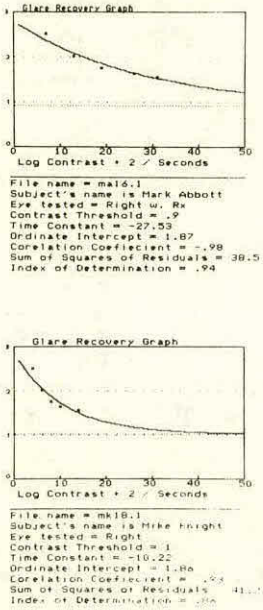

Figure 2
Examples of curves fitted to glare recovery data.

Clinical Results

We determined the norms for the instrument by testing thirty healthy adults between the ages of 18 and 34. They all had visual acuity better than 20/25, no ocular or systemic abnormalities as determined by examination by a physician. Each person in the normal group was re-tested about 3 weeks after the first measurement. The mean time constant for the right eye of the normal group was 13.4 seconds (S.D.= 6.1s). The reliability of the test was assessed by computing correlation coefficients from the results of right and left eyes and also from the results of the first and second visit. The correlation coefficient between right and left eyes was 0.82 and the correlation between first and second visit was 0.80. The test thus shows both reliability between eyes and across time.

Patient Data

Any disease process, such as decreased retinal vascular supply that interferes with the active metabolism required by the retina in order to function optimally is expected to prolong glare recovery. Diabetic retinopathy is a retinal disorder often seen in insulin dependent diabetics of long duration. There is consensus that one of the underlying causes of diabetic retinopathy is hypoxia caused by abnormal capillary bloodflow. To establish whether the glare recovery test is sensitive enough to identify early abnormalities in patients known to be at risk, we tested 10 insulin dependent diabetics. The all had long standing diabetes; the mean duration was 14 years. They all had good visual acuity (better than 20/25) and no or extremely mild visible retinal changes. The mean time constant for the group was 27.1 seconds, significantly longer than the normals ($p<.01$; T-test for unequal N). The kinetics of adaptation for this group were severly abnormal indicating retinal abnormalitity even though standard clinical tests showed no change.

Another group of 10 patients with systemic hypertension were also tested to determine whether glare recovery could be used as an indicator of general vascular status. The patients were between the ages of 30 and 55 (mean 43 years). They had been treated for hypertension for an average of 12 years. We included only the results from the eye with better visual acuity, which was 20/25 or better in all cases. The dynamics of adaptation were clearly abnormal in this group as well. The time constant was 25.0 seconds, significantly different from the 13.4 seconds for the normals. This difference is statistically significant at the 0.01 level.

SUMMARY

This study described a new device for measurement of retinal adaptation. We initially established normal values for this test which measures multiple recovery times from a single exposure to a bright light, using standardized light levels. We computed a time constant of recovery which is independent of contrast sensitivity differences between individuals and consequently free of pre-retinal media contamination. The results on the patients in this study provide evidence for functional loss prior to visual acuity loss or visible fundus changes in patient groups at risk for vascular abnormalities. This test should provide clinicians with sensitive indices at the early stages of both primary retinal disorders and secondary ocular changes produced by systemic disease.

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BACK TO FIRST GRADE
A CASE STUDY IN BLIND REHABILITATION

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ABSTRACT

Onset of severe visual handicaps concurrent with medical problems related to diabetes interfere with continued employment for many skilled professionals. This case study presents the instructional and technological interventions provided at the E. H. Gentry Special Technical Facility of the Alabama Institute for Deaf and Blind, Talladega, Alabama, designed to prepare the client for re-entry into the field of elementary teaching.

INTRODUCTION

The client was assigned to a braille class to develop skills for an alternative communication medium due to visual irregularities not defined by acuities and due to the progressive nature of the disorder. Tactual sensitivity assessment and training were of particular importance due to the reduction of sensitivity often associated with diabetic neuropathy. In addition to these physical readiness skills, psychological issues of adjustment related to the acceptance of the adaptive media needed to be addressed.

BACKGROUND INFORMATION

The client is a 42 year old female with insulin-dependent diabetes, diabetic retinopathy resulting in 20/100 OD and 20/200 OS corrected acuities, and chronic osteomyelitis of the right foot which has resulted in wheelchair dependence for mobility. The client is a former first grade teacher who requested a full range of personal adjustment service, including orientation and mobility, braille assessment, technological aids assessment and adapative ADL classes. The long range goal is for the client to resume her teaching career and independent living.

MATERIALS AND METHODS

The assessment of readiness to learn braille was conducted by the instructor through interview and the use of the Kansas Braille Reading Readiness Book. The client was found to be psychologically ready and willing to accept braille, but tactual sensitivity training on sandpaper, blocks with dot formations, trials using braille paper of different textures, and standard and jumbo dot braille met with minimal success. The client was proceeding to learn the braille patterns through verbal identification and laborious and cumbersome large, bold hand printing using her residual vision. She was using the Perkins mechanical braille writer for transcription.

In an attempt to find a practical way for this client to read braille, the Versa Braille was introduced because the dots on the braille line were of a more solid texture and could be more readily identified tactually. The Versa Braille is a micro-computer made by Tele-Sensory Systems Incorporated, with a six-key keyboard for braille, which stores material electronically on a cassette tape. The stored material can then be recalled and read tactually on a 20 cell braille display line.

The Versa Braille made braille reading possible for the client with a very high accuracy rate. Lessons from the "Illinois Series" which had been entered onto Versa Braille cassettes were utilized. The potential for teaching applications involving personal notes, documents lesson plans, lectures etc. became apparent to the client as she began to write, edit and print. In order to perform these tasks, the client had to be instructed in the use of all the word processing functions.

Transcription, at this point from audio cassette to braille using the Versa Braille, was revealed by the client to be easier when she wrote and copied from her paper visually. The instructor, himself totally blind due to retinitis pigmentosa (RP) realized that another assistive device to maximize the visual preference should be explored. Consultation from the low vision technical instructor was therefore obtained and the client was trained in the use of a closed circuit television to scan enlarged images of her printed work.

The combination of these aids and skills predicated the introduction of the Apple IIe computer, which revealed needs in the area of typing improvement and desensitization to the use of the computer itself. These obstacles were overcome through introduction of braille computer games, which combined the use of the computer keyboard and braille signs and contractions. After success in this training, the client then proceeded to learn and use the most powerful, useful and versatile word processing program for the blind, the Braille-

Edit program written by David Halladay of Raised Dot Computing.

The Braille-Edit program, a word processing program, is designed to work with the Apple IIe It works with a wide variety of screen modes, voice devices, braille devices, and ink-print printers. Braille-Edit can perform a broad range of text editing operations using voice, screen, and braille. This includes the capability to translate into grade II braille and back into regular text. When all the functions have been learned, the client will be capable of performing more word processing than a sighted counterpart because of the braille translators and the interfacing features. Understandably, training an individual to use this program required patience and dedication to the task by client and instructor, as well. The Echo voice systhesizer was used as a re-enforcer and for the instructor to monitor her work, a system which the client may need to use if her residual vision deminishes at some point. These aids may now be used by the client upon re-entry into teaching in any useful combination.

DISCUSSION

Technological innovations provide a key to unlock a promising future for scores of unemployed or underemployed low vision and blind clients through a rehabilitation effort. It is vital for those involved in applying technological aids for use in training or employment settings to remain fully cognizant of the human factors involved, and the financial resources available. The case study presented demonstrates the interaction of individualized instruction, and technological aids in a comprehensive rehabilitation program. Practitioners in the field must recognize that there is no effective cookbook solution to the complex problems presented by clients as they seek to maximize their employment potential through technology.

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3.3

A Portable Reading Machine for the Visually Impaired

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ABSTRACT

The Center for Rehabilitation Technology at the Georgia Institute of Technology in Atlanta, Georgia has developed a portable, relatively inexpensive reading machine for the visually impaired. The reading machine combines a manual optical character recognition system with a voice synthesizer to encode typewritten material into synthesized speech. Compact and affordable, the reading machine has many personal and vocational applications for the blind.

INTRODUCTION

If the visually impaired are to perform effectively and independently in the workplace, communication aids must be made available to them. Unfortunately, many communication aids are too expensive for business or personal investment. Consequently, the GT Portable Reading Machine, a compact, affordable communication system for the visually impaired, has been developed by the Center for Rehabilitation Technology (CRT) at the Georgia Institute of Technology, Atlanta, Georgia. The reading device, which converts typewritten materials into synthetic speech, has been designed to promote independent living and to increase the employability of the visually impaired, particularly the totally blind, at a reasonable cost.

BACKGROUND

Specifically, the goal of this project was to develop a reading machine for the blind that 1) is relatively inexpensive, 2) is compact and portable, 3) can read job-related materials printed on standard 8 1/2" x 11" paper as well as 8 1/2" x 14" legal sized sheets, 4) is operated by hand tracking, and 5) is easy to use.

Compared to other optical character recognition (OCR) systems, the GT Portable Reader is relatively inexpensive. Similar systems with automatic tracking range from $14,000 to $30,000. While the product is still being developed for manufacturing, it is estimated that the the GT Portable Reader should retail for less than $5,000. CRT has minimized the production costs by 1) using "off the shelf" equipment when feasible in designing the GT Portable Reader, and 2) implementing a handtracking reading system which by-passes the high cost of automation.

Basically the GT Portable Reading Machine incorporates the hand-tracked Oberon Omni-Reader OCR, which costs only $500, with the Votrax Type and Talk Voice Synthesizer, which retails for approximately $250. By adapting these two off-the-shelf devices into the GT Reader design scheme, CRT has provided the blind user with a practical, quick method of scanning and reading print by hand.

Hand-tracking is not only economical, it is practical. Hand-tracking allows the blind user to know where he is spatially on a page. Manually locating lines of print can be particularly helpful in business offices which generate spread sheets and other types of memoranda and communications that integrate verbal text with lists of numbers.

In addition, the GT Portable Reader is easy to transport and easy to operate. The portable system measures 19" x 21" x 5.75" and weighs less than 20 pounds. Blind users can learn to work the scanner and manual tracking system in a single session.

PRODUCT DESCRIPTION AND MATERIALS

The GT Portable Reading Machine is housed in a 19" x 21" plastic wedge with a 15° incline for comfortable use. Its reading surface is broader than most OCR reading surfaces; the GT Reader scanning surface measures 14" x 14" to accommodate both vertical and horizontal orientations of legal sized paper. Two replaceable tack strips, developed by Oberon, run down the face of the reading surface to hold paper securely in position during scanning. A raised lip on the left edge of the reading surface serves as a tactile guide for aligning pages to be read. At the top of the deck is a row of LED function mode indicators which are components of the Oberon Omni-Reader.

The LED indicators provide visual signals that correspond to the auditory signals produced when the character recognition scanner head, or read-head, is moved across the mode commands located on the upper left side of the deck. These visual indicators have been retained on the GT Portable Reader to aid sighted users.

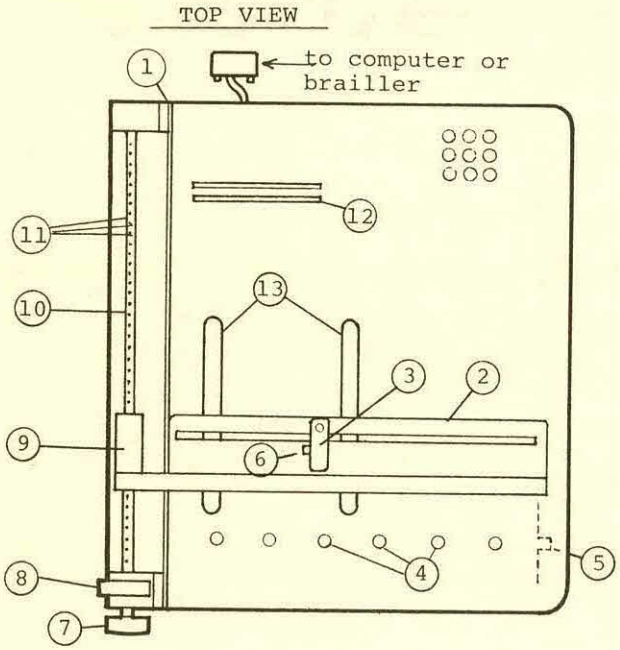
The manual tracking system attached to the left side of the read deck is made up of 1) a stainless steel 21.12" rod with holes corresponding to single, 1 1/2, and double spacing, 2) a spacing selector, 3) a threaded fine adjustment wheel, 4) the Oberon 15.5" tracking guide, and 5) a line indicator that slides along the steel rod and moves the read-head tracking guide vertically on the reading surface. A horizontal slot in the tracking guide accepts the Oberon read-head, and guides it across the lines of print.

The Oberon OCR can decode four of the most common typewritten fonts: Courier 10, Courier 12, Prestige Elite, and Letter Gothic. Attached

to the Oberon read-head is a secondary lens developed by CRT for the GT Portable Reader. The secondary lens blurs the printed image that produces gray patterns which the tone generator reads and responds to when the user is aligning the tracking system.

At the rear of the GT Portable Reading Machine is an output port for connecting either computer or electronic brailler to the reading machine.

Internally, the core of the GT Reader is the Oberon Omni-Reader, an OCR system designed to enter typed text directly from the printed page into a PC. In addition to housing the Omni-Reader micro-processor, LSI chip, Rom chips, discrete-logic ICs and transistors, the GT Reader also houses audio equipment including the Votrax voice synthesizer, speaker, and a tone generator developed by CRT to adapt the OCR system for use by the blind.



1-tactile left margin
2-tracking guide
3-read head
4-tactile bottom margin
5-tone generator switch
6-secondary lens mechanism
7-spacing selector
8-fine adjustment thumbwheel
9-line indicator
10-tracking rod
11-line indicating holes
12-mode commands
13-tack strips

OPERATION

To operate the GT Portable Reading Machine, the blind user aligns a page of typewritten material against the tactile lip on the reading surface. Next, he positions the tracking guide over the page with the read-head in the tracking guide slot.

After aligning the page, he switches on the tone generator which intercepts signals that the read-head sends to the OCR system. The secondary lens on the read-head blurs the printed image, producing gray patterns which the tone generator sees and responds to with a variable noise. The clearest signal is produced when the lens is positioned directly over a line of print, and the user must train his ear to detect the auditory signal produced by the tone generator which accurately locates lines of text.

After locating a line of print, the user searches for the next line of print. Having found the second line and determined the spacing on the page, he adjusts the spacing selector accordingly. Next, he holds the tracking guide down and turns the fine adjustment wheel until the ball plunger contained in the line indicator engages the corresponding line spacing holes on the tracking rod. Now the spacing is set and the tracking guide will stop sequentially on each line of text without the need for auditory signals from the tone generator. However, if the user should lose his place or the paper slips, he can simply realign the system using the tone generator mechanism. (This has not been a problem, however, in our test case.)

Finally, the user presses the rocker switch on the read-head and moves it smoothly across the width of the page. In a few seconds, the system decodes the scanned characters and encodes the line of print into synthetic speech. (The GT Reader can store up to six lines of print at a time so that the user can read continuously.)

If the scan is too fast for the system to read, or if a reading error occurs, the GT Reader produces a beep which signals the user to repeat the line scan.

CONCLUSION

Although the GT Portable Reading Machine, in its present state of design development, has proved to be a practical and effective communication aid for the blind, CRT recognizes that the compact, affordable reader is limited by 1) the current state of OCR and voice synthesizer technology, and 2) by the high cost of the most advanced of those technologies. With these limitations in mind, CRT has designed the GT Reader to incorporate technological advancements as they become available at reasonable prices. As science and industry begin to produce less expensive OCR systems which recognize a greater number of font styles and which eliminate the need

for sighted persons to sort reading material for the blind user by font, and as voice synthesizers are developed with better speech, the GT Reader will incorporate the updated equipment.

CRT is currently working on a more sophisticated version of the GT Portable Reading Machine in conjunction with a company in private industry. The second generation prototype is geared toward student needs as well as vocational applications.

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SOFTWARE MODULES FOR A HAND-SCANNED READING AID FOR THE BLIND

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ABSTRACT

At the VA Rehabilitation R&D Center in Palo Alto, CA, work is under way to develop a compact and versatile reading aid for visually impaired persons.[8] This aid will use solid-state imaging components in a hand scanned camera to capture print images[7], a 16-bit microprocessor to handle image processing tasks, and a custom interface board to to allow it to drive specialized displays already in use by the target populations.[6] The current paper describes three software routines developed for incorporation into the device's image processing module, for: 1) construction of continuous images from data provided by the camera; 2) location of individual characters within those images; and 3) identification of the characters.

INTRODUCTION

There are approximately 500,000 *totally blind* individuals in the United States, of which an estimated 50,000 are veterans. In addition, around 1.5 million Americans of impaired vision are classified as *legally blind*, including perhaps 200,000 veterans.[2] Most of these people have no means of unassisted access to inkprint materials, and would benefit personally or professionally from a portable and relatively inexpensive reading aid with voice and other appropriate outputs.

The objectives of the current project are: 1) to establish specific design and operational characteristics of such a reading aid via simulations using equipment developed at the VA Rehabilitation R&D Center; and 2) to produce a versatile working Computer-based Adaptable Reading Aid (CARA) prototype for field evaluations with the target populations. Historically, one conceptually similar device, the HS-1, was developed and evaluated by Telesensory Systems, Inc., and found to be too difficult to operate by the target populations.[3] Commercially, a low-cost product, the Oberon Omni-Reader, has appeared on the market which incorporates several features deemed attractive in the final CARA prototype, such as comparatively small size, low price, and easy interfacing to personal computers.[4,5] Its operation, however, appears to require a sighted user, making it unsuitable for many in our target population. These two examples point up the importance of matching device operational demands to user performance abilities, as well as establishing that the device's performance meets the needs of the user.

IMAGE RECONSTRUCTION SOFTWARE

In order to make our final device compact, a small hand-guided camera will be used for image input.[7] The blind user rests the camera directly on the page and guides it along lines of text, receiving either tactual or audio feedback to assist in the task. For processing purposes, camera images can be thought of as a series of overlapping snapshots of the text. Individual characters, words and text lines must all be reconstructed from these snapshots. Successful reconstruction requires accurate positional information of each snapshot, because there will be significant variations in the velocity and direction of the camera; it also requires image reconstruction software which operates rapidly enough to keep up with the fastest users.

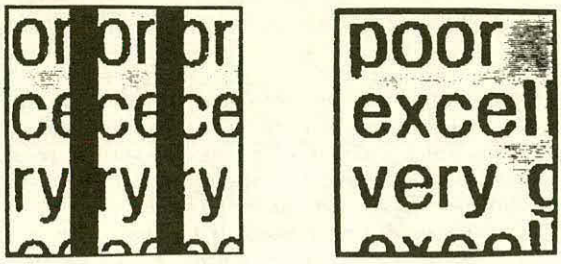
The current software for image reconstruction uses correlation techniques to determine the horizontal distance and direction that the camera has moved since the last snapshot, and a pasting routine which transfers snapshots into their appropriate positions in a Reconstructed Image Buffer. In order to extract maximum speed from the machine, the correlation routine is written in MC68000 Assembly Language, and it keeps all relevant data in the microprocessor's 32-bit registers. The algorithm operates in the following manner:

1. Several horizontal cuts, 1 bit high by 32 bits wide, evenly spaced from top to bottom, are made from the most recent snapshot.
2. Each cut, from top to bottom, is processed separately by storing it in one of the 32-bit registers of the MC68000 microprocessor. Another register is used to store the corresponding cut from the previous snapshot. These two registers are then available for high-speed comparisons.
3. The 24 left-most image data bits in each cut are used for comparisons. The number of these bits that are the same (both zeros or both ones) in both of the comparison registers is added to an accumulated total, which is also stored in an on-chip register. This particular accumulation register holds data relating to a camera movement of zero *pixel*, or picture element, widths.
4. When Step 3 is concluded, the register containing the cut from the previous snapshot is shifted left one bit, and then the comparison of Step 3 is repeated. The number of equivalent bits is added to another accumulation register, corresponding to a camera movement of one pixel.
5. The register containing the cut from the previous snapshot is again shifted left one bit, compared, and the number of equivalent bits is added to yet another register. This process of shift, compare, add and record is repeated a total of eight times. In consequence, eight accumulation registers are used to store data relating to camera shifts from zero to seven pixels.
6. The next lower cut is then processed in the same way by returning to Step 2. This loop is repeated for all cuts.
7. After all the cuts are processed, the totals in the eight accumulation registers are compared.
8. The position with the highest accumulated total is selected as the best match position, indicating how far the camera moved between the two shapshots.
9. With this information, the past routine copies the image into the appropriate section of the Image Reconstruction Buffer.

This routine takes approximately 3 milliseconds to complete a correlation for a pair of snapshots. Thus, 333 such correlations can be completed per second, meaning in practice that a user can scan a page of printed material at speeds up to 60 characters per second without performance degradation.

The two halves of Fig. 1 show the work of this Image Reconstruction Module. Fig. 1A depicts three consecutive snapshots of a page of text as seen by the imaging device. Each individual snapshot is long and narrow, allowing a view onto only a very small part of the text on a printed page. Fig. 1B shows a reconstructed image retrieved from the Image Reconstruction Buffer. The reconstructed image is of high quality, and allows a view onto a much wider area of text — one which is suitable for processing by the next module to be discussed,

whose task it is to locate the individual characters within such a reconstructed image.



A. Three Sequenced Frames B. Reconstructed Image

FIG. 1

CHARACTER LOCATION

After the incoming data frames have been spliced together and placed in the Reconstructed Image Buffer, the location of individual characters must be determined. This segmentation process is necessary since character recognition algorithms require isolated character images as input. Text is usually organized along horizontal lines with relatively large white spaces between lines and smaller white areas between words and characters.

One of the more direct techniques involves first locating the vertical position of the text lines by examining horizontal row sums to find horizontal regions with a high density of black pixels. This is done by counting the number of black pixels in every fourth row and storing the results in a list (or vector). It was empirically determined that each fourth row provided a reasonable compromise between accuracy and speed.

An adaptive thresholding algorithm is then applied to the previous results vector. This process tries to separate the text line from the white background by basing its decision on the statistics of the row sum vector. This is done in a iterative procedure which attempts to identify two discreet populations by looking at the number of occurrences of mostly black and mostly white rows and setting a decision point (the mean of the row sum vector) which will separate these two groups in a relatively distinct manner. After this decision has been made, the process continues to refine what are black and white regions by thresholding the row sum vector at the decision point and then finding the mean of the points that were classified as "black" and the mean of the points classified as "white", and setting the new decision point (threshold) to be midway between these means. This provides a new decision point that is sufficiently insensitive to noise to be effective for further processing (but will not exclude black areas that contains parts of characters).

The binary thresholded results are then passed through a median filter of width five that is used to remove spurious ones or zeroes. This works quite well since the white space between lines is relatively large. A run of ones in the binary vector is then selected as a line.

After the horizontal line of text is located within the Reconstructed Image Buffer, a similar process is applied to this selected horizontal region, using column sums to separate the individual characters. Since the separation between characters on a lines is less than the separation between lines, a higher effective resolution must be used. The result vector is not filtered, since this could cause two characters to run together.

The results of the previous segmentation process provides a rough location of the boundary of the character. Due to the nature of the thresholding algorithm, the descenders, ascenders and other appendages of a character maybe lost. Therefore it is necessary to refine the character boundaries by defining the edge of a character as the point at which the density of black pixels along the horizontal or vertical extent of the character become less than some fixed fraction. After this boundary has been defined, it is extended as necessary (within a specified distance) to make sure that significant protrusions of characters are not clipped off.

An alternate technique for character separation is also under investigation. This procedure utilizes a distinctly different process to identify the boundaries of characters. The previously described algorithm utilizes a *priori* knowledge about the structure of text on a page, while the following is more general, in that it does not make assumptions about rows of characters. This technique is well suited to implementation in advanced integrated circuit computer chips, which would allow for compact, high speed determination of the extent of characters. One can visualize this technique as growing "balloons" around characters until they meet or run off the page. The separating boundaries between characters are the balloon perimeters, after all possible expansion has taken place. If this process is done in all directions from each character, a continuous boundary is formed that encloses the letter.

CHARACTER IDENTIFICATION

Once images have been reconstructed and the location of characters within them ascertained, the next step is to identify the characters. In such Optical Character Recognition (OCR), one of two approaches is most commonly encountered — a *feature extraction* approach, and a *template matching* approach. In template matching, a character is tested against an inventory of stored shapes to determine which one it most closely matches. The approach has both recognized advantages and disadvantages. Past experience has shown that the template approach tends to be reasonably robust, straightforward, and and accurate, once it has been properly tuned. This is presumably why the template matching approach is employed in the lowest-cost commercial OCR device available today, the Omni-Reader. On the other hand, template matching tends to be inflexible, requiring a stored reference shape for every character in a given font, and failure to achieve adequate alignment of character with template during recognition can cause template match error rates to climb sharply.

The current project has selected an approach which incorporates techniques of template matching for various reasons, of both practical and theoretical import. First, when font numbers are kept low, template matching can be relatively quick to implement, yielding reasonable results early on. Second, in the evaluation phase of the project work, it will not be necessary to have an omnifont OCR capability. Indeed, many of the evaluation studies can be carried out with a capability to recognize any common single font. Third, we are looking into the feasibility of utilizing custom designed Very Large Scale Integration computer chips to carry out template matching rapidly and accurately against a large number of fonts. This approach, if successful, will contribute to removing the former major drawback of the approach, i.e., the limited font range.

Our algorithm, a hybrid approach which also contains some elements of simple feature extraction, operates as follows:

1. First, the unknown character is normalized for size. Specifically, it is fitted into a 16 X 24 rectangle. The size is chosen because storage of templates this size is convenient in digital memory. The size normalization process is useful since it provides a standardized data format for the

recognition algorithms to work on. A standard sized data format increases the efficiency of these OCR process since it presents data that is relatively independent of font size.

2. The unknown character, now contained in the 16 X 24 pixel array is broken down into sub-arrays of 4 X 4 pixels. The number of black pixels contained in one sub-array is determined, and subtracted from the number of black pixels contained in the corresponding sub-array of the template character, yielding a difference.
3. This difference is squared and added to a cumulative distance index (which is initialized at zero when starting each comparison between the unknown character and a new template).
4. The algorithm moves to each of the remaining sub-arrays in turn, repeating steps 2 and 3, until all sub-arrays have been processed for a given template. The cumulative distance index at that point represents the square of the distance in n-space ($n = 24$) between the unknown character and template. This approach reduces the problems caused by slight mis-registrations between the unknown character and the template.
5. The second index of dissimilarity is calculated by imposing a grid of lines with a spacing of 2 X 2 pixels. For each vertical and horizontal line, the number of white to black, and black to white, edge crossing pairs is counted.
6. Averaged numbers are obtained for sets of four neighboring columns, and compared against the corresponding numbers from the stored template.
7. Differences are calculated and sums of squares are kept, in order to determine the distance between unknown and template in n-space based on edge crossings. Here too, the approach reduces problems that arise from minor misalignment and noise in the picture.
8. The two indices of dissimilarity are combined into a unified index of dissimilarity. These are then compared against one another.
9. The template against which the unknown character registered the *lowest* index of dissimilarity is chosen as the identity of the unknown character, unless the index is above a certain level, in which instance the program reports it has no confidence in its choice.

Work on our OCR modules is still in its early stage. Using the algorithm described above, in laboratory trials we have demonstrated an ability to recognize letters of a single font with approximately 95% accuracy, providing the scanned characters are of good quality and the solid-state camera is properly focused and aligned. While this is deemed insufficient for incorporation in a reading aid for the blind (97% is considered the minimum acceptable[1]), work to improve the performance of the OCR modules is continuing.

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SOME NEW VOCATIONAL AND COMPUTER-ACCESS SENSORY AIDS

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ABSTRACT

This paper describes the development of a new family of vocational sensory aids to provide more flexible instrumentation of the blind employee's workplace. New aids to computer access and information processing for the blind are also described. These include an auditory data flow indicator, an auditory breakout box, new braille display technologies, and an electronic braille notetaker.

INTRODUCTION

The Smith-Kettlewell Rehabilitation Engineering Center has recently undertaken research into several pressing problems in vocational rehabilitation. Firstly, the difficulties involved in providing specialized job instrumentation for the blind have been addressed through the development of flexible microprocessor-based aids. To address computer-access difficulties for the blind, auditory aids to easier interfacing of computer peripherals have been developed, and a comparative study is under way of two promising technologies for new low-cost volatile braille displays. Finally, traditional braille notetaking methods are being radically improved upon through the development of an electronic braille notetaker.

PROGRAMMABLE JOB INSTRUMENTATION SYSTEM FOR THE BLIND

Background

In many job situations where a blind person is confronted with a need to access information which is normally available only visually, a sensory aid of one description or another is needed. With the exception of certain employment categories such as those involving computer operation, no appropriate generic information-access systems exist for the blind. Consequently, the traditional solution has been to custom design a sensory aid to suit the particular employment situation. For example, a blind refrigeration engineer must obtain some means of reading his pressure gauges, and the normal approach would be to design a sensory aid specifically suited to that purpose, and geared to his particular instruments. This type of sensory aid, which could have an audio-braille or a speech output, would more than likely have to be designed "from the ground up," involving considerable expense.

In fact, the need for extreme specialization in vocational sensory aid design has resulted in the potential market being poorly served. The economies of scale cannot be achieved, and indeed few of these devices even reach production, due to the high unit cost and the small market size for each such device.

In an attempt to overcome this problem, we have devised a microprocessor-based approach to job instrumentation for the blind which allows considerable flexibility in application, so that the one basic design can, with software modifications, serve a number of needs.

Programmable Switchboard Monitoring Systems

First, development in this direction was directed at the typical problem of a PBX switchboard operator who may be required to handle, and monitor the status of, anything from ten to sixty incoming lines. Of course, the switchboard with less than ten lines can be made accessible to a blind operator through the use of a simple light probe. However, switchboards with greater numbers of lines typically require extensive custom-made instrumentation.

As an experimental subject, a client was referred to us through the Sensory Aids Foundation. This individual was required, as part of her job, to operate a thirty-line switchboard at a local bank. Our solution to this problem was to interface the indicator lights on the incoming lines with a Certek SBC 85-1 microcomputer connected to a National Semiconductor Digitalker speech output device. A block diagram of the completed system is shown in Figure 1.

In operation, the microcomputer scans the incoming lines sequentially, calculating the status of each from the actions of the indicator lights. This information is stored and updated every second.

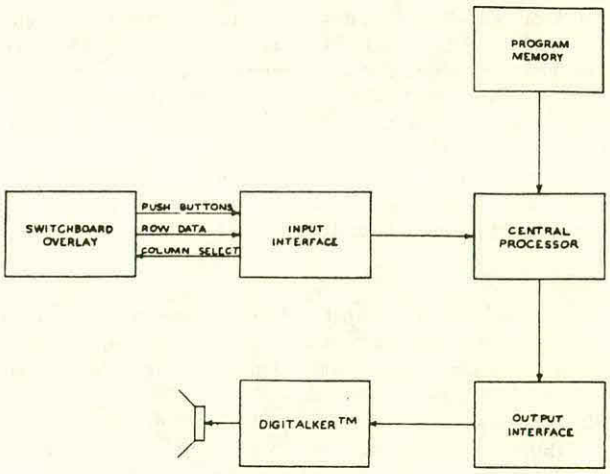

Figure 1
Switchboard Monitor Block Diagram

The device is programmed to announce incoming calls as they arrive, indicating the number of the line upon which the call appears. The user can determine, at any time she wishes, which lines are busy or on hold by pressing a "busy" or a "hold" button (which are the only user controls required). For example, if the user depresses the "busy" button, the computer immediately announces, in numerical order, the numbers of the lines which are busy.

The system software, stored on an EPROM, is written in such a way that no modification is required if the number of lines is increased or decreased, up to a limit of sixty-four. The particular PBX used in this example is an AT&T Horizon model, which is commonly used in industry and handles up to thirty lines.

The interface between the switchboard and the computer, in our initial experimental model, was by means of an overlay containing thirty photo-transistors which directly sensed the status of the light corresponding to each incoming line. This approach was chosen in order to not interfere with the wiring of the existing telephone systems as installed. However, an improved model to be developed will interface directly with the standard telephone cable system, using opto-isolators to eliminate any possiblity of interference with the installed telephone system. Using this approach, no mechanical components such as the overlay mentioned above are needed, and the cost of interfacing the system to any standard telephone system becomes minimal.

The initial prototype was successfully tested on the job and found to give complete satisfaction. We anticipate that the improved model now under development will permit convenient adaptation of any of a large variety of industrial PBXs for use by the blind operators at relatively low expense.

Refrigeration Pressure Gauge Systems

The second example chosen for attention utilizing the approach outlined above was that of a refrigeration engineer who had lost his vision and needed instrumentation to allow him to read several pressure and vacuum gauges.

Using the same single-board computer and speech output device as in the first example, we developed an input interface allowing connection of standard industrial pressure and vacuum transducers via analog-to-digital converters. In this instance, the microcomputer's EPROM was programmed with instructions to cause monitoring of three analog inputs, any one of which could be read out at the command of the user. Once again, user control merely required the activation of a switch corresponding to the desired information.

Generalization of the System

The next stage in development, currently under way, is to design a modular programmable job instrumentation system utilizing the same general architecture as the examples above, but with a wide variety of input interfacing options. These options will include serial digital, parallel digital, analog and raw Linear Variable Differential Transformer (LVDT) transducer data, and will be implemented by designing appropriate "interchangeable plug-in" circuit boards housed in the same chassis as the microcomputer, speech output system, and user controls. Standard software packages, contained on interchangeable EPROMs, will be designed to accommodate as many options as practicable, while custom software modifications, if necessary, could be carried out inexpensively.

Thus, a typical application may require the monitoring of two parallel digital channels and an analog channel. Through appropriate coding of the circuit board edge-connectors, the micro-computer would be alerted to the type of input (if any) interfaced to each of the available channels. Any special input scaling or output vocabulary needed for specific applications would be accomplished through simple software modifications. To sample the status of any input, the user need only depress the switch corresponding to that input, and a value would be enunciated verbally.

We believe that this approach should greatly increase the feasiblity of commercial production of specialized vocational instrumentation for the blind, by making it possible to use the same basic system for a large variety of different applications - thus increasing the economies of scale to the point where commercial production should be feasible. In turn, the commercial availability of such a system, which is adaptable to such a variety of special job situations, should increase the number of such job placements available to blind persons.

AIDS TO COMPUTER ACCESSIBILITY AND INFORMATION PROCESSING

Auditory Data-Flow Indicator

Several electronic devices to enhance computer accessibility for the blind are under development in our laboratories. An auditory data-flow indicator was developed at the request of the Kentucky Bureau for the Blind, and has now been produced commercially as the "Tweedle Dump." This device was designed to provide an indication to the blind user when data transfer is in progress in an RS-232 serial interface, aiding him in knowing whether the interface to printers and other peripherals is operating normally and when the computer is free to receive other input. It is especially useful with modem operations in alerting the user when a long data transmission has finished. The device consists of a simple audio transducer built into a standard interface cable, and driven directly from the data signals.

Auditory Breakout Box

Interfacing the large number of different computers and peripherals now available presents special problems for the blind user. Sighted individuals have access to breakout boxes which can be connected between the computer I/O port and a peripheral. Each input signal line can be connected, through use of jumpers, to any output line, and the device will indicate (through activation of LEDs) when the correct connections

have been made. To make this ability available to blind individuals, we have developed an Auditory Breakout Box (Figure 2), which provides a similar convenient means of trying different connection combinations, and utilizes a tonal sound coding system to indicate whether any selected line is grounded, high, low, or open. This device has been successfully tested by several users, and discussions have been initiated with a potential manufacturer.

New Volatile Braille Display Technologies

Braille computer-access aids are widely preferred by braille-reading blind users to those using synthetic speech, but hitherto these devices, utilizing volatile braille displays, have been relatively expensive and not without reliability problems. We are currently exploring new approaches to the production of refreshable braille displays with a goal of developing a reliable device which could retail for less than $500. The use of inexpensive electromagnetic technology, avoiding the close tolerances and wear problems of conventional solenoids, is being compared with the application of memory metals for low-cost display production. We have commissioned a study by Dr. David Johnson on the feasibility and cost-effectiveness of the latter approach, and initial results are promising.

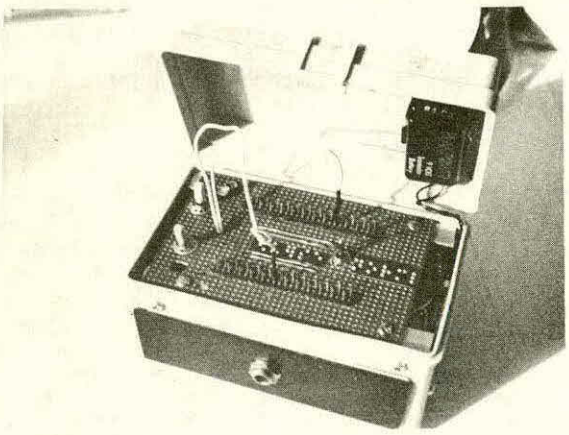

Figure 2
Auditory Breakout Box

Electronic Braille Notetaker

Another development being explored in the area of braille and computer utilization technology is a pocket-sized electronic braille slate. The "Note-a-Braille" is a battery-powered device with an 8-key (plus spacebar) braille keyboard and an 8k-byte CMOS RAM in which the data entered from the keypad are stored. A built-in EPROM converts the braille code to ASCII for downloading into the user's personal computer, an operation which can be done at leisure. Notetaking using this device is both faster and quieter than using a braille slate, and provides a convenient method of "bringing one's computer to class." Because there are thousands of blind people familiar with the simple braille keyboard, advantage can be taken of its small size (compared to a conventional keyboard) for use as a computer input device.

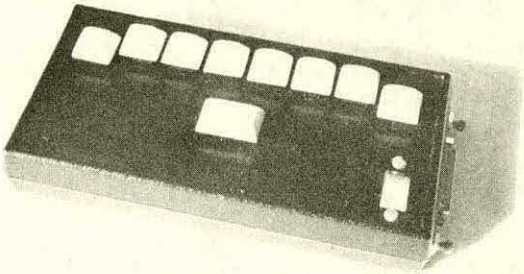

Figure 3
"Note-a-Braille"

Sufficient memory is provided to store approximately eight pages of braille. Extra keys are provided on the keyboard to provide directly such functions as capitalization and 8-dot computer braille codes as desired by the user. Output is compatible with a standard Centronics Parallel Port, and a parallel-to-serial converter is provided as an option to allow direct connection with standard RS-232 ports. The first prototype of the system has been completed (Figure 3) and is undergoing user testing prior to miniaturization and packaging in a truly shirt-pocket-sized enclosure measuring approximately 6 by 3 by 0.6 inches.

Acknowledgements

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DEXTER - A HELPING HAND FOR COMMUNICATING WITH THE DEAF-BLIND

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ABSTRACT

Deaf-blind people who receive information through tactile finger spelling are forced into a dependency which is at best limiting and inconvenient, and which is usually detrimental to their social interactions and vocational fulfillment. To alleviate some of these problems, the Smith-Kettlewell Rehabilitation Engineering Center sponsored a project in a graduate level class in the Mechanical Engineering Department at Stanford University. The result is Dexter, an electro-mechanical device which forms the finger positions of the one-hand manual alphabet. This device has application in facilitating interpersonal interactions, obtaining computer-based information, and promoting telephone communication for deaf-blind persons.

INTRODUCTION

The double sensory loss which defines "deaf-blindness" precludes the use of the main sensory channels for receiving spoken and written language. The resulting social and informational void causes extreme isolation, frustration, and dependence upon others. These problems are somewhat diminished through the use of tactile adaption of the manual communication systems of finger spelling (Figure 1) and/or sign language originally developed to communicate with deaf people.

Unfortunately, not all members of the minuscule group who are familiar with manual communication systems are willing to employ them with individuals who must feel their hands to receive the information. Not only is the tactile contact physically restricting and fatiguing, but it also requires a closeness some find discomforting. Similarly, such interactions limit the deaf-blind person's communication to individuals who are within an arm's reach. Additionally, while tactile versons of the manual communication system may provide deaf-blind people with some degree of substitution for spoken communication, they provide no assistance in accessing written information.

Sensory aids designed to improve the quality of life of the deaf-blind would enable virtually all people to "talk" to them, would allow their interaction to be remote, and would provide access to written information. This would have to be accomplished through information displayed tactually. But in a world where things electronic are promoted over things mechanical, and the freedom of distance is valued over the intimate restrictions of physical contact, precious few devices with tactile displays have been developed.

While braille is a ready-made solution for some deaf-blind people, these are in the minority. The major cause of deaf-blindness is Usher's syndrome, a disease which typically causes congential deafness followed by loss of vision starting in early adulthood. As deaf children, Usher's syndrome patients generally learn to use finger spelling and sign language for interpersonal communication, and print for remote communication (i.e. both for reading and using a telecommunication device for the deaf [TDD]). Their visual loss in adulthood deprives them of using either visual communication system. Yet learning to read braille is often not an acceptable solution to those who have lost their sight as adults (even when deafness is not part of the picture). It is much more comfortable to continue to use the familiar manual communication, despite the restrictions of receiving it tactually discussed above. Learning braille as an adult is a formidable task. Furthermore, it is often frought with negative emotional overtones as learning to read braille is commonly considered the ultimate admission of being blind.

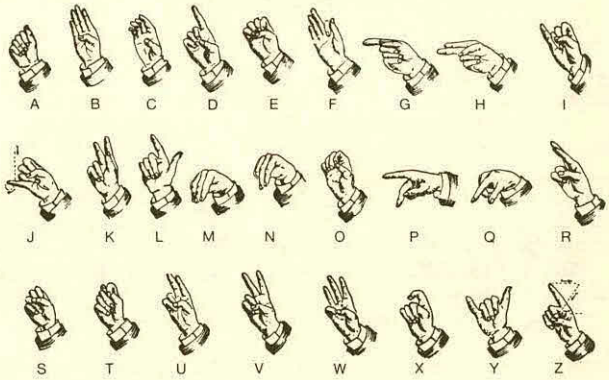

Figure 1. The one-hand manual finger spelling alphabet.

BACKGROUND

In 1977 the communication problems of the deaf-blind were tackled by engineers at the SouthWest Research Institute in San Antonio, who conceived of and fabricated a mechanical finger spelling hand. [2] Their device, like the one reported on here, was designed to form the symbols of the one-hand manual alphabet (Figure 1) on command from a standard keyboard. Despite certain limitations such as an inability to properly form all of the letters and the lack of fluidity of motion which seems to enhance comprehension of tactile finger spelling, the SwRI device was a stunning success in demonstrating the feasibility of the concept - the mechanical hand did indeed provide a vehicle for transmitting information to deaf-blind users. Unfortunately, however, any changes in the way the letters were formed had to be accomplished through hardware modifications, limiting its flexibility for research.

The current project builds upon the experience of the SwRI device and contains features which overcome deficiencies noted by the original users. In particular, the adjustability of finger positions and timing considerations have been addressed.

MATERIALS AND METHODS

The mechanical portion of Dexter consists of five fingers and a thumb joined together at a palm. Each finger operates independently of the others and has a range of motion comparable to that of human fingers (Figure 2). The thumb has two joints similar to those of the fingers. These joints allow it to sweep across the front of the palm. In addition, the thumb is mounted on a pivot joint which allows it to move in a plane perpendicular to the palm. The palm, in turn, is supported by a vertical steel rod which rotates to provide the hand positions necessary to form some letters. A pneumatic rotary actuator provides this motion.

The fingers and thumb are flexed by drive cables. Pneumatic cylinders pull these cables which flex the fingers and thumb while spring-driven return cables open the finger joints to the extended position. The actuating equipment and valving are housed in two separate assemblies below the hand itself.

A microcomputer and accompanying software control the opening and closing of the valves which operate the pneumatic cylinders. Each letter is formed by a timed sequence of these valve operations. Presently, the hand starts each letter from and returns to a partially flexed "neutral" position. Each letter's ASCII value is used by the software as a pointer into an array of stored valve control values. Since 22 valves are controlled, three bytes are allotted to specify the state of each at successive time intervals.

The original student design included an Intel 8085-based STD buss "target system" used in ME218 (Smart Product Design Course) at Stanford. It consisted of the 8085 microcomputer, Forth programming language, memory, and timer support. The timer generated signals that determined the rate of hand motion and how long each position was held. The additional circuitry needed to control Dexter was fabricated on an STD card which plugged into the target system card cage. A single external 12 volt power supply provided the power to activate the 22 valves under computer control. Port latches received the valve control words from the CPU, while Darlington power transistors provided sufficient current to control them. Letters to be displayed on the hand were entered on an IBM PC keyboard which was connected by a serial link to the target hardware.

Current activity is concerned with the replacement of the Stanford computer system with a Zilog Z80-based STD microcomputer system with electronic boards donated by Prolog Corporation. Commercial medium-power DC controller cards replace the student-built wire-wrapped version. The software has been rewritten to 1) accomodate changing of critical parameters such as timing variables, 2) allow characters to be accepted while previous ones are being displayed, 3) permit easy modification of the finger movements for any letter, and 4) incorporate both modem and serial input of characters.

RESULTS

Two deaf-blind clients of the Lions Blind Center in Oakland, California served as subjects to initially test Dexter (Figure 3). Within minutes they were able to "read" the letters, words, and sentences formed by the hand. It was apparent that they were delighted and intrigued with this novel communication system.

DISCUSSION

The results so far are extremely encouraging and suggest that Dexter is a promising communication device for the deaf-blind.

Extensive evaluation of Dexter will be undertaken. Finger positions will be optimized for easiest recognition, and the hardware will be reduced to a minimum. A low cost, simple design to accomplish intelligible finger spelling with a three fingered version of the manual alphabet will be attempted. The use of stepper motors and a memory metal will be investigated as substitutes for the pneumatic drive mechanism. The next iteration will include the ability to translate TDD signals and a general computer interface for vocational and recreational purposes (Figure 4).

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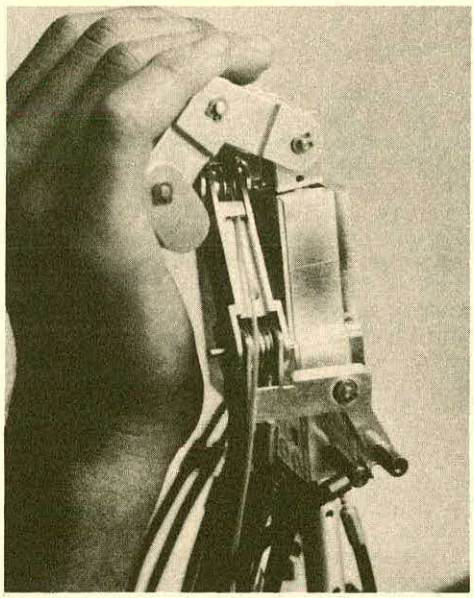

Figure 2. Side view of Dexter.

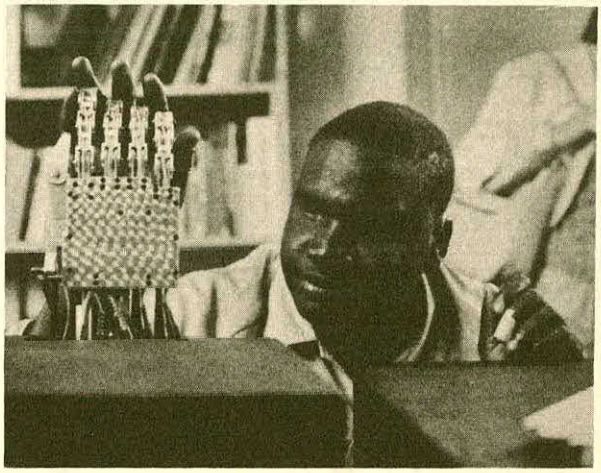

Figure 3. Deaf-blind client using Dexter.

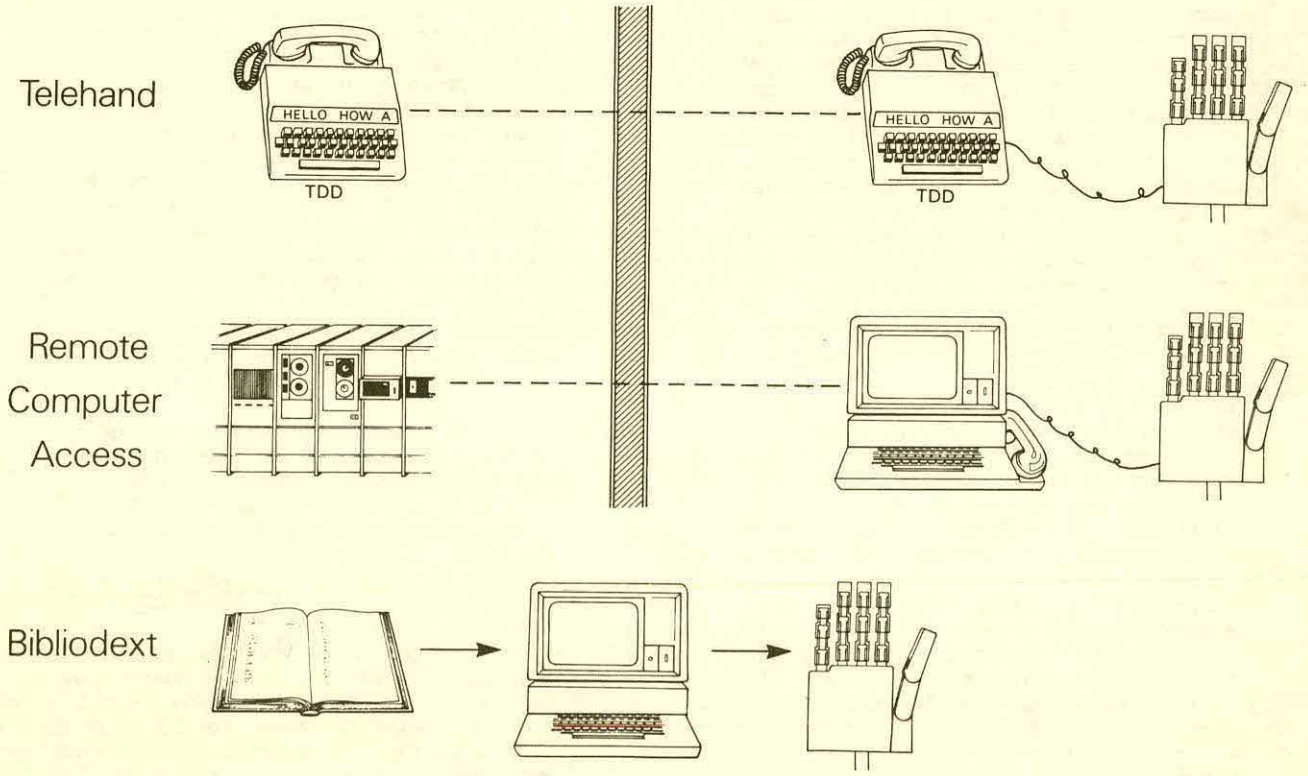

Figure 4. Potential applications of Dexter.

DESIGNING ASSISTIVE DEVICES FOR DEVELOPMENTAL TASKS IN RECREATION

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ABSTRACT

Recreation or play is an important part of a child's psychomotor, social and emotional development. When a disability affects access to and manipulation within the child's environment, all aspects of development are affected. Accessing the play environment must address developmental needs at each age and devices are designed and modified with special consideration to age appropriate activities and the particular disability. In adulthood, play becomes recreation and continues psychomotor stimulation, social interaction and emotional release, all necessary for the total health of an individual.

INTRODUCTION

Recreation or play is one medium through which a child as a developing person learns about interacting with his physical environment, getting along with others and how he feels about himself. Play is an inborn drive and from birth exploration and sensorimotor stimulation provide impetus and rewards encouraging its continuance. Disabilities which limit mobility or cause developmental delays affect the physical, intellectual, social and emotional growth of the child (3,4). At Gillette Children's Hospital along with traditional orthotic and prosthetic prescription, additional care is taken to provide assistive devices which allow children to experience and explore their world as normally as would their peers.

DISCUSSION

Three to nine months

During the first 3-9 months of life, play consists of coordinating hand and eye movements in grasping, mouthing objects grasped, examining objects visually and listening to the sounds they make. A child who cannot reach or perhaps even see objects nevertheless needs to play with them in similar manners. Sitting supports can be easily modified to put toys or objects to be explored within reach of the child by attaching a hoop above the seat. The design should also allow the seat to be used on the floor in the nursery as well as in the high chair for feeding.

By manipulating various toys, a child learns causes and effects, relationships between actions performed on objects and the results, and various characteristics of things (2). A severely involved child with motor and developmental disabilities can appreciate toys and learn from using them. Special momentary switches, commercially available, interface with battery powered toys which come with more complex switches, allowing operation by more gross body movements. Head control of the toy can be achieved using variations of such a switch.

The repetitiveness of play reinforced by pure joy establishes cognitive patterns and can enhance motor control (2).Elements learned during play have more practical applications to be built upon for more functional tasks, perhaps accessing through microprocessors, a communicator or environmental control. The potential is limited only by the child's cognitive abilities which can also be enhanced by the range of play experiences.

Nine to twelve months

Between 9 and 12 months mobility becomes an important part of a child's play activities, first with creeping, leading to standing and walking. We often complain about children this age "getting into everything", but that's one quite enjoyable way of learning about everything. For children with involved extremities, supports mounted on casters are designed to allow them to move about the floor indoors with minimal effort, at speeds comparable to their peers, at the same heights as their peers and have independent access getting on or off. These may be powered with hands and/or feet or anomalous limbs and allow the child to explore this environment at will.

Toys and lower limbs may be supported on the hand propelled caster carts which are easily maneuvered indoors. Without crutches the hands are free to grasp and objects may be transported on the cart without jeopardizing support or mobility. T-foam padding can be used to prevent sores if necessary.

For a child with Lesch-Nyhan syndrome, independent mobility for exploration and socialization must address safe mobility by preventing the self inflicted oral destructiveness of this syndrome. A specially adapted wheelchair along with dental tooth guards allows for independent propulsion and

use of the hands on the lapboard while preventing hand to mouth motions. The cuffs on his wrist track along the rails mounted in the armrest sockets for propulsion and also allow positioning of his hands on the lapboard for drawing or accessing push buttons to the computer used in his school.

Twelve to eighteen months

Between 12 and 18 months discovering and imitating the usefulness of things becomes important. Ambulating in an upright posture, pushing and pulling wheeled objects enter into play activities. The Gillette Crutchless Standing Orthosis and other parapodiums allow the child to participate in these areas of play with hands free to be used in activities other than supporting the upright posture.

Thirty months

After 30 months, the child begins to to broaden his interactions outside the home. There is much wonder in the moving perspective from a swing or merry-go-round. Sitting supports designed for a variety of uses, such as a seat insert rather than a modified wheelchair, enable a child to participate in some of these outside play activities.

Socially, they begin at this age to interact with other children in the neighborhood (5). At first, under the watchful eye of a parent or teacher, and later around the block independent of close adult supervision as they learn to abide by the rules of the game (1) and interact socially with their peers.

Outdoor mobility becomes more important as they must keep up with the other kids. Children with disabilities may require modifications to their tricycles or big wheels to give them a physically safe yet socially challenging experience through which to develop coordination and communication skills. They may be simply sissy bars and foot straps or perhaps more thoracic support.

For children with lower limb deficiencies, hand powered devices are commercially available. Some are low to the ground for independent access. Some are designed to "grow" with the child. The Unicycle attaches to a standard wheelchair. A whirly wheel was adapted with a sitting support for a blind child.

Using self-proprelled devices gives the child independent mobility which allows him to exercise his body enhancing motor as well as social and cognitive functions. They practice reciprocating movements, increase their strength and spatial coordination. These are all very desireable benefits, but there are times when the energy and emotional expenditure is too costly. In such instances, powered outdoor mobility is provided by modifying sitting support and controls on battery powered toys, or modifying bicycles or tandem bikes that are powered with a friend.

Adulthood

Play becomes recreation as we become adults and it gains attributes directed at self achievement and emotional release. Skills learned in play as a developing individual can be carried into the challenges of adulthood. Together with Greg Lais and others at Wilderness Inquiry II, a pulk sled was modified with outriggers to increase stability and efficiency in winter recreational transportation and a people pack for carrying people over portages during canoe trips has been designed. Ice skate blades to accommodate leg length discrepancies have also been made. This is recreation, in leisure activities, for a lifetime.

CONCLUSION

Play is the means by which children acquire tools that will allow them to function in the real world as adults. They learn about their physical environment and how they can control it. Future skills are practiced in a non-threatening situation where mistakes don't count as much. They learn to control their bodies, develop their coordination and strengthen their muscles through enjoyable and challenging repetition. Mastery of such tasks promotes a positive self concept and confidence in their ability to solve problems. By interacting with other children they learn how to get along with others, how to work together towards common goals. Individual talents can be discovered and appreciated. Finally, recreation will provide them with leisure time activities that can give them a lifetime of emotional release from everyday stress and improve their quality of life. Designing equipment which meets developmental needs for play and recreation at each stage can greatly affect a person's potential and quality of life.

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GUIDELINES FOR THE USE OF AUTOMATED LEARNING DEVICES BY PERSONS WHO ARE MULTIPLY HANDICAPPED

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This paper describes a comprehensive approach to the selection and use of automated learning devices by persons who are multiply handicapped. The approach is based on observations of optimal learning situations and includes a philosophical base in addition to considerations for the identification and integration of information about purpose, user position, potential control sites and device options.

Enabling technology of the 80's is pervading home, school, and community settings serving multiply handicapped children and adults. Automated learning devices represent one segment of enabling technology used most often by this population.

Automated learning devices (ALDs) are simple electro-mechanical systems used to control appliances or battery operated toys. A typical ALD system consists of a switch (interface), a control unit, and an appliance; or a switch, a battery interrupter, and a toy. Automated learning devices function primarily as learning tools and as means of enviornmental control for persons who are multihandicapped.

Currently, a wide variety of ALD components are commercially available from many different manufacturers of adaptive equipment. The vast array of automated learning devices creates a challenge for parents and educational or therapeutic service providers who are faced with decisions about WHETHER to use automated learning devices and WHAT automated learning devices to use. These decisions are considerably more difficult to make for persons who are multiply handicapped. Unreliable motor movements, reflex interference, and perceptual or cognitive deficits are among the complicating factors.

Working with teachers of multihandicapped children, the authors observed what appeared to be a common phenomenon in classrooms with access to automated learning devices. One of two situations usually prevailed; either the ALDs were "on the shelf" and not being used at all or, the devices were being used by the children for one function (e.g. independent activity). These observations prompted the authors to study the situation for possible remedies. In most cases the decision to provide ALDs to a specific child was made without systematic evaluation of the purpose of its use. It was assumed that all of the children could benefit from using a switch, and thus, a switch was provided. When teachers reported they were confident that their students were successfully using ALDs, similarities in how decisions were made WHETHER to use the devices and WHAT devices to use were noted.

A comprehensive approach to the selection and use of automated learning devices was developed based on observations of optimal learning situations. The approach includes a philosophical base in addition to considerations for the identification and integration of information about purpose, user position, potential control sites and device options

PHILOSOPHY

The philosophy statements which follow represent a set of beliefs which guide decisions about device selection and implementation. The comprehensive approach stems from these beliefs about the use of technology by persons who are severely handicapped.

1. Automated learning devices are TOOLS which can be used to facilitate learning and to increase independence through environmental control. Thus, ALDs are means to ends rather than an end in and of themselves. Functional interaction with the environment and improved skills in domestic, leisure, community and vocational domains are the ultimate ends or goals for the use of ALDs by persons who are severely or multiply handicapped. Automated learning devices should be selected with an identified PURPOSE as well as the handicapped individual's unique disabilities in mind.

2. It is recommended that decisions to use ALDs be made, whenever possible, by a transdisciplinary team including parents, educators, therapists and other appropriate support staff. The team approach is most critical for clients with severe or multiple handicaps when information from several disciplines must be evaluated.

3. Automated learning devices are one possible technological solution and should be considered along with other forms of enabling technology or adptations. The final decision whether or not to provide ALDs to a given client may ultimately depend upon factors such as device availability, accessability, restrictiveness, cost, portability, ease of maintenance, similarity to devices a non-handicapped person would use, etc. It is important to keep these factors in mind when comparing several possible technological solutions. It is not uncommon or inappropriate for persons who are handicapped to use more than one form of technology. For example, many children use a computer as a learning tool and ALDs to control favorite toys during leisure time.

WHETHER TO USE ALDs

Decisions to use automated learning devices should be based on the needs and capabilities

of the handicapped individual. The following steps may help determine whether ALDs would be appropriate for a multiply handicapped individual.

1. Define user needs in terms of a functional objective.
Specifying the purpose(s) for using ALDs if often assummed or overlooked in the decision making process. Optimal decisions about device selection and use appear to depend upon a clear understanding of the purpose(s) for using ALDs as it relates to the handicapped individual's unique abilities.

This step involves clear definition of individual short or long term objectives that will allow for greater participation in the real world.

2. Identify the skills needed to achieve the objective.
This step involves consideration of the motor, cognitive, or perceptual abilities that a non-handicapped person would need to successfully achieve the objective.

3. Define the barriers which prevent, restrict, or limit the individual from developing the skills needed to meet the objective.
Barriers may be indiviaul or environment related and usually represent a lack of necessary skills defined in step 2. For example, lack of fine motor ability is a barrier which prevents some handicapped individuals from manipulating the dials on tape recorders or radios.

Interferring behaviors such as self stimulation may be barriers which limit functional interaction with the environment for some persons.

4. Define interventions or solutions which may be used to overcome the barriers identified in step 3.
Interventions will include a variety of educational or therapeutic methods. Consider each barrier individually as well as the individual's abilities when completing this step.

5. Determine whether ALDs can be used as a tool to achieve the functional objective using information from steps 3 and 4.
This is a YES or NO decision. It is not necessary to determine what automated learning devices to use at this time. Consider the following when completing step 5:
-Is ALD the least restrictive option for the individual or can the objective be met with a simple adaptation:
-If barriers include lack of a reliable motor response or lack of environmental awareness/interest consider focusing on developing these skills first. ALDs may be used to facilitate the development of a motor response and interest/awareness.

WHAT ALDs TO USE
Once the decision is made to provide ALDs to a handicapped individual the next step in the decision making process is to define specifically WHAT automated learning devices will be provided. Appropriate selection and effective use of ALDs by persons who are severely or multiply handicapped depends upon several critical factors including appropriate selection of an activity, a control site and proper positioning of the user.

Critical factors for ALD selection
Activity selection. The activity refers to the toys, appliances, or materials used in the learning situation. Activity selection is based upon the identified purpose for using ALDs. Additional factors to consider include the user's interest, perceptual abilities and age.

User interest is the most important consideration in activity selection. Many multiply handicapped individuals will not make the required physical effort to interact with unmotivating materials. Interests may be evaluated by systematic exposure to a variety of toys, objects, or activities and observation of affective responses prior to providing ALDs as a means of control. Some severely involved children and adults do not appear interested in attending to or interacting with the environment. Often this apparent disinterest is actually a result of their own lack of control. Interests may develop when these individuals are provided with opportunities to control toys or appliances using ALDs. When the intended purpose is to determine activity preferences, ALDs may also be used as a tool to achieve this end.

Proper positioning of the user. The users position can greatly affect his/her ability to interact with the environment using ALDs. The user should be in a well supported position which allows for concentration on an isolated motor movement rather than on control of the total body. Specific positioning techniques may also help to control abnormal movement patterns and extraneous movements. Decisions about optimal learner position and control site selection are often made simultaneously due to the need to integrate information in the two areas. Typically, a stable position is identified first and adjustments in positioning are made as the motor movement is evaluated. More than one optimal position may be identified for an individual using automated learning devices.

Appropriate control site selection. The motor movement used to activate the ALD system should be RELIABLE (reproducable without enormous physical effort) and VOLUNTARY (intentionally controlled by the user). A reliable motor movement or control site is often easy to identify. Examples include handicapped individuals with good gross motor control of their arm or hand or individuals with isolated movement of other body parts such as a foot or knee.

When the user has difficulty initiating any movement, is dominated by reflex patterns, or has extraneous motion; identifying a reliable response can be very difficult. In this situation, a systematic observation of the user's movements and abilities may help identify a potential control site.

Selection of automated learning devices. Potential ALD components are identified after careful consideration of the above critical factors. The type of switch including special switch features (e.g. sensitivity, feedback, color, size, durability) and the need for a control unit, a timer for controlled reinforcer segments, adaptors, or data collection capabilities are among the decisions which must be made.

GUIDELINES FOR THE USE OF ALDs

A tool was developed by the authors to assist educational and therapeutic teams in selecting the most appropriate ALDs for children and adults with severe or multiple handicaps. Guidelines For The Use Of Automated Learning Devices is based on the comprehensive approach to the selection and use of ALDs outlined in this paper. The tool consists of considerations and if/then rules which help the reader identify possible situations to explore integrating information about activity, user position, control site and ALD components. The ALD guidelines are available from ABLENET, 360 Hoover St. N.E., Minneapolis, Minnesota 55413, USA.

THREE DEVICES FOR USE IN THE SPECIAL EDUCATION CLASSROOM FOR PHYSICALLY IMPAIRED STUDENTS

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ABSTRACT

The Rehabilitation Engineering Center in Wichita, Kansas, has had extensive experience in prescribing devices for use in the physically impaired classroom. Often the devices are ready-made and can be purchased off-the-shelf. Many occasions have arisen, however, that called for special-made devices that have been designed and implemented by staff of the Center. Three such special devices are presented here.

INTRODUCTION

The Rehabilitation Engineering Center (R.E.C.) in Wichita has been principally funded by the National Institute of Handicapped Research and is sponsored by the Cerebral Palsy Research Foundation of Kansas, with a cooperative relationship with the college of engineering at Wichita State University. Some work is done on a fee-for-service basis by members of the staff who are not wholly funded by research. One of the areas of service delivery has been with the public school system in Wichita. Several devices to be used by the teachers, therapists, and aides have been devised and implemented in one of the classrooms for physically impaired students. These devices are used both for teaching aids and for helping to provide aide and attendant care for the students. The three devices that are presented in this paper are examples of the aids that have been implemented.

DEVICES

Adjustable Chalkboard

Chalkboards that are mounted flush with the surface of the wall are inaccessible to students or teachers who are in wheelchairs. A height-adjustable chalkboard which stands off from the wall for wheelchair footrest clearance was designed, fabricated, and installed in an attempt to solve the problem of inaccessibility. See Figure 1. Staff

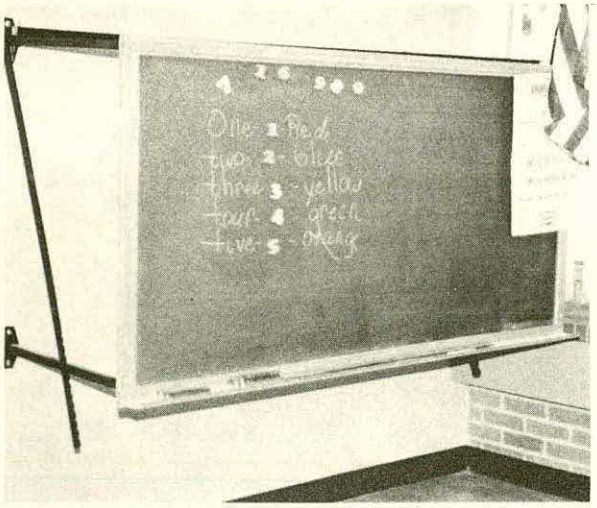

FIGURE 1: ADJUSTABLE CHALKBOARD

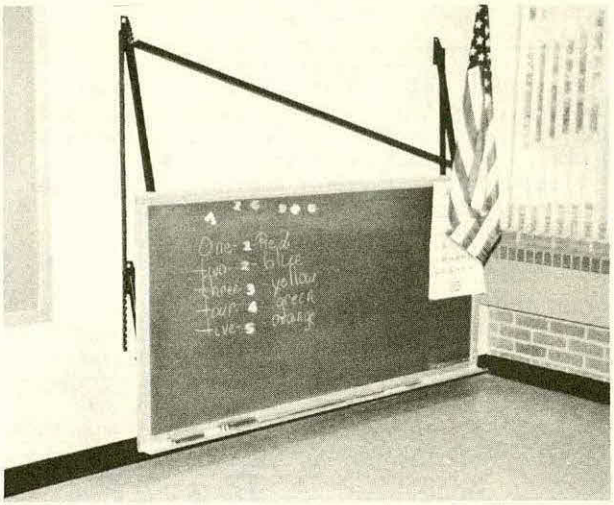

FIGURE 2: CHALKBOARD IN LOWER, STOWED POSITION

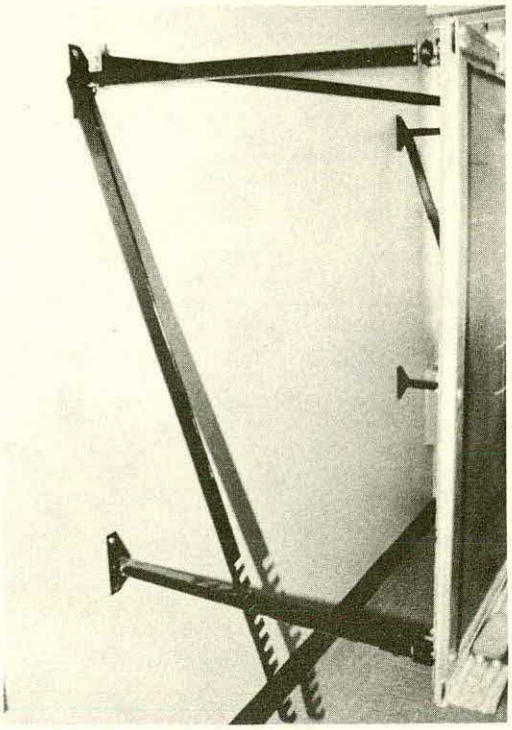

FIGURE 3: CHALKBOARD MOUNTING BRACKETRY

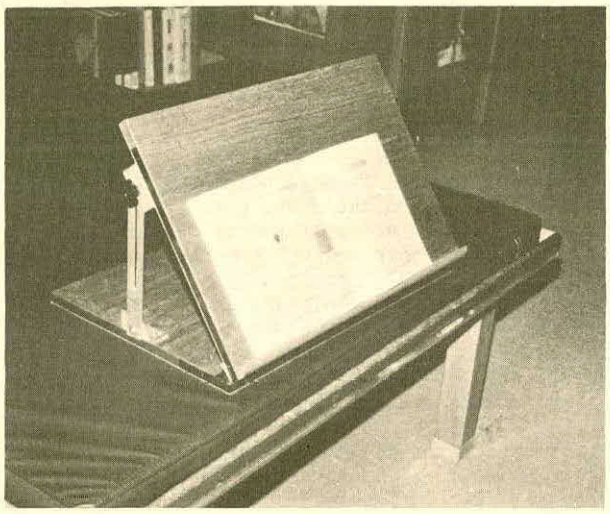

FIGURE 4: BOOKSTAND - RAISED POSITION

in the classroom requested that the chalkboard be capable of storing flush with the wall for the sake of room. It was then suggested by R.E.C. staff that if the board lowered to a position flush with the wall it could be accessible to students who would be on the floor. This was done, and it has proved to be a viable position for the chalkboard. Figure 2 shows the chalkboard in the lowered, stowed position. Figure 3 shows the mounting bracketry for the chalkboard.

Adjustable Angle Bookstand

Reading from a book that is lying flat is sometimes difficult for a person in a wheelchair. The

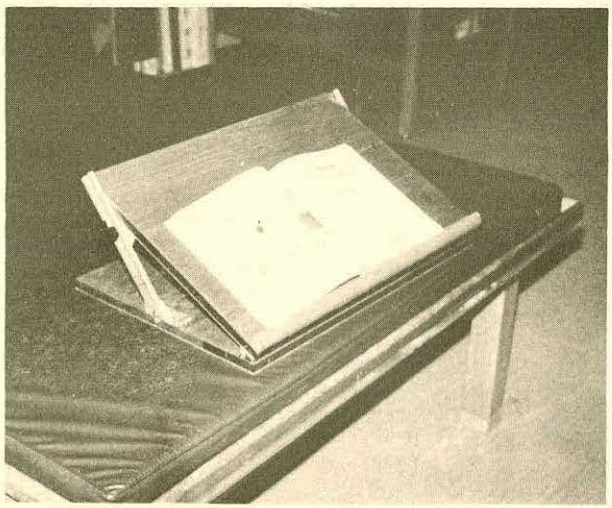

FIGURE 5: BOOKSTAND - LOWER POSITION

bookstand shown in Figures 4 and 5 is one such device devised to overcome this problem. The particular design shown is superior to those that were found in a search for off-the-shelf bookstands. It is heavy, stays in place for headwand and mouthstick users, and is easy to adjust by staff who might have reduced strength in their hands.

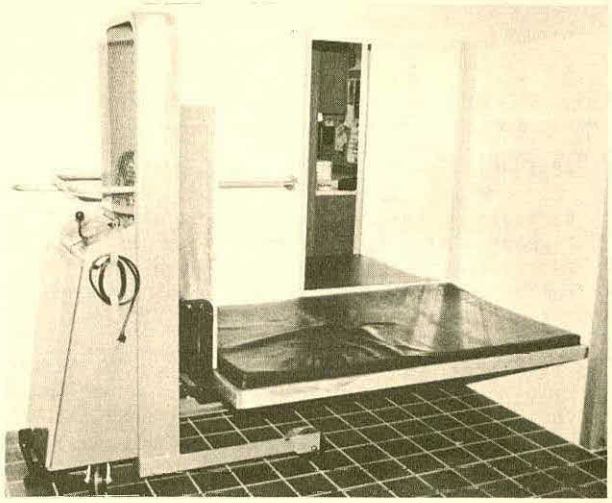

FIGURE 6: ADJUSTABLE HEIGHT CHANGING TABLE

Adjustable Height Changing Table

Members of the R.E.C. learned that some of the most difficult tasks that confront the workers in the physically impaired classroom are helping various students make clothes changes, removal and replacement of body jackets, and braces. Because students are sometimes heavy and of large stature they can be difficult to handle. A hydraulically powered lifting table was suggested. Figure 6 shows this device. A commercially available battery-powered shop lifter was purchased and modified to provide for a changing surface and for safety in use. The changing surface may be lowered to within five inches of the floor and raised to chest high on an average sized adult. Students can be thus transferred from the floor, or from any height wheelchair.

DISCUSSION

School staff members have found these subject devices to be of great benefit. The adjustable chalkboard is being investigated for commercial value. More information regarding the design of these devices can be obtained from the author(s).

ACKNOWLEDGEMENTS

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FUNCTIONAL ARM RESTRAINT FOR CHILDREN WITH ATHETOID CEREBRAL PALSY

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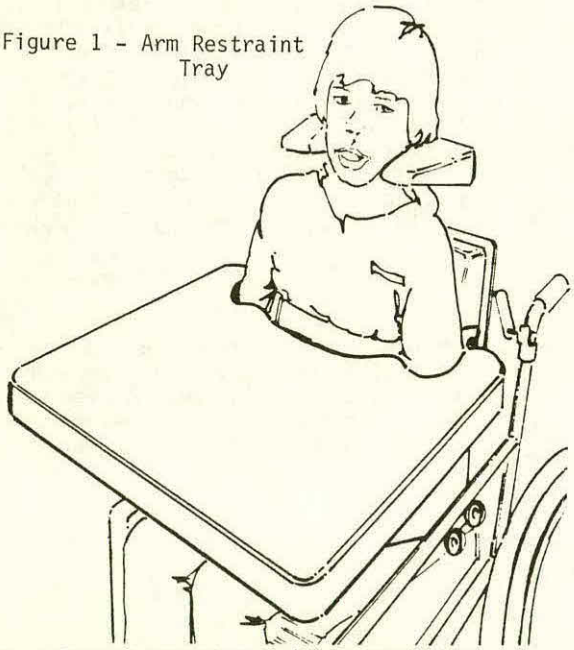
ABSTRACT

Restraining the arms of children with Athetoid Cerebral Palsy has proven to be an effective means of increasing function, stability, and comfort. Fourteen arm restraint trays have been fit to children as part of their technical service provision at the University of Tennessee, Memphis - Rehabilitation Engineering Program. This paper presents attitudinal and functional issues resulting from these fittings.

INTRODUCTION

The concept of restraining the arms of Athetoid children was proposed in 1982 (1). In all, 15 arm restraint systems have been fit. Of the total 15 systems, 14 were arm restraint trays and 1 was a rigid ABS restraint where no tray was needed. This report will present a review of the effectiveness of the arm restraint tray for a specific population group (Figure 1).

Figure 1 - Arm Restraint Tray



The original hypotheses (1) stated that restraining the arms of children with the diagnosis of Athetoid Cerebral Palsy would provide the children with increased stability in the upper trunk and head which in turn would increase the functional potential of the severely physically disabled child. Increased function results from both the increased stability and the ability of the restraint to prevent the arms from dislodging headsticks and other such devices.

The other feature of the original presentation which deserves note was the ability of the arm restraint feature to be applied for specific activities without being necessary at all times. This is accomplished by mounting the tray on adjustable height wheelchair arms.

The young adult who received an arm restraint system which did not include a tray was not included in the study. However a very brief description of his system is included at this time for the sake of clinical completeness. Jeff is a 25 year old man with a diagnosis of tension athetosis. A rigid arm restraint band of ABS was firmly attached to the back of his seating system to hold his upper trunk stable, restrain his arms above the elbow, and to provide a solid stationary attachment for the joystick control of his powered wheelchair system which he accessed with his chin. Jeff has been using his system successfully for two years.

METHOD

A review of fourteen arm restraint tray systems was completed by both reviewing the clients chart and through telephone interviews with the families and when appropriate with the client's teachers. A questionnaire was administered by an occupational therapist on the rehabilitation enginering staff. Except in one case the interviewing therapist had no previous contact with the families questioned. The one exception was the questioning of a parent whose child had expired recently and it was felt that a familiar person would provide the emotional link necessary to make the interview non-stressful.

DISCUSSION

Composite Client Profile

Diagnosis: Cerebral Palsy - Athetoid (3)
- Hypotonia (1)

Age Range: 11-18 Average Present Age: 15.3

Average age at which clients received their first arm restraint tray: 11.9

Mobility: Dependent (10)
Independent Powered Wheelchairs (4)

Communication: Augmentative Communication Device Users - Electronic Devices (5)
- Etran Users (5)
- Other Non Electronic Devices (Picture Boards) 4

Composite Technical History

Arm restraint trays of the standard design - 11

Arm restraint trays using standard trays modified to become arm restraint devices - 3

Arm restraint trays with Etrans as part of the design - 4

Arm restraint of only one arm - 2

Of the fourteen trays fit, two are not longer being used. One was for the hypotonic young man. His spinal curvature became much worse and

his seating system needed to be reclined to a degree that attachment of the arm restraint tray was no longer possible. The other young girl is still using the tray but her arms are restrained on top of the tray with small velcro straps which are sufficiently strong to hold her arms stable. However since these two clients have only recently changed trays, their interview information was still done pertaining to attitudes and performance of the trays during arm restraint use.

INTERVIEW SUMMARY

All 14 families were pleased with the arm restraint trays. Parents and teachers all felt that the children were more functional and comfortable with the tray in place. All felt that the tray added to the child's stability while sitting and in particular while doing functional activities. Three felt the trays were a little difficult to put on but all found them easy to get off. Only one family was worried about removing the tray quickly in an emergency but the two families who had experience in emergency removals had encountered no difficulties. All felt that the trays were attached firmly enough and none felt that repairs were a problem.

Eleven of the children liked the trays but three did not. Two felt too restricted and one could not use her arms to misbehave by knocking schoolwork off her tray. Even though the three did not like the tray, they continue to use it at school as they are more functional.

All families and teachers felt that the tray enhanced the functional abilities of the childs activities which they perform with the tray on including:

* access a communication board either with eye coding, light sensors or light beams;
* feed more easily;
* use head control to access computers with light sensors or single switches.

There were several activities that could not be done with the trays on as efficiently such as operation of an Autocom communication device which required its own mounting, finger painting, and as previously stated - knocking school work off the tray in order to control the environment. With the tray in place, children tended to use augmentative communication devices, including a Minspeak, Etrans, and word boards. Without the trays all reverted to either yes-no responses, eye pointing or vocalizations. None of the families felt that communication was easier with the tray off. Four felt there was no difference and the other ten felt that it was easier. It should be noted that there was no effort to monitor the quality of the communication behaviors.

The final functional activity which was addressed was feeding. All children but one were fed with the tray on at school. The 14th child was fed in a reclined feeding chair. At home the response was quite different. Only one child was fed with the tray on, four were fed out of their seating systems and the rest were fed in their chairs but without their trays.

When asked what they liked most about the tray, all stated the stability, the increased function and safety for the arms. They disliked the weight of the tray and lack of a lip around the tray edge. One parent did not like the method of attachment to the wheelchair and two requested a moisture proof clear overlay for communication boards. None of the families were familiar with the current cost of the arm restraint tray. When asked what they would pay to replace the tray, twelve stated that they would pay whatever it cost, two would pay $200 - $300 and one would pay over $300. However, all but the two no longer using the trays stated that since the trays were essential for school they would have to find the money for replacement whatever the cost.

COMMENTS

It was interesting to note that all families felt that the trays were essential to the childs participation at school even though many did not use the trays at home. The only professional who voiced a dislike of the tray was a physical therapist who was afraid that the tray could not be removed quickly in an emergency. The teacher and family of the same child had no such worries.

It was also interesting to note the number of children who were fed without the trays at home. It could be hypothesized that habit is very hard to break and that families had found a feeding system that worked for them without the tray use and were reluctant to change.

CONCLUSIONS

Arm restraint trays work well with severely physically disabled children with the diagnosis of Athetoid Cerebral Palsy. While using the tray, functional activities were enhanced by the increased stability and comfort the children experienced. Two of the children had only one arm restrained which allowed the dominant arm to be used to control powered wheelchairs, four of the children would not sit in their seating systems without the tray on even for short periods of time, attesting to their feelings of security and comfort with the tray in place. In general, the children liked the trays and even those who did not agreed that they were able to be more functional with the tray on.

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Toward Telephonic Transmission of Sign Language for the Deaf

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ABSTRACT-The work that is presented in this paper deals with data compression techniques that allow for the real-time transmission of messages in American Sign Language (ASL) and fingerspelling using the relatively low bandwidth residential telephone network. The origins of this project are found in the observation that over 500,000 deaf individuals who rely upon manual forms of communication are severely disadvantaged in using the telephone.

INTRODUCTION:

The transmission of visual images rather than auditory signals using the standard telephone system is made difficult because of the bandwidth requirement of a video quality image. The bandwidth of broadcast television is on the order of 1000 times that of the standard voice telephone network.

The results of the investigation have shown that both American Sign Language and fingerspelling can be transmitted intelligibly at digital data rates that are within the range of a commonly available 2400 baud telecommunications mode. The techiques that are employed involve encoding a single video frame (of which there are 30 per second) into a frame consisting of a relatively small number (51) pairs of two dimensional coordinates that correspond to key anatomical sites on the signer's body. These frames are transmitted at a rate of 6 per second. The coordinates of these sites on the body are used to drive a software animator that provides connectivity between the points, and interpolates the missing frames to provide for the flicker free 30 frames per second display. The resulting effect is of an animated stick figure that mimics the movements and, hence, the manual communication of the human signer.

VIDEO BANDWIDTH VS TELEPHONE BANDWIDTH:

The transmission of pictorial information using the existing telephone system is limited by the narrow bandwidth of the switched telephone network. The voice-only nature of the telephone has led to the definition of a usable bandwidth of 3.3 KHz. This is sufficient to allow speech communication at approximately 90% intelligibility. This is referred to as the narrow band system and typifies the majority of residential and business telephone lines.

The narrow bandwidth of the telephone must be contrasted with the bandwidth of broadcast television in order to appreciate the difficulties associated with pictorial transmission. Broadcast video requires a bandwidth of 4.2 MHz or more than a factor of 1000 higher.

The digital rates of communication for these two signals are somewhat different. For the purposes of this discussion it is sufficient to point out that narrow bandwidth telephone digital rates are limited to between 4800 and 9600 bits/sec. and broadcast video requires approximately 40 million bits/sec.

DATA COMPRESSION TECHNIQUES:

The transmission of moving visual images at low bit rates requires considerable processing of the original signal. The bandwidth required for the Bell Picturephone (Crater 1971) and other systems that transmit video quality images is considerably higher than that of the existing switched telephone system. The bandwidth reduction techniques for sign language proposed by Pearson (1981) and Sperling (1981) have difficulty maintaining acceptable intelligibilty at transmission rates that are nearly an order of magnitude higher than the telephone bandwidth. Sperling (1981) indicates that the minimum acceptable bandwidth is in the range of 21 KHz. Since analog bandwidth does not appear to be able to be reduced beyond that point, other approaches must be taken.

Johansson (1973, 1976) has shown that human movement can be characterized by a relatively small amount of consistent trajectory information. As long as the movements of these points is consistent with the expected movements of the human body, the viewer will perceive the moving spots as representing human motion.

Poizner et al. (1981) has extended the point light display to consider the movement of the upper extremities in ASL. They have shown that the point light movements of the shoulder, elbow, and wrist are sufficient to represent ASL signs that do not include hand configuration as a primary feature. Tartter and Knowlton (1981) have shown that point light displays can be used to present hand configurations as well. They use 13 spots on each hand and one on the nose. Signers have been able to conduct meaningful, fluent conversations in ASL.

These point light displays would require the complete bandwidth of an analog signal despite their relative lack of detail. A digital system could represent the 27 spots as 27 pairs of coordinates in a digital image.

REPRESENTATION OF LIGHT SPOT DISPLAYS:

The digital representation of a light spot display requires a pair of coordinates to define the unique location of each point in the frame. The resolution of the entire image need not be the 320 x 480 digital equivalent of an analog video signal. Sosnowski and Hsing (1983) point out that the television events interpreted for the Deaf devote only a portion of the video display to the interpretation. The image of the interpreter usually appears in a lower corner and occupies an area that is approximately 100 x 100 pixels. This has been experimentally shown to provide nearly 100% intelligibility by Pearson (1981), and Sperling (1981).

In a digital display this resolution can be described by 7 bits per axis (actually a 127 x 127

display). The coordinates of Tartter and Knowlton's 19 visible points (only 19 out of 27 can be seen at one time) then require 19 pairs of 7 bit numbers or 7,980 bits/sec. This is much closer to the bit rate allowable on the telephone than any other proposed strategy.

A POINT LIGHT DISPLAY TECHNIQUE:

The point light technique of Tartter and Knowlton omits several important aspects of manual communication. There is no provision for expression with any part of the body other han the hands. Shoulder shrugs, head movement, facial expressions, and lipreading are not available. Of equally important concern is the lack of fingerspelling. The data reduction technique presented in this project, is an extension of the point light concept that will provide the missing features of ASL that have been identified above and will also provide for intelligible fingerspelling.

The proposed technique will be considered to have 51 unique points: arms 3 points (shoulder, elbow, wrist), hands 19 points (all finger joints and tips), face 7 points (eyes, nose, 4 locations on mouth). It is believed that these points track the trajectories of the major components of the body during the production of a sign or fingerspelled word.

CONNECTIVITY PATTERN:

Demonstration of this technique that has been conducted as part of this project uses a connectivity pattern that connects appropriate points in each frame with straight lines. The frames are then displayed on a video monitor at 30 frames/sec. to create a simple moving stick figure.

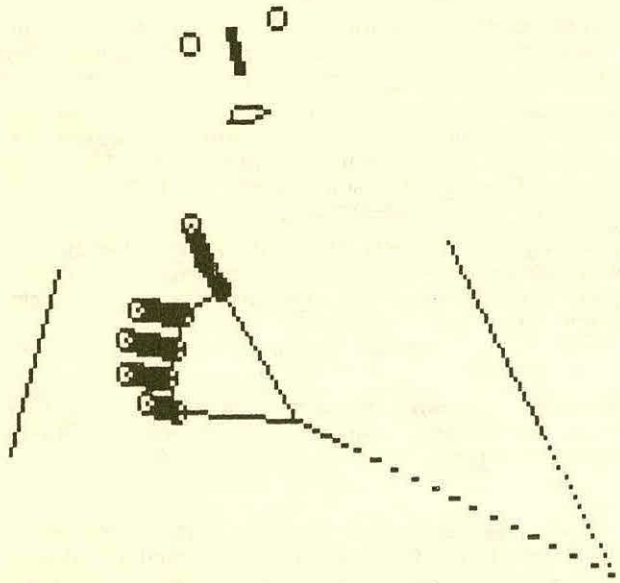

Figure 1. Sample Video Image of an ASL Sign

The stick figure has been enhanced only by making the fingers thicker than the arms so that they are more distinguishable, placing small circles at the eye positions, using a small vertical bar for the nose, and connecting the four points on the mouth to form a continuous opening. None of these affects the definition of the 51 points.

TOWARD TELEPHONE BANDWIDTH:

While 51 points and a connectivity pattern may allow the creation of animated stick figures that present intelligible signs and fingerspelled words, the bit rate required to transmit those points remains in excess of the telephone line capability. The 51 points defined with 7 bits in a 128 x 128 resolution display require 714 bits/frame. At 30 frames/sec., this demands 21,420 bits/sec.

ADJUSTMENT OF THE TEMPORAL RESOLUTION:

Thus far the the data reduction has focused on the spatial resolution required to define the image of the signer. Several attempts have been made to make similar adjustments to the temporal resolution. Pearson (1981) reports of an attempt to reduce the number of frames/sec. that are needed to display intelligible motion. The technique that was employed was frame repeating. The results were unsatisfactory with this technique introducing an unnatural and disturbing effect. Pearson concluded that: "Certainly there seemed to be no phenomenon in the temporal domain akin to Sperling's finding in the spatial domain: in contrast, a temporal resolution approaching that of broadcast television was needed."
Sosnowski and Hsing (1983) were able to achieve some reduction in the temporal domain. They have shown that there is only a modest loss of intelligibility when the frame rate is dropped to 15 frames/sec. Below that, however, the intelligibility drops off dramatically.

FRAME INTERPOLATION:

Since the disturbing effects are caused by changing frames at rates that introduce flicker then it should be possible to eliminate the flicker by lowering the frame rate for transmission of the data by telephone and interpolating between those frames to replace the missing frames with synthesized frames and allow the frame rate to be returned to 30 frames/sec.

The extent to which the frame interpolation will be useful depends upon the spectral content of the trajectory information in the original signal. Since the 51 points are represented in the 30 frames/sec. of the video signal, sampling theory indicates that the highest frequency present in the sampled signal cannot be higher than the Nyquist frequency or 50% of the sampling frequency. This limits the spectral content of the trajectories defined by the 51 points to a maximum of 15 Hz.

The study of the frequency content of a small number of ASL signs reported by Poizner et al. (1983) shows similar results with the 0 to 2 Hz range containing most of the activity.

PREDICTED MINIMUM SAMPLING RATE FOR ASL:

If 3 Hz is the highest frequency component in the signal, the sampling frequency is determined to be minimally 6 Hz. This sampling frequency represents a 5 to 1 reduction in the amount of information that must be transmitted per second. The predicted data rate for the 51 point transmission is reduced from 23,820 bits/sec. to 4284 bits/sec. This is below the 4800 bits/sec. data rate allowed for one way transmission on an unconditioned residential quality telephone. It would be desirable to bring the bit rate below 2400 bits/sec. This is done in a reasonably

nondestructive manner by using frame to frame differences of points rather than the absolute positional values of the points. The final bit rate that is achieved is 2388 bits/sec.

INTERPOLATION TECHNIQUES:

There are several interpolation techniques that could be employed. The method chosen for this project is a Bezier curve interpolation (Artwick 1984). This places a smooth third order curve through the points in the real frames and selects four equally space points on that curve as the interpolated points.

INTELLIGIBILITY STUDIES:

The intelligibility of the stick figure representations was evaluated in experiments using both ASL and fingerspelled stimuli. Nine Deaf graduate students and staff members of the American Sign Language Program at Northeastern University in Boston served as subjects.

Test stimuli were presented on video tape and included representations of a human signer, an animated stick figure with 30 frames per second of original data, and a Bezier interpolated stick figure using 6 frames per second of original data. The animation was done on an IBM AT microcomputer using custom software. The ASL signs chosen for use in the experiment were selected randomly from an elementary ASL text. The fingerspelled words were all proper male and female names. All stimuli were based on a set of tapes prepared for the project by the Communication Disorders Department of Emerson College in Boston.

Subjects were shown the stimuli according to an experimental design that minimized the effect of order and were asked to write their interpretation of the signed or spelled message.

EXPERIMENTAL RESULTS:

The results of both the ASL and fingerspelling experiments were very successful. Analysis of the subjects' scores showed that differences among the three modes of sign presentation (in the case of both the ASL and fingerspelled stimuli) were not significant at the .05 or .10 levels.

This indicates that the interpolated and non-interpolated stick figure signs were as intelligible as the signs produced by the original signer.

CONCLUSION:

The results of this project have shown that it is possible to consider transmitting ASL and fingerspelling over the narrow bandwidth residential telephone network. The reduction of every fifth frame to 51 feature points combined with additional intra and inter frame differential coding provide sufficient compression to allow a data transmission rate of less than 2400 baud. This information provides sufficient trajectory information so that an animation program can interpolate missing frames and display a stick figure that can be accurately interpreted as the original message.

The success of this project has led to the continuation of the effort to pursue the methods whereby the 51 points can be identified in real-time. This will allow the realization of a sign language telephone.

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Assistive Devices for Deaf and Hard of Hearing People

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ABSTRACT

The Gallaudet Assistive Devices program for hearing impaired people is discussed in the following paper. Devices for large area communication, interpersonal communication, television and radio, telephone, and alerting purposes are described. In addition, there is discussion of equipment displays, demonstration programs, fitting of devices to consumers, professional education programming, and ways in which program staff interface with distributors and manufacturers.

INTRODUCTION

In the United States alone, approximately 14 million people suffer from hearing impairment (1). According to the 1982 Bureau of Census report (2), 49% of those people are between the ages of 22 and 64, major years of productivity. An additional 43% are 65 and over. The number of hearing impaired people can be expected to increase because the elderly population is growing and because the amount and severity of noise exposure is increasing.

The vast majority of hearing impaired children and adults have sensorineural hearing losses which are not medically or surgically treatable. The issue then becomes one of finding mon-medical ways of alleviating the communication problems. Hearing aids have serious limitations because they do not function well in noisy, reverberant settings or in situations requiring distance listening. They amplify all sounds within their range equally well, requiring the listener to sort out speech from noise.

People with sensorineural hearing loss experience more difficulty sorting speech from noise than the normal hearing individual because of the "built-in" distortion that is part of the hearing problem. Therefore, while hearing aids might be useful in quiet situations involving one to one communication, they are often useless for group communication, meetings, lectures, television viewing, or telephone communication.

Assistive listening devices can be very useful in such situations because they increase the level of the speech compared to the level of the background noise. This equipment supplements hearing aids and in many situations provides help that hearing aids cannot provide.

BACKGROUND

Most assistive devices require the use of a microphone located close to the sound source. Once the speech enters the microphone, it is processed in such a way that it reaches the listener's ear at the same intensity level. In contrast, environmental noise must travel through the air to reach the microphone, and loses energy in the process. With a hearing aid, the microphone is worn on the listener's head, often resulting in both signals reaching the listener at the same level. Not only is the speech distorted by the noise but it also has lost energy because the listener is usually at some distance from the sound source.

Types of Assistive Devices

Assistive listening equipment includes:

1. Systems for large area communication such as induction loops, FM, and infra-red devices. Such systems are suitable for meeting rooms, auditoriums, theatres, churches, and classrooms.

2. Devices to improve interpersonal communication in a one-to-one situation or in group conversation. A number of systems fall into this category, including adaptations of those used for large area communication. An example of such a system is an FM microphone-transmitter worn by a speaker and an FM receiver worn by a listener.

3. Television and radio devices. There are two major problems associated with television and radio listening. First, the hearing impaired person frequently needs to use more volume than is comfortable for others in the environment. A second issue involves the ability to hear the television in the presence of competing noise or reverberation. The microphone of an assistive listening device picks up the signal as soon as it leaves the loudspeaker and transmits it to the listener's ear without attenuation.

Since competing noise is being attenuated because it must travel through the air to reach the listener's ear, the signal to noise ratio is improved. The problem of too much volume for normal hearing listeners is solved because the hearing impaired listener adjusts his or her volume on the device rather than the television. FM, infrared, loop, and hard wire technologies have been adapted for radio and television use.

4. Telephone communication presents special problems because the hearing impaired consumer cannot speechread or use gestural information normally available in face to face communication. To alleviate these difficulties, some individuals use hearing aids with built-in telecoils which pick up magnetic leakage from the telephone receiver. Hearing aids may be used with the telephone acoustically using the hearing aid microphone, but room noise and acoustic feedback may cause problems. Some hearing impaired people use amplified telephone handsets or portable amplifiers attached to the telephone with or without hearing aids.

5. Alerting device. There is a significant population of deaf individuals for whom listening devices are not sufficient. Fortunately, visual and vibratory systems are available to meet some of their communication needs. There are many tele-communications devices for the deaf (TDDs) on the market to facilitate telephone communication; a closed captioned decoder is available to make captioned television programs accessible; many alerting devices using flashing lights or vibrators are available to monitor important sounds in the home.

Most assistive listening devices can input to the listener's ear in one of three ways: 1. through their own earphones, 2. directly hard wired to the amplifier of the personal hearing aid, 3. through the hearing aid by means of magnetic induction coupling using the hearing aid telecoil. The input mode is selected to meet the individual needs of the client.

THE PROBLEM

Although many assistive listening devices are available on the market, there are very few quality control requirements in place. There are very few standards for equipment performance, methods for evaluating devices, or comparative data. Very few audiologists recommend or dispense assistive devices primarily because of the confusion in the state of the art. At the same time, hearing impaired consumers and family members try to obtain communication devices any way they can. Often this desire results in purchase of inappropriate equipment. Hearing impaired consumer groups have tried to fill the void by establishing their own assistive devices centers. Consumers, however, lack the technical and audiological background necessary to evaluate such equipment.

ONE SOLUTION

To try to meet the need for a professionally managed assistive devices program, the Department of Audiology at Gallaudet College has established a model service delivery/demonstration program. The heart of this program is the Assistive Devices Center which is divided into display areas, each one demonstrating one category of device. A large number of television, telephone, alerting, large area, and interpersonal communication systems are available for demonstration to consumers as well as individual fittings to meet specific communication needs.

The basic objective of the Gallaudet assistive devices program is to provide professional assistance and information about communication devices to hearing impaired consumers and their significant others. In order to accomplish this goal, an aggressive outreach program is being established to educate consumers and professionals about the availability of assistive devices and methods of obtaining them. Equally important is good liason with distributors and manufacturers to develop smoothly functioning distribution systems and channels of communication concerning suitablility of equipment and desirable modifications. Following is a discussion of ways in which both needs are being met.

Consumer Services

Demonstrations. In order to familiarize hearing impaired consumers and their significant others with the various types of equipment on the market, their advantages and limitations, and how to obtain quality service, the Gallaudet Assistive Devices Program offers demonstrations for any interested group either in the Center or in an outreach capacity. During group or individual demonstrations, clients receive the following information:

1. What are assistive devices?
2. Who uses them?
3. When are they used?
4. Why are they needed? Why are

hearing aids not sufficient?

5. Where can they be obtained.?

Individual Fitting. Typically, a fitting procedure consists of several visits involving basic hearing assessment, admini-stration of a communication needs questionnaire, hearing aid evaluation, assessment and fitting of assistive devices, a trial period of some sort, and follow-up visits. Assistive listening devices vary in their power and frequency response characteristics; some are capable of a variety of internal adjustments. Therefore, the audiologist must have complete information about the degree and type of hearing loss the client presents prior to fitting of devices. Most assistive listening devices are used with wearable hearing aids. Therefore, the status of the client's personal amplification is an important issue; it must be evaluated and if necessary remediated before an assistive device can be recommended. Recommendation of a visual or vibratory device is simpler because the status of the hearing and the hearing aid is usually not a crucial prerequisite.

After administration of the assistive devices questionnaire, basic hearing and hearing aid evaluation, the client and clinician evaluate equipment that might meet communication needs, and make a decision. The client receives practical information concerning portability, battery life, price, versatility, etc. He or she is then referred to an approved vendor who will offer a short trial prior to or following purchase of the device.

Professional Education

Nurses, physicians, social workers, educators, administrators, speech-language pathologists, and psychologists are potential resource people for hearing impaired clients. They need the same information as consumers plus knowledge of referral sources in their communities so that they can consistently refer clients to appropriate facilities. In addition, these professionals need to be comfortable with the use and care of assistive device equipment in order to assist consumers.

Audiologists, as the only professionals trained to fit assistive listening devices, require information about fitting procedures and administrative concerns involved in managing assistive devices programs. Finally, trainers of professionals need to learn the most effective procedures to educate front line people who work directly with deaf and hearing impaired consumers.

The Gallaudet assistive devices program is developing training programs for professionals and trainers of professionals. The training package for resource people includes the same information provided to consumers plus sections on use and care of assistive devices and referral procedures. All of these topics are included in the training program for the audiologists plus specific information on fitting auditory devices. The program for trainers of professionals focuses on teaching techniques in addition to the areas included in the other training packages. By training professionals, more consumers can be reached more efficiently.

Interface with distributors and manufacturers

The equipment in the Assistive Devices Center has been obtained: 1. directly from manufacturers through their representatives, 2. from distributors who also supply local vendors, and, 3. from local vendors themselves. Local vendors supply primarily alerting devices and TDDs. Most of the devices have been obtained on a consignment basis with the understanding that the supplier updates equipment and keeps the audiologists informed of new technology. The audiologists, in turn, share with distributors information on which devices seem to be most useful to consumers and what technological changes are desirable.

Because the Gallaudet program does not dispense equipment, it has been necessary to develop a network of local dispensers to sell recommended devices. Criteria for selection of dispensers includes interest in dispensing devices, knowledge of equipment, and willingness to cooperate with audiologists from the Assistive Devices Center particularly in the solution of special problems. Other criteria of importance include ability to establish good rapport with clients, willingness to spend the time necessary for fitting, instruction, and follow-up, and fair prices. It is also considered essential to select dispensers also fit hearing aids because hearing aids and assistive devices are so frequently interfaced. The audiologist and dispenser work as a team, conferring as often as necessary.

Many of the assistive listening devices fittings involve considerable creativity on the part of the audiologist because the choice of an optimal system often needs to be highly individualized. An individual frequently presents several communication problems; it is

considered desirable to develop a single versatile system to resolve as many of these problems as aa possible. This means that an assistive device system may require the use of several microphones, different switching options, or different ways of inputing the assistive device to the listener's ear. Several examples follow:

Mr. Jones is a severely hearing impaired college student rooming with another hearing impaired person and one with normal hearing. He needed a system to allow him to listen to television so that his hearing roommate would not be disturbed while studying. In addition, he needed a systems that would tune out the very loud stereo music played by the other hearing impaired roommate. The problem was solved by the use of a system that directly connected Mr. Jones' hearing aid to the earphone jack of the TV, turning the TV speaker completely off and allowing Mr. Jones to listen through his hearing aid without interference of music from the stereo. There were times, however, when all three roommates wished to watch television together. In order to make that possible, a microphone had to be substituted for the direct jack connection so that the TV loudspeaker would continue to function. Mr. Jones found that he could also use the remote microphone for one-to-one interpersonal communication by clipping it to his clothing.

A second example involving the courtroom problems of a trial attorney, was described in a publication by Fernandes (3). In order to successfully function in the courtroom, Mr. R needed to clearly hear three talkers: the judge, the witness and the prosecutor. The problem was solved with a two-channel FM receiver equipped with a switch for changing from channel to channel. One transmitter functioning on one channel, was used by the judge who wore a microphone attached to his or her clothing. An additional microphone on an extension cord was used by the witness. The second transmitter, functioning on the second channel, was worn by the prosecuting attorney. Mr. R. switched channels on his FM receiver as each person spoke.

In order to develop such individualized systems, close interaction between audiologist, manufacturer, dispenser and client are necessary. Not only does the client benefit, but the state of the art is advanced because new systems are developed.

CONCLUSIONS

Assistive devices are truly a new frontier. They are in their infancy, with unlimited potential for development of new systems. Close cooperation between manufacturers, audiologists and dispensers can result in rapid advancement in the state of the art. Education of consumers and professionals can increase opportunities for communication rehabilitation. These are the goals of the Gallaudet assistive devices program.

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COMPRESSION-LIMITED AMPLIFICATION FOR THE PROFOUNDLY DEAF

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ABSTRACT

A master hearing aid with adjustable compression-limiting, and adjustable pre- and post-limiting frequency response was developed to provide improved audibility of the speech spectrum for subjects with profound sensorineural hearing loss. Evaluation involved the training and testing of 10 subjects using a speech pattern contrast perception task. Subjects were trained under an "equalization-only" condition and an "equalization-plus-compression" condition. Five of the subjects showed benefits as a result of training but in none was there improvement with the introduction of compression. Several subjects showed an improvement of dynamic range with training, - reducing their need for compression. A subsidiary experiment with simulated deafness confirmed that compression adds little to a reduction of effective dynamic range once appropriate equalization of frequency response has been accomplished.

PURPOSE

The purpose of this study was to investigate the potential role of compression-limited amplification, combined with pre- and post-limiting frequency-response adjustment, in the management of children with profound, sensorineural hearing loss (i.e., hearing losses in excess of 90 dB arising from damage to sensory receptors or sensory nerve fibers). Such children have a dynamic range, (i.e., the range from threshold of audibility to threshold of discomfort), that is both narrow and frequency-dependent (1) . If these children have sufficient frequency and time resolution, a combination of compression-limiting and spectral equalization should improve speech perception by rendering the acoustic speech spectrum more completely audible than would be the case with conventional amplification (2) . It is possible, however, that some profoundly deaf individuals are only able to make use of the information contained in amplitude-envelope variations (3), in which case compression-limiting would tend to reduce speech perception performance. Earlier work with compression amplification has produced mixed results (4,5,6,). Much of this work, however, has employed subjects with hearing losses of less than 90 dB. Such subjects typically have dynamic ranges of 30 dB or more, and with a constant average speech input are less likely to need amplitude compression than are the profoundly deaf.

EQUIPMENT

A Master Hearing Aid with adjustable compression limiting was built for this study. The essential components are shown in the block diagram of Figure 1. The preamplifier incorporates a compressor-AGC circuit with a fast attack time (10 ms) and a slow release time (adjustable from 0.5 to 2 sec). As long as the peaks of the input signal are above the knee of this first compressor, the following circuits receive a signal with a constant peak level and an unmodified dynamic range. A "speech" filter is then used to "flatten" the speech spectrum by compensating for the high frequency roll-off found in the normal speech signal. A set of pre-limiting filters permits further adjustment of the relative strengths of the high and low frequencies, and these are followed by a pre-limiting amplifier with adjustable gain. When the gain is set to unity, the peaks of the speech signal fall just below the knee of the second compressor and no limiting takes place. As the gain is increased, the compression-limiter is activated, the reduction of dynamic range being equal to the increase of gain. The possible range of compression is from 0 to 25 dB. The compression-limiter has a fast attack time (10 ms) and a fast release time (adjustable from 10 to 100 ms). This compressor is capable of recovering within the space of a single speech sound and thus equalizing the peak intensities of consecutive sounds. Following the compression-limiter is a second set of filters which permit the compressed speech spectrum to be matched to the needs of the individual listener. The power amplifier is followed by an attenuator that permits the peaks of the output signal to be varied in 2 dB steps from 70 to 130 dBSPL.

EVALUATION

Subjects

Ten profoundly deaf children aged 12 to 14 years participated in an evaluative study. Their hearing losses (better-ear average thresholds at 500, 1000, and 2000 Hz) ranged from 90 to 110 dB. All had sensorineural losses, acquired before the development of language, and were enrolled in a school for the deaf.

Testing and Training

Speech perception was measured using a Speech Pattern Contrast (SPAC) test (3) . This test measures the ability of subjects to perceive 4 suprasegmental and 8 segmental contrasts in a binary choice-format. For the purposes of the present study, only 4 contrasts were used: vowel height (e.g., fool vs. fall), vowel place (e.g., fool vs. feel), final consonant voicing (e.g., seat vs. seed), and final consonant continuance (e.g., seet vs. cease). Stimuli were presented in the carrier phrase "you will mark .... please",

and subjects were given feedback on performance, with a chance to hear the stimulus again in the event the wrong choice was made. Stimulus presentation, response input, scoring, feedback, and replay were managed by a computer-interactive audio-tape system developed for this project.

Design

Subjects' audibility thresholds and highest comfortable listening levels for warble tones were measured through the Master Hearing Aid at octave intervals from 250 thru' 8000 Hz. These data were then used to adjust the pre- and post-limiting filters, and the compression-limiter, for maximum audibility of the speech spectrum. This condition we refer to as "equalization-plus-compression". By reducing the compression to zero, we produced a second condition which we refer to as "equalization-only". Subjects were trained for 8 to 10 sessions in the equalization-only condition, followed by 6 to 8 sessions in the equalization-plus-compression condition. They then returned to the equalization-only condition for between 2 and 8 sessions. (Individual schedules varied according to the demands of the educational program and the constraints of the school calendar). Each session lasted 20 minutes and subjects were seen for 2 or 3 sessions a week. Five of the 10 subjects worked on the vowel height and vowel place contrasts. The other five worked on the consonant voicing and continuance contrasts.

Results

Learning. Five of the 10 subjects showed clear evidence of learning through the experiment (as represented by a significant correlation between score and session number) for at least one of the contrasts on which they were trained. A sixth subject showed evidence of learning during the equalization-only condition but not during the equalization-plus-compression condition.

Effects of Compression. For nine of the ten subjects, the addition of compression raised neither the absolute performance, nor the learning rate above those observed with equalization alone. For the tenth subject, the addition of compression caused a drop in performance on one of the contrasts from around 70% correct to around 50% correct (i.e., chance level). The contrast in question was final consonant voicing, - a contrast that can be cued by durational differences alone.

Dynamic Range. All testing and training was done at the subject's highest comfortable listening level, this being determined anew at the beginning of each session. Since speech detection threshold was also determined, it was possible to track any changes in the dynamic range for speech. In 4 of the 10 subjects there was a significant increase of dynamic range during training, indicating a reduced need for compression limiting.

Discussion. These results do not provide support for the hypothesis that the addition of compression to frequency equalization improves speech perception in the profoundly deaf. In one case, in fact, the evidence was in the opposite direction, i.e., that the addition of compression reduced speech perception. The findings are therefore in keeping with earlier studies that have shown that a proper frequency response takes care of many of the difficulties encountered by hearing-impaired listeners and that the use of compression may add nothing significant to the information available.

The result should, however, be regarded as preliminary and therefore interpreted with caution. The profoundly deaf are a heterogeneous group and only 10 subjects were represented here. Moreover, only a restricted number of speech contrasts were evaluated in this study, and these were not necessarily the ones most likely to benefit from compression-limiting. It should further be noted that only the fast-acting compression limiting was removed during the equalization-only condition. The compression-limiting AGC circuit was still operating and served to maintain the speech peaks at a constant level, while the equalization circuits served to compensate for the high frequency roll-off in the speech spectrum and the frequency-dependent thresholds of the hearing-impaired subjects.

The case of the subject whose performance deteriorated under the equalization-plus-compression condition is interesting, especially since the contrast involved was final consonant voicing - a contrast that is perceptible on the basis of amplitude-envelope cues alone. It was hypothesized that the compression, by amplifying the sound immediately before and after the test word, could have removed word-boundary cues. To test this, she was retested using isolated vowel-consonant syllables in a three-interval, forced-choice task. In this situation she scored almost 100% in the equalization-plus-compression condition.

CONCLUSIONS

1. These results do not provide support for the use of fast-acting, compression-limiting as a means of solving the dynamic range problem of the profoundly deaf.

2. They should be regarded as premature, however, since neither subjects nor speech pattern contrasts were not fully represented in this study.

3. There was one subject in whom compression-limiting produced negative results. Additional data suggest that the compression made it difficult for this subject to identify word-boundaries.

FOLLOW UP

As a secondary test of the hypothesis that appropriate frequency-equalization may accomplish much of what can be accomplished by a single band compression system of the type used here, an experiment was performed with simulated deafness. Eight normally hearing subjects listened to speech in 70 dBSPL of white noise. Using monosyllabic words, scored phonemically, performance/intensity functions were plotted under 4

conditions: constant gain at all frequencies, frequency equalization by high frequency emphasis, 20 dB of compression limiting, and equalization and compression combined. As predicted, the slopes of the functions were very similar for the last 3 conditions, and all three were markedly greater than for the first condition.

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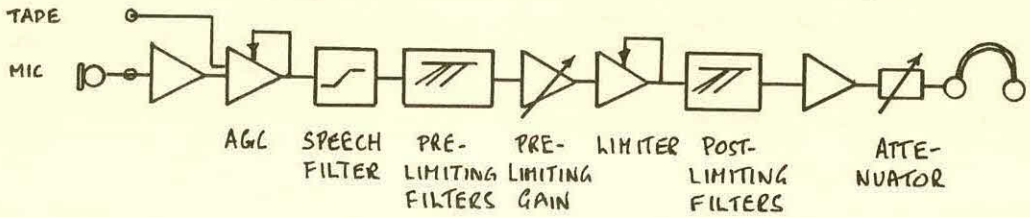

Figure 1. Block diagram of essential components of master hearing aid. Actual system has separate channels for right and left ears, adjustable release times for AGC and limiter, a set of 24 dB per octave low-pass filters to control acoustic feedback, and peak clipping as an alternative to compression-limiting.

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EVALUATION OF REDMASK NOISE REDUCTION FILTERING

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ABSTRACT

Performance-intensity functions for monosyllabic words presented against a background of broadband noise (0 dB and 15 dB S/N) were obtained for normal and hearing-impaired subjects through a filter designed to reduce the effect of background noise. The REDMASK filter places the noise spectrum parallel to the listener's threshold. The purpose of such filtering is to reduce the masking of noise while preserving intelligibility and quality. Normal hearing subjects exhibited higher maximum speech discrimination scores than did the hearing-impaired subjects for both noise conditions. The scores of the hearing-impaired subjects plateaued at lower sensation levels than did the scores of normal hearing subjects. For both groups, maximum scores in the 15 dB signal-to-noise ratio condition were equivalent to scores which could be expected in quiet.

INTRODUCTION

While many hearing-impaired listeners are satisfied with the hearing aids they are using, many others find their hearing aids to be of limited value. A common complaint is of difficulty in using a hearing aid in noisy situations. Two types of problems are identified by hearing aid users: difficulty in understanding speech-in-noise, and the annoyance with having background noise amplified to uncomfortable levels.

In the past, fixed high pass filtering has been used in hearing aids in an attempt to reduce background noise. More recently, several methods have been implemented for automatic suppression of low frequencies, which include level dependent control of cut-off frequency or slope of a high pass filter, or level dependent compression of low frequencies (1). All of these methods depend exclusively on the level of the background noise and suppress only low frequency energy.

An alternative method of filtering would be listener specific; i.e., information about the listener's hearing, as well as information about the noise characteristics would be used to calculate the filter which would keep noise parallel to the listener's threshold in the frequency region through 4000 Hz. This type of filtering called REDMASK could be expected to reduce the masking of background noise. The potential advantage of this method over high pass filtering is that low frequency speech energy is preserved. The presence of energy in all frequency regions should allow good speech intelligibility and result in more acceptable quality than high pass filtering.

The purpose of the present experiment was to determine the effect of this type of filtering on the intelligibility of a speech signal at two signal-to-noise ratios (0 dB and 15 dB).

METHODS

Ten normal hearing and ten hearing-impaired listeners served as subjects. The hearing-impaired listeners had moderate to moderately-severe sensorineural hearing losses and were experienced hearing aid users.

A digital master hearing aid system (2) was used to obtain the psychoacoustic information necessary for specifying the filter characteristics for a given listener and then for simulating those filters. The master hearing aid consists of a MAP 300 array processor which is controlled by a DEC LSI-11/23 minicomputer. The array processor is used to generate non-speech test stimuli, to calculate the filter for a subject based on certain measurements obtained on the subject, monitor subject response, analyze data, and check the output of the array processor.

For the first part of this experiment, the system was used to generate one-third octave bands of noise for the measurement of threshold and loudness discomfort level (LDL) at all one-third octave frequencies from 200 Hz through 4,000 Hz. A 12,000 Hz sampling rate was used. The output of the system was passed to a digital-to-analog converter. The analog signal was low pass filtered (5,000 Hz), amplified, attenuated, and sent to a Knowles CI 1955 hearing aid receiver mounted in a post-auricular hearing aid case. The hearing aid case was worn by the subject in combination with a custom earmold.

For the second portion of the experiment which involved speech testing, the output of a two-track tape recorder was mixed at the specified signal-to-noise ratio, low pass filtered (5,000 Hz), and sent to an analog-to-digital converter (12 bit). A 12,000 Hz sampling rate was used. The rest of the system is identical to that used in the first portion of the experiment.

An adaptive up/down strategy (3) was used to obtain threshold and LDL levels from the subjects. Instructions for LDL were those of Cox (4). The filter characteristic was calculated to mirror the listener's threshold for the one-third octave bands of noise and was simulated using the array processor. The listener's LDL data were used to limit the output of the system for subsequent testing.

During the second stage of testing, performance-intensity functions for speech-in-noise were obtained for the two groups of listeners under

two conditions, 0 dB signal-to-noise ratio (S/N) and 15 dB S/N. For the latter condition, the overall level of the noise signal was dropped by 15 dB on the input to the system. Recordings of twelve randomizations of the Northwestern University (NU) Auditory Test No. 6 (5) were used as the speech stimuli. The speech was mixed with broadband noise at the specified signal-to-noise ratio prior to digitization and filtering. For the 0 dB S/N condition, presentation levels were chosen to place the noise in each one-third octave band at -10, -5, 0, 10, 20, and 30 dB sensation level (SL) re. the listener's threshold for the one-third octave bands of noise. In the 15 dB S/N condition, while the overall level of the noise was dropped by 15 dB the speech levels were identical to those used in the 0 dB S/N condition. Presentation level, speech test protocol, and signal-to-noise ratio were randomized across subjects.

Subjects listened monaurally. Each subject wrote his responses for the speech discrimination task. All testing was done in a sound-treated room.

RESULTS AND DISCUSSION

Speech recognition scores in proportions were converted to arcsine units to stabilize the error variance. Four separate analyses were done to determine whether or not scores varied significantly as a function of presentation level. A repeated measures analysis of variance model was used.

For both normal hearing and hearing-impaired subjects, the factor of level was highly significant at both S/N ratios ($p < .001$). Mean scores for each level at each signal-to-noise ratio appear in Table 1. As can be seen from inspection of the table, for both groups of subjects mean performance increases as a function of presentation level. Mean performance of the normal hearing subjects is similar to that of the hearing-impaired subjects at levels through 0 dB SL, but performance of the normal hearing subjects is much better than that of the hearing-impaired subjects at positive sensation levels.

Table 1. Mean scores on the Northwestern University Auditory Test No. 6 (percent correct) for ten normal hearing and ten hearing-impaired subjects at six presentation levels (0 dB and 15 dB Signal-to-noise ratio) through the REDMASK filter.

	-10	-5	0	10	20	30 dB SL
Normal Hearing						
0 dB S/N	11.8	31.2	54.0	80.4	82.8	83.6
15 dB S/N	10.6	31.0	52.8	87.6	94.6	97.4
Hearing-Impaired						
0 dB S/N	13.6	33.0	47.8	53.6	53.4	50.8
15 dB S/N	15.8	37.8	54.8	67.0	67.8	69.6

The Scheffe procedure was used for post hoc testing. For the normal hearing subjects, the pattern of performance was similar at -10, -5, and 0 dB SL at both signal-to-noise ratios. The mean scores at these three lowest levels (-10, -5 and 0 dB SL) were significantly different from each other and from the higher levels ($p < .05$). As can be seen from Table 1, scores increase systematically until 10 dB SL. At 0 dB S/N, the mean scores at the three higher levels (10, 20, and 30 dB SL) did not differ significantly ($p > .05$). In contrast, in the 15 dB S/N condition the mean score at 20 dB SL was significantly higher than the score at 10 dB SL, and the mean score at 30 dB SL was significantly higher than that at 20 dB SL. Thus for the normal hearing listeners, performance plateaus at 10 dB SL in the 0 dB S/N condition, while performance continues to improve with increasing level in the 15 dB S/N condition.

For the hearing-impaired subjects, the pattern of performance was similar at the two lowest levels (-10 and -5 dB SL) in both S/N conditions. Mean scores at these two levels were not significantly different ($p > .05$), but were significantly lower than scores obtained at the higher presentation levels. For the 0 dB S/N condition, the mean score at 0 dB SL did not differ significantly from the scores at the higher sensation levels ($p > .05$). For the 15 dB S/N condition, the mean scores at 0 dB and 10 dB SL did not differ significantly from each other, but were significantly lower than scores at the higher levels ($p < .05$). Thus for the hearing-impaired listeners, the mean performance for the 0 dB S/N condition plateaus at 0 dB SL, while for the 15 dB S/N condition performance plateaus at 10 dB SL.

The performance-intensity functions obtained on the normal hearing subjects show that filtering in a manner which minimized the level of noise at all frequencies allows high levels of performance on the NU Auditory Test No. 6. Scores for the 15 dB S/N condition are similar to those which would be obtained in quiet. Since the performance of the hearing-impaired subjects plateaued at lower sensation levels than did the normal hearing subjects, it is of interest to compare this group's mean score in quiet through a headphone with a flat frequency response (TDH 49) to those obtained through the REDMASK filter. The mean score (quiet) of the ten hearing-impaired subjects on the NU Auditory Test No. 6 administered at 30 dB SL re. the pure tone average was 80%. This score is similar to the maximum score obtained in the present study in the 15 dB S/N condition. Thus the hearing-impaired subjects listening through the REDMASK filter were also able to obtain maximum scores for the 15 dB S/N condition, even at low sensation levels (10 dB SL).

CONCLUSION

The use of the REDMASK filter thus appears to be potentially useful in maintaining intelligibility, while still maintaining comfort. Such a filter would be simple to implement in a wearable hearing aid. The requisite psychoacoustic measurements

could be made at the time of prescribing a hearing aid and the appropriate filter could then be activated automatically in the presence of noise.

Further work is planned to compare performance with this type of filtering to that of selective amplification and selective amplification combined with high pass filtering. The effect of REDMASK filtering on quality will also be evaluated.

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COMPARISON OF FOUR HEARING AID PRESCRIPTION TECHNIQUES USING A COMPUTER CONTROLLED HEARING AID SIMULATION SYSTEM

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ABSTRACT

A computer controlled system for the simulation of hearing aids was designed to allow for efficient and precise manipulation of hearing-aid characteristics. This system was used to compare the frequency-gain characteristics prescribed by four hearing-aid prescription techniques. Speech intelligibility scores and paired-comparison judgments of intelligibility and quality were obtained at three speech presentation levels for eleven hearing-impaired subjects. The effect of frequency-gain characteristic was found to interact with speech presentation level for the three performance measures.

INTRODUCTION

A number of different predictive techniques for hearing-aid prescription have been developed over the years (1, 2, 3, 4, 5). There have been few comparative assessments between these techniques, however, because of the difficulties involved in making such comparisons. Part of the problem is the lack of instrumentation of efficient flexibility to allow for rapid, precise comparisons between hearing aids prescribed by the different procedures. Recent advances in computer technology offer a solution to this problem in that hearing aids can be simulated in real-time with great precision. An additional advantage is that the test procedure can be computer controlled.

MATERIALS AND METHODS

A system of the above type has been implemented using a programmable digital filter controlled by a microprocessor. In this system the patient wears a conventional behind-the-ear hearing-aid case in which is mounted a microphone and hearing-aid receiver. The output of the microphone is fed to a pre-amplifier which then serves as the input to the programmable filter. The output of the filter is fed to a power amplifier and then to the hearing aid receiver worn by the patient. In most hearing-aid evaluations, the patient is seated in a sound-treated test room. For this situation, it is most convenient to use hardwire connections between the test equipment and the transducers worn by the patient. For those cases in which the patient walks about freely, a dual FM transmission link (from patient to test equipment and vice versa) can be used. It is also possible to obtain greater control of the signals reaching the ear by bypassing the microphone and using a direct electrical input to the test equipment.

The central component of the test system is a programmable digital filter. The filter is, in essence, a transversal filter with a maximum of 4096 coefficients. The digital filter has its own analog-to-digital and digital-to-analog converters (12 bit, with a programmable sampling rate). For the purpose of hearing-aid simulation, sampling rates of 15,000 Hz were used. In order to avoid aliasing errors, low-pass filters with a 6000 Hz cut-off frequency were used at both the input and output of the digital filter.

At present the system is designed to simulate the frequency-gain characteristic of a hearing aid. The system is not designed to simulate the internal noise or non-linear distortion generated by a typical hearing-aid. Simulation of these characteristics will be considered after satisfactory methods of measurement have been developed.

In order for the system to simulate a given hearing-aid, the desired frequency gain characteristic is first entered into the computer. Typically, this would be a previously measured frequency-gain characteristic stored on a floppy diskette. Alternatively, an idealized frequency-gain characteristic could be typed into the computer. The frequency-gain characteristic is then modified to take into account any differences in acoustic coupling between the real hearing aid and the simulated version. The filter coefficients for the prescribed frequency-gain characteristic are calculated by the computer, converted into binary form, and transmitted to the digital filter. Once the filter coefficients have been entered, the filter can operate on a stand-alone basis.

In order to use the system for paired comparison evaluations of hearing aids, it is necessary to switch instantaneously from one simulated hearing aid to another. This was done by subdividing the transversal filter into two independently controlled parts, one for each simulated hearing aid.

The test system is not limited to the simulation of hearing aids, but can also be used for generating test stimuli and for measuring hearing aid characteristics. For a more detailed description of the test system see Levitt, et al. (6).

An experiment was performed, using the above equipment, in order to compare four techniques of hearing aid prescription. These were based on the methods developed by Lybarger (2), Skinner (7), Byrne & Tonisson (3), and Levitt, et al. (5). These four methods can be subdivided into two groups, those that depend on threshold measurements (Lybarger and Byrne & Tonisson) and those that depend on loudness measurements (Skinner and Levitt, et al.) Threshold, dynamic range and loudness measurements were obtained for each

subject using the test system described above. Subjects were fourteen adults (mean age = 64 years) with sensorineural hearing loss (PTA = 49 db HL) and a minimum of one year of hearing aid use. The subject's aided ear was chosen as the test ear.

Performance on a speech intelligibility test (Auditec recording of the Northwestern University Auditory Test #6) in noise (S/N=+5) and judgments of speech intelligibility and quality were obtained for each of the frequency-gain response characteristics prescribed by the four hearing aid selection methods. A paired-comparison tournament strategy was used to find the best of the four frequency-gain characteristics. Two tournaments were undertaken for each subject. In the first, the subjects were instructed to select the best frequency-gain characteristic on the basis of the intelligibility of the speech signal. In the second, the subjects were instructed to select the best frequency-gain characteristic on the basis of the quality of the speech signal. The two tournaments and the speech intelligibility tests were repeated at three output levels. The first output level was the level to which an average male voice at a normal vocal effort (8) would be amplified given the prescribed gain for each method. This level was the reference level. The second output level was obtained by increasing the gain of the hearing aid by 10 dB and represented a case in which the speech would be at a loud vocal effort. The third output level was obtained by adjusting the gain of each hearing aid to the subjects' comfort level for speech. Comfort levels were established using a procedure described by Pascoe (4). The presentation of the frequency-gain response conditions were randomized within a speech level condition. The order of speech level presentation was counterbalanced.

RESULTS

A repeated measures analysis of variance of the speech intelligibility data showed a statistically significant interaction ($p < .001$) between frequency-gain characteristic and speech output level. Intelligibility scores were higher at the loud speech level for the two threshold based methods (Lybarger, Byrne and Tonisson) whereas, for the dynamic range procedures (Skinner, et al. and Levitt, et al.), the higher scores were obtained at the normal speech level and at the comfort level. The differences between percent intelligibility scores for the four frequency-gain characteristics were largest at the normal speech level and decreased at the loud and comfortable speech levels. The smallest differences occurred at the comfortable speech levels.

A similar result was obtained for the paired comparison judgments of relative intelligibility at the normal and loud speech levels. The agreement between the two speech intelligibility measures was not as good at the comfort levels where the differences between intelligibility scores were small. When the criterion of quality was used, once again, the dynamic range procedures were consistently preferred at the normal speech level. The agreement between the paired-comparison judgments for quality and the other two performance measures was not as good at the loud and comfortable speech levels. A summary of these results appears in Table 1.

Table 1. Rank order of frequency-gain characteristics as a function of speech level and performance measure (percent intelligibility = %(I), paired-comparison judgments of intelligibility = PC(I), paired-comparison judgments of quality = PC(Q). The frequency-gain characteristics are listed in order from best (1) to worst (4).

	Normal Speech Level			Loud Speech Level			Comfortable Speech Level		
	%(I)	PC (I)	PC (Q)	%(I)	PC (I)	PC (Q)	%(I)	PC (I)	PC (Q)
1	D 66%	D	D	A 73%	A	A	B 69%	A	D
2	C 62%	C	C	B 68%	D	D	A 64%	D	A
3	A 58%	A	A	C 62%	B	C	C 62%	C	C
4	B 42%	B	B	D 60%	C	B	D 61%	B	B

A = Threshold Based Method (Lybarger)
B = Threshold Based Method (Byrne & Tonisson)
C = Dynamic Range Based Method (Skinner, et al)
D = Dynamic Range Based Method (Levitt, et al)

CONCLUSIONS

The rank order of the four frequency-gain characteristics based on performance on speech intelligibility tests and judgments of speech intelligibility and quality changed as a function of speech presentation level. Performance on the speech intelligibility test was best for the dynamic range procedures at the normal speech level and best for the threshold based procedures at the loud and comfortable speech levels. The more efficient paired-comparison techniques provide similar, but not identical results to speech intelligibility tests. The results of this study suggest that similar performance will be obtained for the four prescriptive hearing aid techniques if the listener is able to control the speech level. In spite of the small differences in intelligibility scores for the four frequency-gain characteristics at these levels, listeners are able to distinguish among the frequency-gain characteristics on the basis of speech intelligibiligy and quality judgments.

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VIDVOX: FINAL REPORT ON A FEASIBILITY MODEL FOR A SPEECH RECOGNITION COMMUNICATION AID FOR DEAF PERSONS

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ABSTRACT

The first phase of the Vidvox Project to develop a speech recognition communication aid for hearing impaired persons is complete. The speech recognizer employs continuous speech recognition and prosodic analysis algorithms, and produces phonemic text on a large computer. The Human Factors Investigation has explored whether a deaf reader can learn to read the output of the recognizer at rates usefully close to real speaking rates without an inordinate amount of training, the best strategies for handling recognizer errors, and how best to present the recognizer's output to the reader.

INTRODUCTION

In 1981, Sensory Aids Foundation initiated development of a communication aid for profoundly deaf individuals. The target device, designated Vidvox (Latin: video = I see, and vox = voice) would display spoken speech in the form of a stream of phonemic symbols.

After initial surveys to determine the state of the art of speech recognition technology and pertinent human factors research (1, 2), the project was divided into two parts. The first was aimed at establishing the feasibility of a phoneme recognition computer algorithm which would convert speech to a string of phonemes for display as phonetic text; the other was to determine whether a deaf reader could learn to read the output of the recognizer at rates usefully close to real-time speaking rates without an inordinate amount of training, and how best to present the phoneme string on a display. One assumption, based on results and assumptions from past Advanced Research Projects Agency (ARPA) projects, was that a recognizer with 60% or greater accuracy would be adequate.

The Vidvox Feasibility System development was conducted by Bolt Beranek and Newman Inc. (BBN), Cambridge, MA, with Dr. Michael Krasner as principal investigator. The Human Factors Investigation was a joint project between BBN and The Center for Communications Research, Inc., Rochester, NY. Drs. A.W.F. Huggins and Robert Houde were co-principal investigators. Sensory Aids Foundation defined, directed and funded the projects. Funding was provided by a gift to Sensory Aids Foundation.

This paper contains material presented in the Final Reports of the two projects (3, 4).

BACKGROUND

Printed text is a form of information transfer in which language can clearly be perceived at rates comparable with those of speech. Reading rates of skilled readers are often as high as 300-400 words per minute, comparing favorably with speech rates, which typically range from 120-180 wpm. The effect of removing the spaces in English text has been reported to reduce skilled reading speed to about 50% of its normal value (5). Thus, if it were possible to perform real-time computer speech recognition on speech, the resulting text could also be read in real time. Unfortunately, the present state of the art in speech recognition suggests that real-time recognition of unrestricted fluent speech will not be feasible for some years except for applications that involve very small vocabularies.

Vidvox was proposed to fill this gap. The initial two projects were intended to determine the feasibility of development of a phonetic-syllable speech recognition communication aid, and to examine the human factors requirements for using such an aid.

The first of the two Vidvox projects, the Feasibility System, developed, refined, and implemented speech recognition algorithms on a large computer and measured typical error rates (3). Performance was not in real-time. Error rate, type, and effect on reading rate were major research interests.

The second project investigated the human factors requirements for using a device such as Vidvox, (e.g., the rate for reading phonetic text, effects of segmentation, errors, spacing, font, etc. (4)).

People can clearly learn to read phonetic text when word boundaries are marked by spaces in the printed text and the text contains no errors. A European journal, Maitre Phonetique, publishes papers entirely printed in International Phonetic Alphabet symbols. The effect of removing the spaces in English text has been reported to reduce skilled reading speed to about 50% of its normal value (5), but if Maitre Phonetique can be read at high or even normal reading rates, the implication is that this might still leave a rate sufficiently high to read phonetic text at rates typical of speech.

When phonetically-spelled syllables transcribed by a stenographer and processed by sound-spelling correspondence algorithms were presented without word boundaries (and included 5 to 15% operator keying errors), a few readers could still read the results at useful speeds if sufficient training time were allowed. The stream of phonemes was segmented into syllables and the segmentation was performed by a human operator who understood the message. The error rates were low and probably reasonable, since they were made by a human listener/transcriber (6, 7).

However, Vidvox recognition is performed by a computer algorithm, and the results are not space-

separated, error-free phonetic script, nor are the segmentation, error rates and error types typical of human listening or speech errors.

MATERIALS AND METHODS

Feasibility System

The phonetic recognition approach selected for the Vidvox Project was a recognition system in which acoustic-phonetic features and measurements of the speech spectrum were combined in a powerful statistical formalism, the hidden-Markov model (HMM). In addition, further development combined these with a triphone context model, in which the middle phoneme was conditioned on the two (left and right) adjacent phonemes, or, in the event one or both of these were not present, selecting an appropriate left, right, or context independent model. The technical report prepared by Krasner contains a detailed description of the system, including the training data used to implement the HMM algorithms (3). (For a more complete mathematical review of HMM's, see (8).) The algorithms were developed on a VAX 11/780.

Recognizer performance with only 30 minutes training data and 256 spectral clusters resulted in 81% correct phones; 94% class correct; 11% substitution errors (in class); 9% substitution errors (out-of-class); 3% deletions; and 13% insertion errors. The overall performance was significantly higher than any previously published reports, and this was accomplished with very little training.

Human Factors Investigation

The overall aim of the project was to investigate whether deaf readers could learn to read the output of the speech recognizer. The materials selected for training ideally should be generated by the recognition system being tested. However, the Vidvox Feasibility System ran many times slower than real time, and was being developed in parallel with the human factors work.

The training materials therefore were generated using Speech Plus's CallText, which can convert machine-readable English text into synthetic speech. For our purposes CallText could also output the string of phonetic symbols derived internally as a command string for an analog speech synthesizer.

Test materials consisted of short stories published as reading instruction materials for 11th and 12th grade high school students. Many stories were available, they were on high-interest topics, and they were controlled for language structure and vocabulary level.

The experiments consisted primarily of tracking tasks. Test sentences from the stories were presented on a BBN BitGraph advanced graphics terminal. The subject controlled the rate at which new sentences were presented. The time between successive sentence call-up and the numbers of words and syllables in the sentence were recorded on the computer so that a reading speed could be calculated for each sentence. Experimental sessions lasted about an hour each, with subjects reading between 50 and 500 sentences in a session. The useful reading rate lower limit was considered to be 30-35 wpm.

Eight experiments were used to arrive at our results. Testing was done with hearing and non-hearing subjects with high-level language skills.

The objectives of the first three experiments were to determine which phonetic alphabet and segmentation scheme were easiest to learn to read. Two alphabets (one based on Webster dictionary pronunciation marks, the other a form of regularized English (Regfont) which spells phonemes as they often appear in regular English text) were selected for possible use.

Testing was done both without segmentation and with "frequency" segmentation based on a suggestion by Victor Zue in which a hyphen was placed before each longest legal initial consonant cluster. An exceptions list for commonly occurring segmentation errors would improve this scheme, but was not implemented in this project (9).

The general findings were that Regfont with frequency segmentation was easiest to learn to read. The deaf subject's performance was well within the range of that for hearing subjects, and both groups read within the useful range.

The next two experiments examined the effects of slurred speech and inclusion of various levels of "human-like" speech errors on reading rates and accuracies. Reading rates ranged from an average of 25-30 wpm for 1/3 errors to as low as 14.1 wpm for very "errorful" material. When substitution and insertion errors were replaced with a single symbol (the pound sign #) to assist in quickly identifying phonemes containing errors, the errorful material was somewhat more readable.

The usefulness of prosodics was examined in Experiment 7. Underlines were used to mark stressed syllables. Of the seven hearing impaired subjects who read with stress marks, three read faster and four read slower than the baseline. On average, reading rates dropped 4.47 wpm with the stress marks. The consensus was that marking syllable stress was not a large factor in making phonetic text more readable.

Simulation of an interactive system was explored in Experiment 8 with three hearing subjects. The story material was run through an error simulation program which generated a different set of errors for each of eight copies of each sentence. Subjects looked at as many of the copies as necessary to decipher the sentence, piecing together the readable parts of each copy to form a whole. Reading rates were very slow, ranging from 13.3 wpm to 6.3 wpm for three subjects. Subjects found the task tedious and would not want to use the system in the real world.

A second set of tasks was set in which errors of the type and frequency produced by the speech recognizer (machine errors, not "human-like" speech errors) were simulated and inserted. Deciphering the text with recognizer errors proved difficult because the reader had no information about which phonemes might be in error. The computer errors were not acoustically reasonable, not

the same types made by human listeners, due to the nature of the HMM recognition algorithms. The reading rates were too low to be useful, and the approach to reading degenerated to puzzle- or problem-solving, rather than reading.

RESULTS

Speech Recognizer

The overall recognition rate of 75%-80% phonetic accuracy achieved by the recognizer alogorithm exceeded previously published results for a single-speaker continuous speech recognition system, and was above the assumed required accuracy of greater than 60% set for the Vidvox Project (10). The recognizer development significantly advanced the state of the art of phonetic recognition, which is the basis of continuous speech recognition. The demonstration system produced phonetic displays for testing purposes, though much of the testing was done on simulated systems because of parallel development of the recognizer with the human factors investigation.

Human Factors Investigation

The results of the first four experiments show that, with a reasonable amount of practice, post-lingually deaf readers can learn to read error-free phonetic script segmented into syllable-sized chunks (but without word boundaries) at useful rates. Error rates such as those typical of the Feasibility System speech recognizer render the system less than acceptable. Substituting a single common signal for all erroneous phonemes helped, but required highly accurate identification of the phonemes in question. Marking those vowels that carried stress helped some readers, but hindered others. When errors were machine-generated, not "human mistake" type, and were unpredictable, readability dropped to well below useful levels, even with highly motivated test subjects and high-accuracy recognizer performance. One project investigator's estimate of required recognizer accuracy for useful readability was that it be greater than 95%.

What was not investigated was whether or not the recognizer makes patterned or consistent errors. Knowledge of such patterns and long-time use might increase reading rate for some users.

The Human Factors Investigation Final Report recommendation was that the best future approach for a communication aid probably lies in using improved word-level recognizers instead of using the phonetic string approach.

Sensory Aids Foundation and the Vidvox Project principal investigators are unaware of any other human factors investigations similar to those explored in this project. It is hoped that the experiment designs and results will prove valuable to other researchers addressing communication problems of deaf persons and the use of speech recognition technologies in rehabilitation. Abstracts and copies of the project Final Reports containing full descriptions and results are available to those requesting them. Interested parties should direct their requests to Yvonne Russell at Sensory Aids Foundation.

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STIMULATING THE MANUFACTURING AND DISTRIBUTION OF REHABILITATION TECHNOLOGY: ECONOMIC AND POLICY INCENTIVES AND DISINCENTIVES

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ABSTRACT

Interviews were conducted with industry officers to ascertain perceptions of the potential effects of various economic factors and governmental policies. An attempt was made to identify disincentives to industrial participation in the field of rehabilitation technology and to discover incentives that might promote increased commercial activity. Findings are reported for the following issues: Financing technology acquisition; product identification, selection and evaluation; market data, federal regulations, patents, professional education and training, and federal funding for development and manufacturing activities.

INTRODUCTION

The slow diffusion of quality rehabilitation technology into common usage by people with disabilities has been a major concern of many groups: industry, rehabilitation professionals, potential users, and even the United States Congress. In 1982, the Congressional Office of Technology Assessment (OTA) released a report which discussed its conclusions relating to the production, marketing and diffusion of rehabilitation technologies. The report, entitled Technology and the Handicapped, presented a series of options which might be taken by Congress to stimulate the development and diffusion of innovative technology intended to increase the independence, productivity and quality of life for people with disabilities. A significant proportion of these options concerned means to stimulate the involvement and commercial success of industries engaged in the production and distribution of these technologies.

This paper describes preliminary findings of one project designed to study basic economic factors and policies which affect industrial efforts to operate successfully in the rehabilitation technology marketplace. This study has had, as a secondary goal, the validation of the assumptions and conclusions of the OTA report. This investigation has taken the form of an in-person, in-depth interview with key officers of industries with extensive experience in the manufacturing and distribution of rehabilitation technology.

BACKGROUND

As an integral part of the Electronic Industries Foundation Rehabilitation Engineering Center's primary goal of increasing the availability of appropriate rehabilitation technology, research was designed to study the effects of economic issues and public policies upon company decision-making and practices among the industries which produce and distribute these products. The study had as one of its missions the investigation of the disincentives to industrial participation in the field of rehabilitation technology manufacturing and distribution. Simultaneously, efforts were made to examine potential incentives which might contribute to the amelioration of these existing disincentives and, when implemented, result in increased industrial participation this field.

METHOD

A list of approximately 40 industries which represented a broad range of rehabilitation and medical manufacturers and distributors was created for possible inclusion in an in-person survey. The companies represented diversity on several continua--disability groups served, size of annual sales, kind and extent of manufacturing capabilities, form of marketing and distribution employed, and degree of third-party payment coverage anticipated.

An interview instrument was developed to explore the disincentives perceived by corporate officers to limit industrial participation in the rehabilitation technology field and to elicit comments regarding possible incentives which might remedy existing deterents to involvement.

Among the topics addressed in the instrument form used in interviews conducted to this time are the following:

Type of product line, size of annual sales volume, motivation for entering rehabilitation market, manufacturing capabilities, distribution approaches used, professional organization participation, identification and selection of new products, product evaluation, marketing strategies, role in professional and user training and equipment service, perceived effects of government regulations (e.g., FDA or FCC), source of marketing data, source of procurement finances, source of research and development funding, perceived role of federal intervention in research and development and/or product engineering funding, perceived role of federal intervention in stimulating market through financial assistance to end-users for technology acquisition, patent policy, effects of liability or other legal issues, potential role of large industries in rehabilitation field, potential effect of increased public relations activities in rehabilitation technology field, general views of other major issues and problems relating to industrial involvement with rehabilitation technology.

The preliminary findings reported in this paper

are based upon interviews with corporate officers in 13 of the original list of industries. Each officer was either the chief executive officer or designated product or marketing manager. Researchers believe that these 13 companies represent the broadest possible array of companies involved in the manufacturing and distribution of rehabilitation and medical technology. Annual sales for these corporations range between an estimated $500 thousand and $2 billion with the median being between approximately $15 million and $20 million. The small number surveyed to date makes it impossible to draw firm conclusions regarding the perceptions and attitudes of the industry as a whole, but certain trends have already emerged and warrant cautious citation in this report.

TENTATIVE FINDINGS

Interview responses differed significantly when discussions concerned products and practices. Attitudes toward policies, however, evoked a high degree of concurrence. The following paragraphs summarize opinions of 13 corporate officers interviewed. Minority opinions will be noted. Nine topics will be reported.

Financing Technology Acquisition

The limited financial resources available to the end-users of rehabilitation technology were identified by nine of the thirteen persons interviewed as being the primary problem confronting industries concerned with the manufacturing and distribution of products in this field. Two of the remaining four placed this issue as the second most critical issue. These eleven interviewees felt that public sector efforts were needed to stimulate the market by creating improved funding programs for disabled individuals. Limited tax credits, government backed guaranteed loans, low interest loans, and broader third-party payment coverage through Medicare and Medicaid were the most frequently supported concepts.

Eleven of the thirteen interviewees addressed each of the funding options to some degree. Regarding partial tax credits for end users, several interviewees supported a 50% tax credit with a maximum dollar ceiling. One interviewee felt that tax credits should be limited to technology used in employment. The issue of Medicare and other third-party payments also received several additional comments. Primarily, they were in the form of advocacy. Four interviewees stated that the current application of medical necessity regulations emphasizes the acquisition of less expensive, and therefore, frequently lower quality technology. These devices often must be replaced because of a shorter unit life cycle. It was recognized that a balance between cost containment and human need must be struck, but these efforts have recently often tended to be short-sighted. Further, interviewees stressed that "life enrichment" devices, health and "wellness" devices, as well as "medically necessary" technology should be covered by Medicare. However, there was emphasis placed by one person to insure that "convenience" devices be excluded. Finally, one interviewee stated that no company should have products covered unless it could offer training and service support for their products. One interviewee stated uncategorically that public insurance programs should not be used to cover additional rehabilitation technology.

Product Identification

Twelve of the companies do not perceive identification of new products to be a problem. Most of the interviewees stated that new ideas are brought to the firm each week for consideration. Developers seek them out. The vast majority of the devices brought for consideration are eventually rejected either because they are not commercially viable or because they do not fit existing product lines. One interviewee, however, stated that there have been difficulties in dealing with developers because many do not understand that rehabilitation technology rarely produced wealth. Other developers refuse to permit alteration of a design.

All of these companies stated that internal research and development is an ongoing activity. Each believed that the company worked closely with the potential consumer market, both the end-users and the rehabilitation and medical professionals. Thus, they believed that product identification was handled adequately within the company.

Product Selection

Twelve of the interviewees stated that the two most important criteria used in the selection of new products were the following: (1) projected return on investment when weighed against anticipated risk, and (2) insuring that the new product fit into existing product lines thus improving the delivery of distribution and customer services. One of the interviewees emphasized that the selection of new products was based primarily upon the determination that customers would have access to financial resources.

Product Evaluation

All companies stated that internal evaluation of products takes place prior to releasing them into the market. Three of these companies indicated that the less sophisticated aids received only cursory evaluation by a few potential customers, professionals, or dealers. All companies indicated that the "high tech" devices were evaluated by staff with the aid of external consumer and professional assistance. Nine interviewees stated that federal funds should not be used in rehabilitation technology evaluation for several reasons. First, the evaluation cannot be timely enough to meet the needs of the industries. Second, these evaluations emphasize efficacy rather than user acceptance, a fact which can lead to release of a non-commercially viable product. Third, this effort is duplicative to regular activities of a company seeking valuable products in a small marketplace. Further, the limited federal funds for rehabilitation technology should be used to stimulate the market rather than evaluate

devices. Three interviewees did state that external technical evaluation was needed, but even this evaluation should be done in conjunction with the developer to facilitate inclusion of refinements and to permit observations of user acceptance and enthusiasm as well as performance.

Market Data

All interviewees stated that it would probably be helpful to have improved marketing data, but no company felt that it was essential. Each felt that the company had a handle on the market size. The difficult, and intangible question, was how many people might actually buy a specific device. Demographic data could not help answer that question. Four companies stated that data regarding the potential market would have been very valuable at the time sales first began. Experience in the field has produced internal knowledge of the field although the information is not systematized.

Federal Regulations

None of the interviewees felt that federal regulations under the direction of FDA or the FCC produced disincentives for involvement in the rehabilitation or medical technology fields. Most felt that compliance was not a problem once the agencies were regularly consulted as part of normal procedures. One company, for instance, reported that a product was dropped from consideration early in the company's existence because of the fear of attempting to comply, but subsequent experience eliminated this concern.

Patent Considerations

All companies presented a somewhat ambivalent position toward patents. All believed that patents were important. They would be sought whenever possible, and patented products brought to them for consideration received extra attention. Patents, for instance, were important to reduce the time before competition might appear. They might also be helpful in dealing with the international market.

Professional Training

The four interviewees mentioned the issue of professional training as one of the most significant problems confronting the industry. Professionals involved in the selection and prescription of new technology are perceived by these interviewees as lacking the knowledge necessary to promote appropriate aids and devices. One individual, for instance, said that no efforts should be made to increase third-party payment coverage until a mechanism could be established to train these rehabilitation and medical professionals systematically in the performance of rehabilitation technology. Regular in-service training programs are needed even though training is already considered to be a major part of the companies' marketing budget.

Federal Funding for Development and Manufacturing Activities

Eleven companies expressed strong reservations regarding extending federal funds for the research and development of rehabilitation technology by private companies. The concerns centered around two points: objective administration and timeliness of support. Most interviewees stated that the federal government cannot be expected to have knowledge of current priorities in the rehabilitation technology field. Who, it was asked, could set the criteria for selecting which company or product was to receive support? Most of these companies have had federal R&D funds in the past, but concern over the attachment of strings was repeatedly expressed. Four representatives mentioned the desire to hold research findings proprietary. Further, the four companies mentioned the issue of timeliness. The time between selecting a project area of interest, requesting funds for initial work, and the actual acceptance and granting of an award is too lengthly for a company to keep competitive.

The issue of federal support for manufacturing was also addressed. Nine companies expressed interest in having some federal support for tooling costs, the making of molds, or other initial costs associated with production. Five companies, however, strongly emphasized that cash awards were not wanted because strings would undoubtedly be attached. Rather, special investment tax considerations were thought to be appropriate for companies engaged in rehabilitation technology endeavors, efforts aimed at human need, a socially redeeming activity. It was pointed out that other industries receive special investment tax credits for other, less human-related activities. Two of the thirteen, on the other hand, stated that all companies should be treated equally under federal laws and regulations. Thus, no federal support should be granted to rehabilitation technology firms either for development or manufacturing, either through awards or tax considerations.

In summary, the companies tended to agree in principle that the demand-side of the market needed more public stimulation than did the supply side. Further analysis of the data are planned, and additional verification of the findings will be sought through further administration of the survey instrument.

ACKNOWLEDGEMENTS

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6.2 The Design and Testing of a Tutorial Manual That Allowed Users To Learn a Multifunction Aid System Without Expert Support

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Abstract: To see if a complex aid system could be learned from a manual without direct support from experts, a special tutorial manual was designed for a multifunction, computer-based, communication aid system called the Trine System. During the first two weeks of the Trine System's beta test, 13 users with their parents, therapists, and teachers were asked to learn the Trine System with the guidebook without contact with people who knew the system. After two weeks, six of the users were asked to demonstrate what they had learned. Five of the six users, as well as the father of the sixth user, had mastered the basics of the system.

In addition, five of the users and ten parents, teachers, and therapists were asked,"How likely is it that a user could learn the Trine System by using it with the guidebook without personal help from the supplier?" Eleven of these people rated this as "Very Likely" and the other four rated it as "Likely". No one rated it as "Unlikely" or "Very Unlikely".

The primary goal of the guidebook design was to allow people with normal intelligence who had no computer or complex aid experience to learn the Trine System without expert help from a printed document as physically accessible as possible to the disabled user. Secondarily, the guidebook was designed for small manufacturers to easily produce, maintain, and update in very small printing runs. To accomplish these goals, the Trine guidebook design makes use of structured documentation techniques, positive multimode presentations, and desktop publishing. As a result, the design includes many features not often found in aid system manuals. This study suggests that emulating these features may help users get the most from their aid systems and reduce the cost to manufacturers and dealers of personal support.

Question: Can Better Manuals Allow Users to Learn Aid Systems Without Expert Help? Aid systems are often underutilized because users are unable to learn from the documentation provided and are unable to get the expert support they need locally. Many users can't get expert help locally because they are in less populated areas and/or they are dealing with a manufacturer who has no local aid system integrator. In this case, many manufacturers are forced to provide aid systems via the mail and support them directly. For manufacturers who must use their technical experts to support systems directly, this can be doubly bad. It not only adds to the cost of the systems, but by tying up their designers, it also slows down the development of improved systems.

It is often suggested that if manufacturers would spend more effort developing quality tutorial manuals, they could mitigate these support problems. The better the manual, the less direct support they have to do. In the best case scenario, the tutorial manual would be good enough to allow users along with their local support group (parents, teachers, therapists, etc.) to master their aid systems without any additional help from experts on that particular aid system.

Testing: The Trace Center Tested the Idea In Developing the Trine System. The Trace Center decided to test the idea that a good tutorial manual would allow users to learn a complex aid system without help from anyone already an expert on that particular system. As a result, two major activities were added to the development of a portable computer based communication aid system called the Trine System. First, great emphasis was put on the design and production of a tutorial manual that we call *A Journeyer's Guide to the Trine System* (hereafter referred to as the Trine guidebook). Second, part of the beta test of the Trine System was designed to see whether or not people could learn to use the communication aid system from the guidebook and their own experimentation, without help from people familiar with the system.

What Is the Trine System Like? In order to understand how effective the tutorial guidebook it is necessary to understand what the Trine System is like and what there is to learn. The Trine System is a multifunction aid system whose primary components are a software program, an Epson HX-20 portable computer, and an optional speech synthesizer. It has three main functions: providing a portable writing system; providing a portable speech output system; and providing access to a standard computer through a standard keyboard emulating interface.

The Trine System is designed to be easy to use by beginners while providing flexibility for more advanced users. It uses menus and dedicated keys for most functions. However, it's flexiblity means there is a lot to learn if a person wants to master everything. It has 143 different menus and there are about 33 functions on dedicated keys including 22 editing functions. Four of the keys cycle sequentially through three different states. The Trine System offers six different writing workspaces, a conversation workspace, and keyboard emulating interface workspace. It can use a built-in or external printer,and a built-in or external speech synthesizer. It has a two mode abbreviation expansion system that can store about 400 twelve character user programmable expansions. A built in cassette drive stores abbreviations as well as up to 18 pages of writing organized into three different volumes. Almost all program parameters are user settable. To cover all these features the guidebook is itself 327 pages with 123 major topic articles.

Who Were the Test Users? As part of the Trine System evaluation a beta test was carried out with six users from Wisconsin (Level 1 users) and seven users from other areas (Level 2) users. Figure 1 profiles the test users, showing their sex, age, grade level, and whether or not they walk or use a speech synthesizer. Level 1 users ranged in age from 11 to 40 and in grade level from third to the second year of college. Users in Level 2 ranged in age from 10 to 47 and in grade level from third through college. Each group had one female. Two of the level 1 children walk, and one Level 1 person along with three Level 2 persons use a speech synthesizer with their Trine System.

User	Sex (Female / M)	Age	Grade	Ambulatory (Yes / N)	Uses Speech Synthesizer (Yes / N)
			Level 1		
#1	M	11	3	N	Yes
#2	M	40	4	N	N
#3	M	11	5	Yes	N
#4	Female	11	5	Yes	N
#5	M	15	4	N	N
#6	M	30	14	N	N
			Level 2		
#7	M	47	16+	N	N
#8	M	15	9	N	N
#9	M	10	3	N	Yes
#10	M	27	12	N	Yes
#11	Female	17	11	N	Yes
#12	M	17	3	N	N
#13	M	15	9	N	N

Figure 1: Profile of Test Users

An important point to keep in mind about the beta test users is that in order to participate they had to already have an Epson HX-20 computer because the project could not afford that many computers. This means that they all had some experience with the Epson computer before they got the Trine System. Thus, they were not naive users. However, we feel that their experience still relates to other beginning users because the Trine System is very different from software that they used previously. In fact, they may have had some extra problems due to expectations about how they thought it should work from their other experience .

The Test Users Had Only the Trine Guidebook For Support During the First Two Weeks: The overall beta test was designed to evaluate the Trine System as a communication aid system and the results are given in the Trine beta test report[1]. One procedure of the beta test was designed to see if the six Level 1 users could learn the basics of the Trine System without help from experts. These people were

given the Trine System and the guidebook and asked to learn it without help from the Trace Center. If they called with a problem they were preferred to the guidebook in general, not to any particular section.

Results: Five of Six Users Mastered the Basics In Two Weeks. After two weeks the Level 1 users and their associated support persons came to the Trace Center for evaluation. Susan Fishman, beta test coordinator for the project, then tested them to see whether they had mastered the basics of Trine System use. The following is her conclusion in the beta test report:

After two weeks of using the Trine System without any direct training by the Trace Center,subjects in Level 1 were asked to demonstrate basic functions of the aid. Five out of six of the subjects were able to get into different workspaces, delete characters, lines, and pages, move the cursor in all directions, insert words and print on the built-in printer. In the case of the one subject who was not able to perform all these functions the father was able to demonstrate them. The skill demonstrated by either the subjects themselves or their primary trainers suggest that it is possible to learn the basic operation of the Trine System using the training manual in conjunction with hands on exploration without external support or training. All subjects in Levels 1 and 2 agreed that they were provided with sufficient information to learn the basic functions of the Trine System and did not require any direct training. In addition, three subjects had incorporated the system into their classes or daily environment within two weeks.[1]

What The Test Users, Their Parents, and Clinicians Thought About Learning From the Trine Guidebook: Between the two and four week point Level 1 and Level 2 users were interviewed on their thoughts about the guidebook. In three cases parents and clinicians were interviewed instead of the users because they were learning the system themselves and then teaching the users who were at the third and fourth grade level. In addition, two parents and five clinicians who also learned the Trine System were interviewed.

The five users, five parents, and five clinicians interviewed about the guidebook were asked, "How likely is it that a user could learn the Trine System from using it with the guidebook (with help from family, friends, and inexperienced clinicians) without personal help from the the supplier." They were given four choices: Very Unlikely; Unlikely; Likely; or Very Likely. The results are shown in Figure 2. Eleven of these people rated this as "Very Likely" and the other four rated it as "Likely". No one rated it as "Unlikely" or "Very Unlikely".

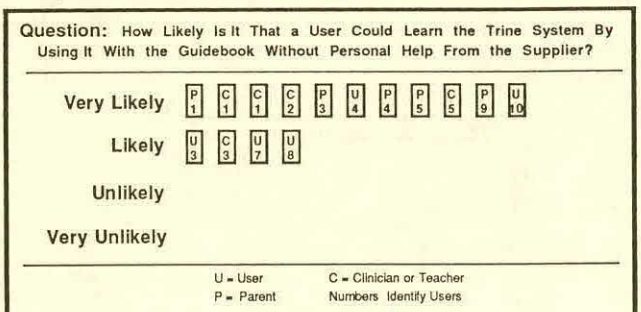

Figure 2: Results of Interviews

Thus, users, parents, and clinicians strongly endorsed the idea that the Trine System could be learned from a tutorial manual without the need for training by experts in the system. Considered with the results of the beta test finding discussed above, this indicates it is likely that manuals similar to *A Journeyer's Guide to the Trine System* can reduce the need for personal training and other direct support for other complex aid systems.

Design: The Trine Guidebook Design Makes Use of Structured Documentation Techniques, Positive Multimode Presentations, and Desktop Publishing. The primary goal of the guidebook design was to create a printed document as physically accessible as possible that would allow a disabled user with normal intelligence, a sixth grade reading level, and no computer or multifunction aid experience to learn the Trine System without expert help. Secondarily, the guidebook was designed for small manufactures to easily produce, maintain, and update in printing runs of as few as 50 copies. To accomplish this three major approaches were incorporated into the design.

The first major approach used is called structured documentation. The author learned the structured documentation approach from a 2 day workshop, Structured Documentation, presented by The American Institute for Professional Education (Carnegie Bldg., 100 Kings Rd., Madison, NJ 07940, (201)377-7400). According to the presenter, Lynn Harris, structured documentation evolved from the intense documentation and training requirements for computer software in companies like IBM, Wang, and Digital Equipment Co. Its overriding goal is to make it possible to design, produce, and maintain effective documentation in the most cost effective and time effective way. Structured documentation can be done successfully by any size company. The three main features of structured documentation are that it is: 1) modular, 2) structured in a top-down fashion, and 3) written using clear communication techniques

The second major approach was to design, write, typeset, print and assemble the guidebook in-house using the Apple Macintosh computer system. This has since come to be called desktop publishing. Because most small aid system companies print 100 or fewer manuals at a time, the cost to produce quality tutorial manuals using traditional publishers and bookmaking techniques is prohibitive. Desktop publishing offers the possibility to aid system companies of making quality manuals. The overall design and layout of the guidebook was strongly determined by technical limitations of the Macintosh. It turned out that these limitations severely increased the effort required to create the guidebook. Happily, most of these limitations have been removed in the latest equipment.

The third major approach was to incorporate new ideas about different cognitive learning styles by using a positive multimode presentation of key information. To communicate effectively to a wide variety of people, the guidebook presents the same information in a variety of ways. Information was always presented positively in the logical sense. Negations were actively eliminated. The five main types of presentation were: 1) a summary, 2) straight text, 3) a follow along example, 4) a realistic illustration of the actual device, and 5) a tabular instruction list. This redundancy is relatively cheap when implemented with desk top publishing equipment. It was found from interviews with the learners that even in our small sample of 15 that the group was almost evenly split for their primary learning among all five types of presentation. It is felt that the positive style and redundancy in the guidebook contributed greatly to its success. This is like the organization and redundancy in a Space Shuttle. It's a necessary redundancy if it is to work.

The Trine Guidebook Includes Many Features Not Often Found In Aid System Manuals: The structured documentation techniques, positive multimode presentations, and desktop publishing approach led to a tutorial manual design with features not often found in aid system manuals. Figure 3 shows a typical topic layout. In this paper features of the guidebook design can only be briefly listed:

Overall Features of *A Journeyer's Guide to the Trine System*:

1) Modularity
2) A Multimode Presentation
3) Textual and Visual Redundancy
4) Topics Defined by the Actual Tasks a User Wants To Do
5) Each Topic Is Self Contained
6) Reader Encouraged To Only Look At Relevant Topics
7) Directly Addresses Reader
8) Positive Statements Only
9) Written For a 6th Grade Reading Level
10) Descriptive and Mnemonic Titles
11) Descriptive Headlines
12) Summaries In Each Topic
13) Examples For Each Task
14) Descriptive and Mnemonic Page Numbers

Physical Features:

1) Made With Half Sheet Size Three Ring Binders
2) Binders Have Slip In Covers
3) $6^1/_4$" x $8^1/_2$" Pages
4) Large $^5/_{16}$" Binder Holes and Sheet Lifters
5) Pages Printed On Colored Copypaper Stock
6) Tabs Printed On Colored 110# Cover Stock With Mylar Reinforcing

Organization Features:

1) Six Distinct Color Coded Parts: 1 Opening, 2 Using the Trine System, 3 Troubleshooting, 4 User Tips, 5 Connecting & Adjusting, 6 Reference & Index
2) An Instant Demonstration On the Very First Page
3) Tabs At Every Section
4) Complete Table of Contents At Its Own Tab
5) The Contents of the Section Listed At Its Beginning
6) The Main Points of the Section At Its Beginning
7) The Contents of Adjacent Sections Listed At the Beginning of Each Section
8) An Expandable User Tips Section For Notes With Helpful Tips Written By Actual Users
9) A Distinctive Red Troubleshooting Section At Its Own Tab
10) A Complete Glossary At Its Own Tab
11) A Complete Index At Its Own Tab

Layout Features:

1) A Basic Layout Repeated For Each Topic
2) Each Topic Has Text Page and a Facing Visual Exhibit Page
3) Headlines 18 pt. Bold Times-Roman Type (72 pt. to an inch)
4) Ruler Lines To Set Off 12 pt. Helvetica Summaries
5) Text 10 pt Helvetica With Bolded Keynames and Menu Items
6) Eye Catching Symbols In the Left Margin Marking Special Paragraphs
7) Exhibit Page Has 14 pt. Caption
8) Realistic Screen Illustrations
9) Realistic Keyboard Illustrations
10) Tabular Instruction Lists

Production Features

1) Designed In-House using the Apple Macintosh and Macpaint
2) Typeset In-House Using Apple Macintosh. The edition tested was done on 80 dpi Apple Imagewriter and the second edition (Figure 3) was done on 300 dpi Apple LaserWriter.
3) Printed Via Photocopier (Except Tabs Which Were Offset Printed)

Conclusion: This study suggests that tutorial manuals like *A Journeyer's Guide to the Trine System* with the features listed above can allow users of complex computer based aid systems to learn these systems without expert help. This type of manual might allow more users to get the most from their aid systems and might reduce the cost to manufacturers and dealers of personal support.

1. Beta Test Report for DOE Grant 300-83-0267, Technology Compensatory Activities, July 1985, Trace Research and Development Center, University of Wisconsin-Madison.

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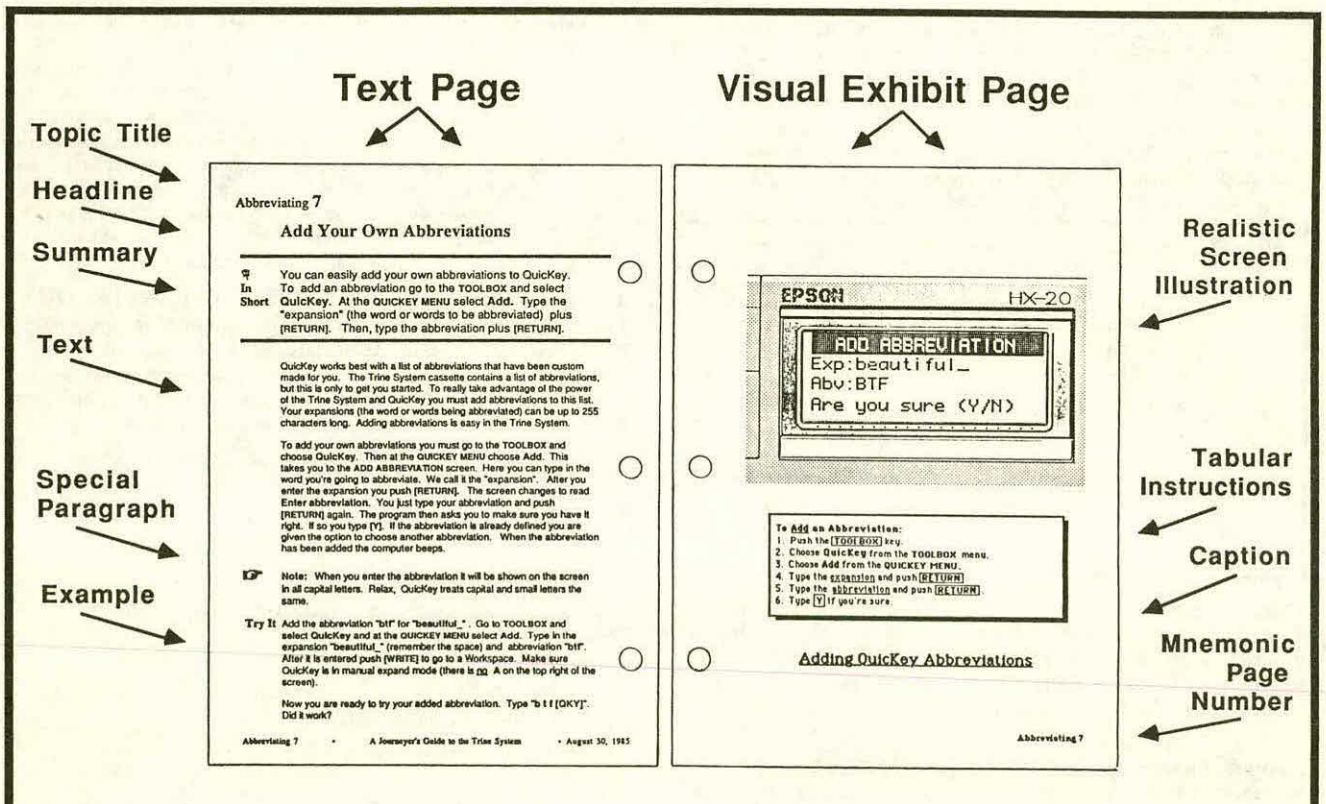
Address: Trace Research and Development Center, 1500 Highland Avenue, Madison, WI 53705, (608)263-5683



Representative Topic Layout

Figure 3

AN EVALUATION PLAN FOR COMMERCIALIZATION ENHANCEMENT OF NEW REHABILITATION PRODUCTS

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ABSTRACT

The paper emphasizes the significance of integrating objective evaluation components into the rehabilitation product development and commercialization processes. A general plan that includes a discussion of the functions served by informal field trials and formal clinical evaluation of user ready rehabilitation products is proposed.

INTRODUCTION

It has been clearly demonstrated that technology increases opportunities and alternatives for independence among disabled people. However, the methods for addressing and, hopefully, accomplishing commercialization of technology applications are not so clearly defined. The process appears lengthy, complex, and encumbered with seemingly endless hurdles. As repeatedly indicated, the need to develop and refine methods that will enable efficient advancement of appropriate and functional technical aids into the market still remains.

A component of the product development process that many times appears weak, or maybe even missing, is a plan for objective evaluation. Though the performance of extensive evaluation is still not a guarantee for the product successfully reaching the market, it does offer some insurance through realistic, objective matching of the proposed solution to the problem.

To help deal with the issues associated with rehabilitation technology evaluation and product advancement, the Southwest Research Institute Rehabilitation Engineering Center (REC) was established in mid-1983 through a cooperative agreement with the National Institute of Handicapped Research of the U.S. Department of Education. The REC focuses its research on the development of methods and criteria for evaluation of new rehabilitation products and clinical techniques to advance their commercialization. Activities of the Center primarily relate to engineering testing, safety analysis, user field trials, and clinical evaluation.

THE EVALUATION PLAN

The proposed plan for rehabilitation product advancement is based on the Center's experience with previous demonstration evaluations and informal discussions with rehabilitation product manufacturers. It was developed collaboratively by staffs of the REC and the University of Texas Health Science Center-San Antonio, Department of Physical Medicine and Rehabilitation. This conceptual plan emphasizes a two-level approach to user evaluation of new product developments. The plan was devised to distinguish between the functions served by the informal user field trial and the formal clinical evaluation. Evaluation activities conducted at both levels offer specific information applications that are critical to decision making at different developmental milestones of the product.

Frequently, a product developer (commercial manufacturer or independent inventor) requires objective evaluation information and expert opinion quickly and within a defined time constraint. This brief, but intensive effort has been termed user field trial evaluation by the proposed plan. An objective broad view of device effectiveness, potential applications, and user acceptance is gained on a short-term level. The information obtained from this experience may, in fact, later form the basis for the study design of the formal clinical evaluation.

Relatively similar information may result from the performance of a formal clinical evaluation. However, this activity is designed to investigate specific aspects of device safety, effectiveness, reliability, and utilization according to a formally approved protocol. It may be conducted as a multi-facility study or even examine long-term utilization patterns. By its very nature, the formal clinical evaluation was never intended to be accomplished quickly or offer ready response.

When the REC staff initially identified and then examined the merits of each of these evaluation functions, a plan incorporating these efforts into the general product development process evolved. The plan is presented to emphasize the significance of product development evaluation and is not intended to offer a comprehensive view of production and manufacturing, each complex processes within themselves. An overview of the plan includes the following stages, which correspond to the accompanying flow diagram.

Stage 1. Development of User-Ready Product

An equipment developer has already discovered that a need exists for a specific device. In response to that need, the appropriate resources were tapped which resulted in the production of a functional prototype unit or working model. At this critical point in the development process, the developer begins to contemplate and, perhaps, question the safety and effectiveness of the device, its applications, and user acceptance. Many devices may not progress further. At the same time, many developers are interested in

continued advancement of their products or at least in exploring that potential.

Stage 2. User Field Trial Evaluation

An efficient method for quickly determining a product's general potential and gaining objective feedback is an informal user field trial. Although the number of available (prototype) devices must be considered, this trial can involve as many users as desired, but generally a minimum of two participants is necessary. These users must be appropriately matched to the particular product, instructed on its use, and be of a device-oriented nature to insure that the device will, in fact, be evaluated and adequate feedback offered. During a two-week trial period, usually an adequate interval for most items, each user is instructed to note their experiences with the equipment. If possible, the user/evaluator is encouraged to incorporate experiences and comments into a written summary, which is especially beneficial information for the product developer. A post-trial interview with each user/evaluator, usually only an hour in duration, is conducted by the field trial manager. Customarily, user/evaluators are considered consultants and are paid a reasonable fee for their services. The field trial offers a timely response to the developer concerning the product's effectiveness, its applications, user acceptance, and any additional predetermined considerations. Most importantly, the information has come from an objective viewpoint.

Stage 3. Device Pre-Production Design and Tooling

Pending positive results from the user field trial, the decision to proceed with production and manufacturing of the device has been made. Any information that has resulted from the field trial has been incorporated into equipment design considerations. The manufacturing elements are activated to begin toolmaking for production.

Stage 4. Device Commercialization and Marketing

Assuming that the pre-production stage has resulted in the manufacture of the device, commercialization advancement has continued. Efforts are now focused on product marketing. Strategies and tools that promote the visibility and sales of the product are employed. These efforts are primarily directed toward equipment prescribers (medical and clinical professionals, vocational rehabilitation personnel), third party payers, and, in some instances, disabled consumers. When considering equipment purchase, most prescribers and payers tend to rely heavily on documentation that indicates demonstrated success and substantiates product safety, effectiveness, and reliability. The basis for the development of this documentation stems from formal clinical evaluation results. Objective clinical data that support claims made by the developer/manufacturer may weigh directly (and heavily) on the purchase decision.

Stage 5. Formal Clinical Evaluation

Concurrent with the pre-production and manufacturing phases, formal clinical evaluation of the device should be formulated and begun. Formal evaluation represents a massive coordination effort, including enlistment of cooperating clinical facilities and staffs and development of evaluation program design and protocol with contributions by the device developer, engineering staff, clinical staff, and statistics personnel. These activities are followed by seeking approvals from Institutional Review Boards, procurement of evaluation items, patient screening, selection, and scheduling. All is accomplished prior to commencement of the evaluation. Information previously obtained during the field trials is especially useful for the preliminary development of evaluation protocol and patient questionnaires. Though the field trial generally provides a fair prediction of formal evaluation results, a larger user population operating under a prescribed protocol is prerequisite to the statistically significant substantiations of device safety and effectiveness.

Stage 6. Data/Results Collection

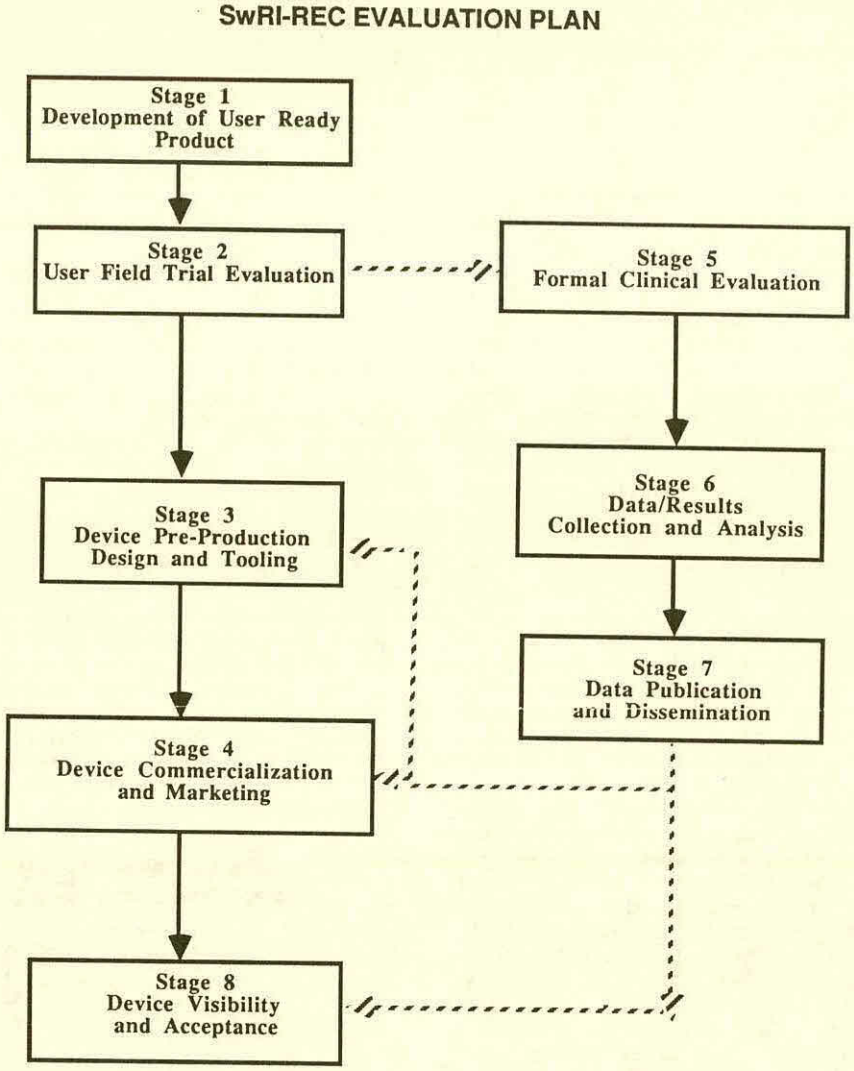
Through the course of the clinical evaluation, data is collected by the patient/user and by members of the evaluation team according to the protocol. Following the prescribed evaluation period, an in-depth interview is conducted by the clinical team with each user. All results are compiled, discussed, and reviewed by the evaluation team. With the assistance of the statistician, the results are analyzed for significance.

Stage 7. Data Publication and Dissemination

The evaluation method and information resulting from the clinical evaluation activities serve as the basis for the development of clinical journal articles and professional papers submitted for publication. The information also provides the basis for claims made in commercial sales and promotional literature.

Stage 8. Device Visibility and Acceptance

With the support of the clinical evaluation data and published reports and literature, the product stands to potentially gain visibility and acceptance in the market among users and prescribers. Clinical evaluation effort have helped to insure product success through demonstration and substantiation of safety, effectiveness, and user acceptance.



CONCLUSION

Other programs and organizations have developed and successfully demonstrated the validity of particular models for rehabilitation product evaluation. However, the SwRI REC will continue to formulate evaluation plans and models and refine them through demonstration in order to respond to the differing and changing needs identified among the members of the rehabilitation community. Clearly, the SwRI program cannot make gains without the participation and cooperation of other REC programs, rehabilitation product developers and manufacturers, evaluation facilities and staffs, and particular government agencies. Comments and recommendations are invited.

ACKNOWLEDGEMENTS

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OPPORTUNITIES FOR NASA TECHNOLOGY TRANSFER IN REHABILITATION

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ABSTRACT

The Technology Applications Team (TATeam) at Research Triangle Institute, North Carolina, serves as the technology transfer agent in the dissemination and utilization of NASA's aerospace technology to problems in medicine and rehabilitation. NASA has formalized this active transfer process in its Technology Utilization (TU) Program. Here, medical and rehabilitation specialists, medical device manufacturers, and Federal health agencies, together with the TATeam and NASA scientists and engineers, follow a methodology that promotes the transfer of NASA technologies into new, commercially available products or processes. An example of technology transfer in rehabilitation is a wandering notification device for memory-impaired older persons.

INTRODUCTION

The term "technology transfer" has recently gained popularity, or been coined as a "buzz word", to define the process of applying existing technologies into areas or problems that are different from the original use of that technology. Another definition, loosely based on terminology used in the Second College Edition of the American Heritage Dictionary, is that technology transfer is the shift of a body of knowledge from one place to another. However, both of these definitions lack an essential element of the technology transfer process, and that is; technology transfer is not complete until the technology or knowledge has been successfully used in its new application, either as a new commercial product or process. This paper will present a well-defined methodology for transferring technology from the National Aeronautics and Space Administration (NASA) to applications in rehabilitation and rehabilitation engineering.

BACKGROUND

As part of the Space Act of 1958, Congress incorporated into NASA's charter "...the widest practical and appropriate dissemination of information concerning its activity and the results thereof." In 1962, the NASA Technology Utilization (TU) Program was organized to uphold this obligation. Today, the NASA TU Program includes applications engineering, as well as information dissemination programs. NASA TU has grown into a nationwide network consisting of ten NASA Field Centers (located at the nation's primary space and flight centers, such as Kennedy Space Center, Johnson Space Center, and Ames Research Center), Industrial Applications Centers, and the Technology Applications Center at Research Triangle Institute (RTI).

The Technology Applications Team (TATeam) at RTI serves as technology transfer agents in the application of NASA technology to problems in rehabilitation and medicine. The TATeam, whose members include biomedical, rehabilitation, and industrial engineers; life and social scientists; and computer specialists, serves as active transfer agents, based on the acceptance that passive information dissemination is not always sufficient for successful technology transfer. This group of professionals, together with consultants from industry, medicine, Federal health agencies and associations, and research and academic institutions, employs a technology transfer methodology that is described below (Figure 1).

METHODOLOGY

Problem Identification and Needs Assessment

The Technology Applications Team collaborates with outside sources to identify problem areas in rehabilitation (for all populations of handicapped individuals, including physically and developmentally disabled and the frail elderly) which may be amenable through the application of technology, and resulting in a new device or process. Groups assisting in problem identification include Federal agencies, rehabilitation research and engineering centers, clinicians, and consumer organizations.

NASA Technology Utilization Project Development and Decision Points

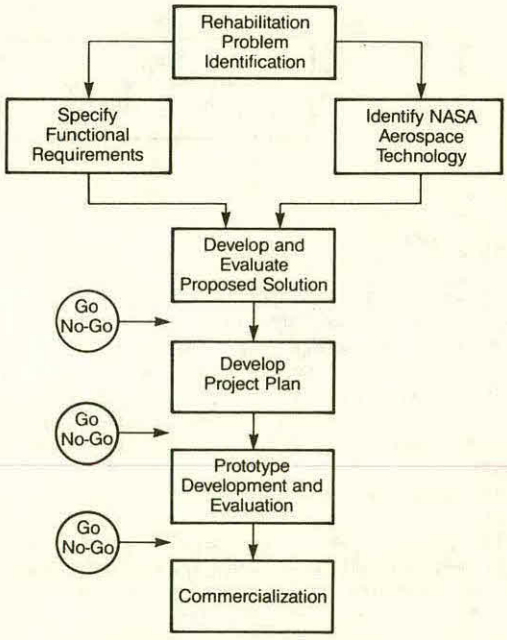

Figure 1.

After a problem area has been identified, the TATeam conducts a needs assessment study to validate the impact of the problem and the availability of potential solutions. This will include an extensive literature review, as well as consultation with experts in the field, service providers, and potential end-users of the new product. The TATeam will also identify any commercially available devices designed to address the problem. At this early stage, it is also important to both assess and promote manufacturer interest in the new product including any perceived barriers to commercialization.

Thorough preliminary evaluation is perhaps the most important step in the technology transfer process. At this point, the goals are to (1) identify a significant problem area, (2) precisely define the target population, and (3) select a project which has a demonstrated need, in other words, one which has the greatest potential for commercial success. (Remember, our definition of technology transfer states that the technology must be successfully used in its new application.)

Specify Functional Requirements

After selecting the project and specifying the characteristics and needs of the target population, the TATeam works with the problem originator (such as a rehabilitation engineering center) and other experts in the field to develop functional requirements for a proposed solution.

The TATeam may use experts in the field and manufacturers of existing products to help evaluate the functional requirements. Another approach that may be used is focus group discussions. Focus groups usually consist of 6 to 10 persons who, because of a common background or interest, will help a manufacturer or investigator evaluate a new idea or proposed features of a new product. The functional requirements will most likely be modified as a result of this review process.

Identify the Applicable Technology

The TATeam must identify and prioritize the technologies needed in the functional requirements. Then, in order to identify the applicable NASA technologies, the team prepares a concise statement of the problem, including constraints and specifications for the solution, for distribution to the ten NASA Field Centers. At each center, a Technology Utilization Officer distributes the problem description to appropriate personnel for review. As part of their role in the Technology Utilization Program and through their in-depth knowledge of specific technical areas, scientists and engineers at each center then respond with suggested solutions based on the most appropriate NASA technology.

The TATeam also identifies other relevant technologies (possibly available through commercial sources) that may be appropriate for the proposed solution. All the technologies needed to develop the new product or process, and identified through NASA or other sources, are evaluated for their relevance, availability, and cost-effectiveness as part of the final solution.

Develop and Evaluate Solutions

The TATeam, problem originator, and NASA scientists and engineers develop a proposed solution based on the functional requirements and incorporation of the identified NASA technologies. At this point, the proposed solution once again undergoes a critical review, this time to assist NASA in deciding on whether to continue the technology transfer project.

Parties who help evaluate the recommended solution include experts in the field who can help determine the solution's feasibility, potential impact on the problem, and cost-effectiveness; Federal health agencies who can assess the development and evaluation costs; and regulatory agencies who may be aware of commercialization obstacles. The TATeam also reviews the proposed solution with industry, especially those manufacturers interested in undertaking the development of the product. Here, it is important for manufacturers to evaluate the features, the technical feasibility of the solution, the potential development costs, and critical marketing considerations for the new product. It is unlikely that a technology transfer project will continue past this point if manufacturers feel that the size of the market or technical feasibility of the solution do not justify the development and marketing costs.

Develop Project Plan

The TATeam, together with the NASA Technology Utilization Officer (usually representing the Field Center associated with the identified NASA technology), NASA scientists and engineers, and any participating sponsors (such as a rehabilitation research or engineering center), develop a plan and cost estimate for the commercial development and evaluation of the potential new product. Relevant issues addressed in the project plan include costs for engineering design, hardware development, and clinical evaluation stages; possible sources and methods to secure co-funding; time constraints on modifying the applicable technologies; and any patent or license requirements that must be considered.

In addition, the plan must present a means to secure a collaborating manufacturer. This may be through a competitive procurement procedure, including issuing a formal Request for Proposals (RFP). The project plan should present clear goals for the manufacturer, as well as a manageable time frame for completion, including project milestones together with identifiable goals for each stage of development.

Monitor Prototype Development, Evaluation and Commercialization

The selection of a collaborating manufacturer is based on the demonstration of their ability to successfully complete the project. Among the factors considered are technical merit of their proposal, including demonstration of an understanding of the problem and project objectives, together with a knowledge of the required technologies; organization capabilities, including personnel qualifications, manufacturing resources, and previous success with related product's; management plan; co-funding commitment;

cost estimates for product development; and marketing strategies.

At this stage, the Technology Applications Team assumes a less visible role in the new product's development. The selected manufacturer together with appropriate NASA personnel undertake the prototype development, evaluation, and commercialization of the new product. The TATeam serves to monitor the project and act as liaison between NASA and any co-funding or co-sponsoring organizations, as necessary.

CURRENT TECHNOLOGY TRANSFER PROJECTS IN REHABILITATION

A Wandering Notification Device for Memory-Impaired Older Persons

Memory impairment is a significant problem for both independent and institutionalized older persons. While memory impairment can manifest itself in many ways, one symptom commonly found in the demented elderly (those persons suffering from Alzheimer's Disease or other dementing illnesses) is wandering behavior. Wandering in itself is not harmful; it may help relieve tension, or promote social interaction. However, wandering can often lead to consequences which may physically endanger the wanderer, or force family caregivers to institutionalize an older family member.

The TATeam, working with a consortium of five Federal agencies and experts in gerontology and memory impairment, has developed functional requirements for a system to detect wandering and notify both the wanderer and caregiver if a potentially harmful situation exists. These requirements incorporate some capabilities of commercially available wandering notification systems, while extending the features to include memory prompting capabilities for the wanderer, and also addressing wandering in both home and institutional settings. The functional requirements have been evaluated through focus group discussions with family and professional caregivers, experts in the field, and manufacturers of existing systems.

The TATeam has identified applicable NASA technologies in two-way communications, signal locating and signal tracking, and voice cueing. Expertise in these areas is being contributed by personnel at the Johnson Space Center and at Ames Research Center. At present, the TATeam is formalizing the proposed solution and preparing a project plan to select a collaborating manufacturer. If the sponsoring agencies decide to proceed to the next phase of the project, competitive procurement and an award to a manufacturer will be made by Fall, 1986, and an engineering design leading to prototype development will be completed by Spring, 1987.

CONCLUSION

A methodology for the active transfer of NASA aerospace technology to problems in rehabilitation has embodied the definition that the technology must be used in a new commercial product or process. It is hoped that an ever increasing technology base will be utilized to help overcome barriers to independent living for disabled persons.

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ABSTRACT

Developing a method for accessing an augmentative communication system for a physically disabled client is an integration of many components. This paper will review those components and their implications in selecting a method of access. Although emphasis is placed on assessing a client's physical ability, cognitive and psychosocial factors are also considered. A single case example is used to illustrate the evaluative process and development of an access system.

It is the intent of this paper to provide a knowledge base for those involved in the selection of augmentative communication systems and stimulate interest in areas where further research and development are needed.

INTRODUCTION

For the professional charged with developing an augmentative communication system for a physically involved client the choices are numerous. When the decision is made to utilize an electronic system the professional must choose an appropriate selection scheme, symbol system and access method.

Determining the selection scheme and symbol system appropriate for a particular client will be based primarily on cognitive functioning. While selecting a method of access will be influenced primarily by the person's physical ability.

At times choosing the appropriate method of access becomes a guessing game. However, there are some basic principles that can help alleviate the trial and error component.

MATERIALS AND METHODS

Hierarchy of Muscle Group Utilization
Silverman, in his book Communication for the Speechless (2), outlines a hierarchy of muscle groups to utilize when selecting an access system.

Upper extremity. The first group of muscles considered are those of the upper extremity. Use of the upper extremities allows two means of access; direct or indirect. Direct access is achieved through the use of a keyboard as in a typewriter. The VOIS 130 by Phonic Ear is another example of a direct access device. Keyguards and expanded keyboards can be used to facilitate direct access where minimal impairment of motor function exists.

Indirect access to the communication device is achieved through the use of switches. Switches frequently used with the upper extrmities include arm slot, tredle, and leaf switches.

Neck or face. The next group of muscles in the hierarchy are the muscles of the neck and face. Access may be either direct or indirect. Use of a head pointer may allow the person to directly access a keyboard. Indirect access can be achieved through the use of eyebrow wrinkle, pneumatic or specially mounted tredle or leaf switches.

Lower extremities and trunk. The muscles of the lower extremity and trunk are the last groups of muscles considered. Direct access using these muscle groups is limited to those individuals who display significant dexterity of the lower extremity and foot. Indirect access can be achieved through use of a slide switch, tredle or leaf switches.

Active Range of Motion
Active range of motion refers to the range of movement a person has at a given joint. Following the hierarchy of muscle group utilization active range of the upper extremities is assessed first. This information will assist in determining if a direct access method can be used. If an indirect system is used the information assists in determining switch placement.

Coordination of Movement
While considering range of movement the coordination of that movement must also be a concern. If a direct access method is being considered an assessment of coordination assists in determining if a keyguard or expanded keyboard is needed. In considering an indirect access system, assessing coordination assists in selecting an appropriate switch. The following components of coordination must be considered.

Accuracy. The accuracy of a person's motoric response will affect the size and placement of the system.

Speed. The time required for a person to initiate a motoric response will need to

be considered. This information will be pertinent when considering a selection scheme.

Fatigue. Fatigue will affect the speed and accuracy of a motoric response. How quickly the client fatigues then is an important consideration. Where more than one accessing method is being considered fatigue may be the determining variable.

Muscle Tone and Reflexes
Because abnormal muscle tone and primitive reflexes are frequently present in client's with a neurological insult they must be considered here.

Primitive reflexes. Clinically there are two primitive reflexes that have the greatest influence on choosing a method of access. The assymetrical tonic neck reflex (ATNR) and the symmetrical tonic neck reflex (STNR) are postural reflexes which effect muscle tone throughout the body in response to changes in the position of the head and body (1).

When present these reflexes will influnce the client's ability to maintain eye contact with the communication device as well as his ability to access the device through direct or indirect methods.

Muscle tone. Muscle tone can be described as the amount of tension present in a muscle. Muscles at rest maintain a certain amount of tone. In the neurologically impiared individual this tone may be affected. Hypertonicity refers to the presence of increased muscle tone. Hypotonicity refers to decreased muscle tone. The presence of either of these conditions will affect the person's active range of motion, speed and accuracy of motoric response.

Positioning
The position the client will be in while using the communication device is our next consideration. If the client is ambulatory, or will be using the device in different positions, such as in a wheelchair as well as in bed the device may be designed to be portable and the accessing system must be protable as well.

The effect of motor output on the client's overall positioning must be observed. If the motoric requirement for accessing the device is too great there may be a detrimental affect on the client's ability to maintain a functional position, thus affecting his ablility to access the device.

The effect of gravity on the movement required to access the device will most likely be seen when assessing coordination of movement. At times it is possible to negate the effect of gravity by placement of the switch.

Placement of the Access System
Once a method of accessing the communication device has been selected, the placement of that system, as well as placement of the communication device itself in relation to the client is assessed.

Field of vision. While the communication device itself will be in the client's field of vision, thought should be given as to whether or not the access system is also within his field of vision.

In the clinic, it has been observed that some clients will attend more to the switching device than the communication device. This may be attributed to several factors. First, the motoric output required to make a response is too great. Second, the client may have poorly functioning kinesthetic and proprioceptive systems and needs the extra input of visual attending. Third, it is an unlearned motoric response and visual attending to the access system will decrease through practice, much as one learns the keys of a typewriter or a piano.

Whether or not to place the accessing device in the client's field of vision is an area devoid of research and full of conjecture.

Practical Application-A Case Study
T. is a ten year old male, in residence at a state school for the mentally retarded. His diagnosis includes mental retardation and cerebral palsy. He is unable to sit independently and exhibits poor head and trunk control. His movement patterns of his head, arms and legs are influenced by a strong ATNR. His seating system does not provide good inhibitatory positioning, a tray is not used on his wheelchair.

Using the hierarchy of muscle group utilization as a reference T. was evaluated for an indirect accessing system. Active range of motion of his upper extremities was found to be limited and greatly impeded by the force of gravity and the influence of the ATNR. In sidelying he presents increased use of his upper extremities. However an accessing device that would be useable in the sitting position was needed.

Consideration was then given to the neck and face. Again the influence of the ATNR was of primary concern. Any switching system that required turning of his head would have a negative impact on his overall body position. Oral motor skills were also not developed sufficiently to use a pneumtic switch.

Finally the lower extremities were assessed. There still existed the

influence of the ATNR but it was less pronounced than in the upper extremities. The most consistent motor response that could be elicited was extension of his right knee. It was decided this motion would be used and the next step was initiated, finding the appropiate switch.

Commercially available tredle and leaf switches were considered. However an acceptable mounting system was beyond design. Attention was then turned to designing a homemade switch system utilizing a microswitch. The microswitch is enclosed in a wooden box which is attached to the base of his wheelchair. A small piece of wire leads from the microswitch to a velcro loop that goes around his right ankle. A fishing swivel and clasp were used so that the velcro loop can be detached allowing the box to remain permanently attached to his chair if desired. Training T. to use the switch was initiated using adapted battery operated toys. Proper placement of these training devices as well as his communication device was emphasized to minimize the influence of the ATNR on his ability to access the devices.

CONCLUSION

Assessing clients for an augmentative communication device is a complex task for anyone. At Pauls Valley State School we have found an interdisciplinary approach to be useful. Utilizing the expertise of a speech pathologist, occupational therapist, and rehabilitation engineer has led to the development of communication systems for multihandicapped individuals whose communication attempts in the past have been limited to the "twenty questions" system.

ACKNOWLEDGMENTS

The author wishes to acknowledge the administrative staff of Pauls Valley State School for their support in securing the staff and materials needed.

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FIVE NEW PIECES OF EQUIPMENT FOR QUADS

Arthur Heyer
EXTENSIONS FOR INDEPENDENCE

For most handicapped individuals, equipment is essential to be able to function productively, whether it is at school, at home or at work.

In the course of my life as a quadriplegic, I have undergone a large number of experiences involving the use of equipment and the lack of it. I know how it feels to want to do something, but not being able to; or how it feels to be able to do something which is generally not expected from an individual with my disability.

In the next paragraphs I will describe the few latest pieces of equipment which have added comfort and sense of freedom to my life.

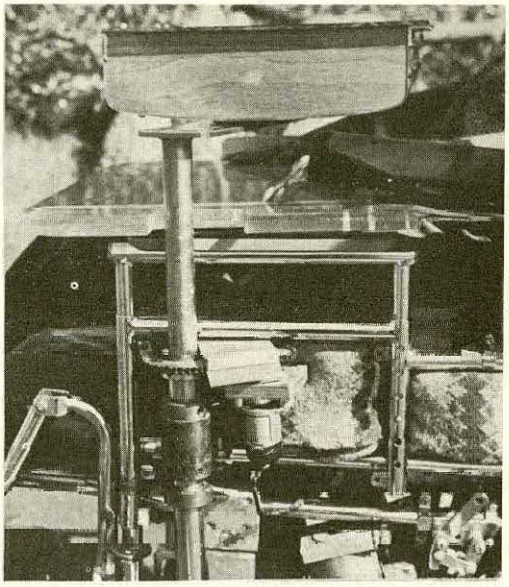
SWING-AWAY PORTABLE DESK

Next to my mouthstick which is my "hands", and my power wheelchair which is my "feet", a PORTABLE DESK I carry on my wheelchair everywhere I go has been a most useful piece of equipment.

This portable desk is designed to provide wheelchair users with the convenience of an organized mobile workstation. It is basically a box whose top slides or lifts open (both types available) and serves as a working surface. Things for use at school, the library, etc., such as books, a tape recorder, or a portable computer, notebooks, pencils and papers can be carried inside. Optional built-in ramps can be used by mouthstick or frail hand users to move things onto working surfaces at different levels, such as a main desk, or to move things from inside the box onto its top.

Until recently, this portable desk was an item that had to be custom made to suit individual needs. The way of mounting on the wheelchair was the main reason for the need to customize. For example: different types of disabilities required different ways of positioning the portable desk. Also usage needs varied: one individual required that he could move his portable desk away for driving his van. All Individuals with arm motion want the top of their portable desk to be under their arms and flush with their armrests. Those with no function of arms or hands benefit from having their Portable Desk over their arms. This way they make use of the greatest working area possible. Wheelchair size and type, as well as the user's size, were also factors that required precise custom making.

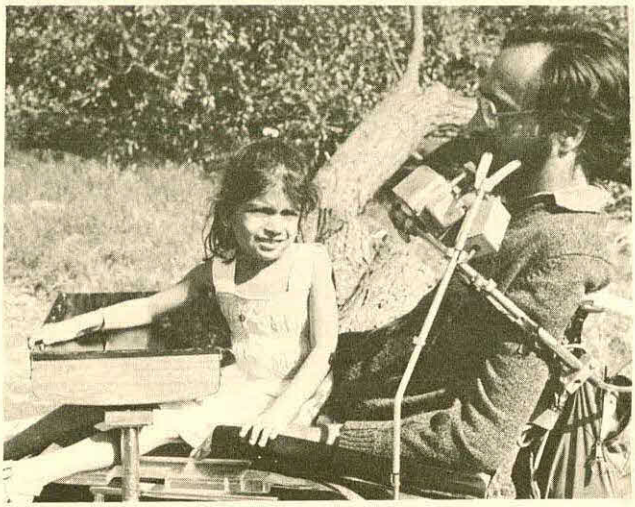
A Universal way to mount a Portable Desk on a wheelchair was finally achieved. It is a flat stainless steel arm welded to a telescoping post which mounts on one side of any wheelchair.



Mounted on this arm, the Portable Desk can be raised or lowered, placed over or under the arms, closer or further away to suit individual needs, and can also be swung away.

A motorized version gives freedom and fun to high-level quads. For example, leg positioning can be checked by swinging the PO away; or quads who drive can make room for the steering wheel, and still have their PO within their rich on their side.

I have had great fun carrying my little girls on my lapboard under my PO. They like to see how I work on my computer, and they are of great help by being my hands.



BLANKET REMOVER

Knowing how much to cover when going to bed was a problem for a long time until I had two pieces of equipment made and installed: a window opener and a blanket remover.

Years ago, I had the idea of a motorized device for removing or replacing a blanket. I let years pass without implementation, trying to solve my problem with available means.

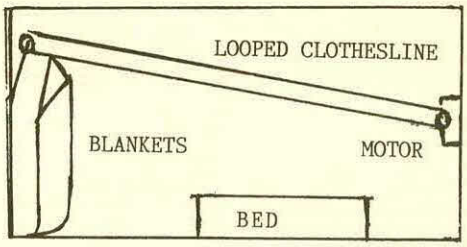
Automatic control of temperature at one single level does not work on a person who only feels temperature changes through blood circulation: I feel too cold or too worm when I am put in bed; one hour later my feeling of temperature changes and I need adjustments.

For all those years I had to call somebody during the night to come and make adjustments. The problem became psychosomatic, with fear at the time of going to bed that I was not going to cover enough or that I would cover too much.

A motorized window opener helped to diminish the problem. This way I had some control of the temperature in my room by opening or closing the window at will.

The approach works relatively well in winter. However, it is ineffective on warm days where the outside temperature is about the same as that inside. In any case, I still needed adjustments almost every night.

The problem was finally solved when I had installed an electro-mechanical device to remove or replace a blanket. The device is basically like a motorized looped clothesline extending a few feet above my bed lengthwise from wall to wall (see drawing below).



An aluminum tube with a length equal to the width of the bed was used in a way of a trapeze suspended on two cords. Two blankets were sewn around the aluminum tube.

Notice in the drawing that the bed is separated from the wall and also that one of the pulleys is substantially lower than the other. This arrangement allows for the blanket to be high against the wall when not in use and to cover more fully when being used.

STICK-ON KEY-LOCK

In a previous paper (Reference #8) I had described a keyboard adapter, which was an acrylic overlay for a computer keyboard, with spring-loaded keylocks.

The mechanism I am describing now fulfills the same purpose: to allow one-handed or mouthstick users to depress two or more keys of a keyboard simultaneously,...but in a much simpler way:

The new keylocks are not spring-loaded. They don't automatically move over the key when the key is depressed. Instead, the keylock, designed as a lever with two inclined planes at its tip, is moved, by hand or mouthstick, over the key in order to depress it. The key then remains depressed until the keylock is pushed back.

FIVE NEW PIECES OF EQUIPMENT

The new type of keylock comes with its own independent base or pad, with double-sticky tape on its bottom. It can be stock by the key it is to lock, directly on the keyboard, or on an overlay which some computers may require.

One keylock has been designed to fit most Apple computers, and some of the other brands as well. In general, a key to lock should have a flat surface next to it in order to accommodate the keylock pad. Fortunately, the keys that need to be locked are located at the edges of the keyboards on most computer brands, and there is generally enough room next to the these keys to accommodate the keylock.

For some computer models, one Key-Lock can operate two adjacent keys alternatively, such as "CONTROL" and "SHIFT" on the Apple IIc and others. This feature was introduced on our keylocks with a double bevel tip.

EASY-LOAD-A-DISKETTE SYSTEM

It is a set of simple and inexpensive aids which allows frail hand and mouthstick users to interchange floppy disks quickly and unassisted.

How it works:

One aluminum tray is attached to each disk drive. The tray is used for the user to lay a floppy disk in position for proper feeding into the disk drive.

Wire hangers, which attach to each floppy disk, allow the disabled user to transfer the disks from a storage case to the tray on the disk drive, or vise versa, by means of a specially designed hook.

Who can use the system?

Anybody who can turn pages of a book or type, by hand, by foot, or by mouthstick or headwand, should be capable of using the system.

CONCLUDING REMARKS

Being a consumer of my own creations has given me the unique oportunity to fully test a new device before marketing is considered. My choice to become an engineer was a good one; and I encourage others to persue their career desires in spite of their handicaps...especially if they have the oportunity of having their functional needs resolved (through equipment?).

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A HIGH SPEED INTRAORAL SWITCHING DEVICE FOR THE MOBILITY IMPAIRED

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ABSTRACT

This paper describes the development and operation of a jaw operated proximity switch which replaces a specific foot function for a paraplegic patient. The jaw operated switch discussed here activates a piano pedal depressing mechanism for a professional pianist who was injured in an automobile accident that resulted in paralysis from mid-chest down. The magnetic proximity switch is a departure from conventional mouth operated devices and may have wider applications for other patients affected with spinal cord injuries. Advantages of the magnetic proximity switch are 1) the jaw activated movement is non-fatiguing and unobtrusive; 2) it does not interfere with phonetics, talking or breathing; 3) the patient is able to maintain control with a high degree of accuracy and coordination; and 4) it is easily mastered in a short time.

INTRODUCTION

An estimated 8,500 persons[1] in the United States are affected yearly with spinal cord injuries and 1,400 of these injuries occur at the C1-C4 level[2]. Persons afflicted with quadriplegia, paraplegia, and multiple amputations, and a percentage of people afflicted with severe arthritis, cerebral palsy, muscular dystrophy and multiple sclerosis have extreme mobility limitation, but often retain excellent mouth/jaw dexterity and movement. These patients are able to interface with vocational and environmental equipment such as powered wheelchairs, communication devices, etc., through such devices as sip and puff pneumatic switches, chin and eyebrow single action switches and a variety of oral devices.

The variety of control mechanisms that interface with independent living and vocational equipment are selected on an individual basis with particular attention to:

1) the user's abilities
2) ease of learning and operation
3) effectiveness of control relative to user requirements
4) esthetic and cosmetic appearance
5) interference with other functions such as vision and movement
6) sanitation and contamination
7) medical considerations.

Even when the above criteria are carefully evaluated in the selection of a control mechanism, there is often a functional trade-off such as unnatural or fatiguing movement, poor appearance, or slow operation[3-6].

BACKGROUND

A female professional pianist and teacher was injured in an automobile accident seven years ago with resultant paralysis from mid-chest down. Although the patient was still able to play the notes on the piano, she was unable to use her legs and feet to depress the foot pedals, which not only enables the pianist to sustain the tone of those notes required by the music score, but also provides a resonance to the music. Several attempts had been made to attach jointed sticks from her waist to the sustain pedal, but without success. The piano and music were always an integral part of this patient's life and without the use of her legs the fullness and completeness of the music was not possible.

The Georgia Division of Rehabilitation Services referred the patient to the Georgia Institute of Technology's Center for Rehabilitation Technology. Their task was to develop a device that could mechanically mimic the human ankle in depressing the sustain pedal. The Georgia Tech engineers coupled an electro-mechanical drive system to a depressor lever that was instantaneously responsive to an operator's signal. It had sufficient torque to depress the piano pedal and to instantly respond to a pedal release signal.

The problem of activating the control switch emerged. The patient's hands were occupied in playing the notes. A sip and puff mechanism, often used as a control device for persons afflicted with paraplegia and quadraplegia was unacceptable because it interfered with breathing and it lacked the speed and precision required for intricate passages of music. Some musical compositions require up to three sustain strokes per second.

In attempting to resolve this problem, the patient was referred to Emory University School of Dentistry for evaluation. The patient's oral condition and masticatory capabilities were evaluated and found to be within normal limits. Requirements for activating the depressor mechanism and the patient controlling device were then analyzed. It was determined that jaw movements, particularly short protrusive movements, could best be utilized. Protrusive movements can be very precise and rapid and appeared to best meet the demands of operating the pedal depressor mechanism.

MATERIALS AND METHODS

A Hall-effect proximity switch to be imbedded in an acrylic resin splint was selected as the patient's controlling device. This switch is activated by a magnetic field exciting a solid state circuit and requires no actual contact of

metal bodies. It is activated by having one side exposed to the north magnetic field of a magnet. There are no moving parts and a single three-lead wire is the only connection required. A one-eighth by one-eighth inch cylindrical rare earth Cobalt magnet to be embedded in an acrylic resin splint was selected as the activating component because it has an extremely powerful magnetic field relative to its size. Both major components are of sufficiently small size so as to be easily incorporated in the protheses.

The Hall-effect switch was selected to allow the use of an opto-isolator unit which provided a maximum degree of safety. The only connection between the mechanism and the switch is a light path--thus isolating the patient from the high powered electro-mechanics. The opto-isolator actuates an electronic motor control circuit with optical position sensing feedback which in turn controls an induction motor that powers the depressor lever.

The switch circuit is powered by a 9-volt battery pack that attaches to the back of the patient's wheelchair.

Impressions were made of the patient's maxillary and mandibular arches, casts poured, undercuts blocked out, and duplicate casts fabricated. Jaw relation records were secured at sufficient vertical dimension to accommodate the splints and the modified casts were mounted on a Hanau Articulator. It was critical to position the switch and magnet in the splints with the appropriate distance between them. This precise distance was critical to ensure that the switch would not be in the magnetic field and inadvertently activated when the patient closed in her habitual closure position. A small battery powered test light was connected to the components and used to position the switch and magnet so that the boundary of the magnetic field was 1.5mm to 2mm anterior to the Hall-effect switch when the patient's jaw was in habitual closure position.

The splints were processed with clear acrylic resin, returned to the articulator and the location of the magnet and switch verified with the test light. The splints were then broken out, polished, fitted to the patient's mouth, and adjusted for ease and comfort of operation. The switch and magnet were placed in their respective slots in the splints, again verified with the test light for placement, and incorporated into the prostheses with clear autopolymerizing acrylic resin.

RESULTS

The pedal depressor mechanism lever was set up and adjusted to a piano. The patient attached her mouth switch wire to the switching circuit and moved into place at the piano. She struck a chord, made an almost imperceptible jaw movement, the sustain pedal went down instantaneously; she raised her fingers from the keyboard and the chord was sustained. She made another slight jaw movement, the pedal raised and the sound of the chord stopped. One hour later the patient was still playing, using the jaw switch without muscle tenderness or fatigue. Her jaw movements and their relation to note sustainment, though requiring some degree of concentration, had become relatively automatic. Six months after delivery, the patient has had no problems related to the use of the jaw operated proximity switch and plays complicated pieces of music successfully and completely.

CONCLUSIONS

Several important facts were learned during the course of this project. Some are:

1) Mouth and jaw dexterity are retained in most high level spinal cord injuries.
2) A back and forth sliding jaw movement is a natural and non-fatiguing motion. This motion can be controlled with high accuracy and coordination.
3) Dental techniques permit the implant of miniature switches in partial plates that do not interfere with talking, eating, drinking, or head movement.
4) It is extremely fast and easy to learn how to operate a jaw activated switch.

These criteria have generated a new development project now underway at Georgia Tech and Emory University for a passive, wireless switching and controlling interface for high level quadriplegics. Findings from this project will be made available at the Minneapolis conference.

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INCREASED INDEPENDENCE THROUGH COMPUTERIZED VOICE RECOGNITION FOR PERSONS WHO ARE SEVERELY PHYSICALLY INVOLVED AND PROFOUNDLY MENTALLY RETARDED

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ABSTRACT

A person who is profoundly mentally retarded and severely physically handicapped learned to voice activate a computer system to gain control over various electrical appliances in her environment. This was achieved by interfacing several off-the-shelf computer items and using newly-devised training strategies with the subject. Data indicated the subject benefited from using this computer system which allowed her freedom of choice and independence for the first time in her life. This research documents the potential accomplishments of persons who are not only severely physically handicapped, but also profoundly mentally retarded if appropriate technology and training techniques are applied to their needs.

INTRODUCTION

There are more than two million individuals in the United States whose physical and/or mental impairments are so severe that they are unable to interact with the environment in a normal and effective manner. Persons with profound mental retardation and severe physical impairments are among the most difficult to serve. They are typically bed- or wheelchair-bound with very limited control over even gross motor movements and are often capable of making only guttural or unintelligible sounds. They are usually totally dependent on others to discern their basic needs and desires and to act on them. Often these individuals are denied by their handicaps and society's response (or lack of response) to them, the social interaction, opportunities for productivity, and personal fulfillment to which everyone is entitled.

BACKGROUND

A partial solution to this devastating situation was offered by research recently completed in the Bioengineering Program at the National Headquarters of the Association for Retarded Citizens. A major goal of the Bioengineering Program is to develop and apply technological aids to assist individuals who are severely or profoundly mentally retarded in achieving increased autonomy and an improved quality of life. Appropriate technology in many cases can compensate for their limitations, extend their abilities, and provide them with more normalized experiences. The general purpose of this specific study was to determine if a computer system equipped with voice recognition and environmental control capabilities was an appropriate tool for use by these individuals. While environmental control systems have been used by persons with severe physical handicaps such as quadriplegia, it had never been determined whether someone with profound mental retardation and severe physical impairments could learn to purposively use such technology or if the benefits of such use would be substantial.

The specific questions to be addressed by this study were whether a person who was profoundly mentally retarded, severely physically impaired, with minimal non-speech vocalizations could: 1) understand and relate to the concept of "control", 2) emit designated vocalizations with enough consistency to voice-activate and control a computer; 3) understand the relationship between emitting a vocalization and the resultant activation of an electrical appliance; 4) associate specific vocalizations with the activation of specific appliances; 5) display preferences among the available appliances; and 6) learn to de-activate an appliance once it was activated.

Thus, the basic intent was to configure an assistive device to intervene for the subject at her choosing to provide some freedom of choice over significant aspects of her environment -- and thereby extend her range of personal independence.

MATERIALS AND METHODS

The subject selected for this study was a 42 old resident of a large institution for persons with mental retardation. She possessed no self-help skills, had almost completely unintelligible vocalizations, was evaluated as profoundly mentally retarded with an IQ of 19, was confined to a bed or gurney chair all of her waking hours, and was dependent on others for the fulfillment of all her needs. The electrical appliances under computer control were chosen to match the desires of this particular subject based on feedback from parents, caregivers, and professionals through an interdisciplinary team process.

For the computer system to operate correctly, the subject had to be able to vocalize sounds consistently. The target sounds were selected after extensive evaluation of the subject's speech with assistance from a speech pathologist.

Four vocalizations were designated to cause the computer system to activate different appliances. A sound approximating the word four operated a videotape of classical stories of short duration. A sound approximating the word move activated a vibration massage pad. A sound approximating the word ray turned on a radio tuned to a country music station. A sound approximating the word bee operated a videotape of the subject's family who resided a great distance from the subject. A fifth vocalization, the word off, turned off any appliance that was operating. Each appliance was programmed to turn off after five minutes if the subject did not de-activate it. This was to prevent endless operation of an appliance if the subject did not understand the de-activation process.

Caregivers who assisted with the data collection were trained in the client training procedures and in the operation of the computer system while the subject was being familiarized with it. The entire computer system was housed in a metal cabinet in a large activities room. The monitor was mounted on a swivel metal bracket on the side of the cabinet for easy viewing by the subject.

A single-subject multiple-baseline-across-behaviors research design was employed to evaluate the subject's interaction with the computer system. The software for the computer was designed to record the vocalization attempts by the subject, the appliances that were activated, and the length of time the subject chose to leave them in operation. To permit clear interpretation of any results, not more than one appliance could be in operation at one time.

The study was designed so that the availability of each of the four appliances to be activated and de-activated was staggered across time. First, one appliance was available to be voice activated; after recording a stable rate of vocalizing its corresponding word, a second appliance was also made available to be activated; and so on until all four appliances were simulataneously available. Typically, there were two sessions of approximately 30 minutes duration per day, for a total of 70 sessions.

Initially, baseline data were used to determine when the vocalizations achieved a stable rate before they were linked to any appliance. Stability was achieved by Session 7 so baseline was terminated on the first targeted vocalization and the staggered availability of the appliances was begun. The videotapes of the classical stories could be activated by the subject vocalizing the word four and de-activated by vocalizing the word off. By Session 25, data indicated the second intervention should be introduced. She could now activate both the classical story videotapes by speaking four and the vibration massage pad by speaking the word move, and turn either appliance off with the word off. This continued until Session 36 when activation of the radio was made contingent upon the word ray. The same procedure was implemented in Session 53 with the videotapes of the family and the word bee. The subject was now able to activate the classical stories videotape, the vibration massage pad, the radio, and the family videotape.

Some of the technical complexities of this computer system were a function of the components required for its operation: a Scott Instruments Shadow/VET interface card and the Shadow Vet System operating system to process voice input from a microphone, a Bell & Howell interface card providing computer control over the functions of a videocassette recorder, a Bi-Comm Systems PC-1 powerline controller interface card permitting computer control over AC appliances and real-time clock interrupts, and a disk drive controller card and disk operating system. Each of these components utilized their own proprietary machine-code driver routines which needed to be integrated through a master routine. These drivers, along with the application software and record keeping routines that were developed, were required to be functional within the 64K memory environment of the 8-bit central processor of an Apple II Plus computer. The Shadow/VET voice training and Shadow/VET KEYVET software, which allowed for transparent keyboard operation, were also patched and modified by the Bioengineering programmer using the standard interface outlined in the Shadow/VET manual.

RESULTS

The subject learned the relationship between her vocalizations and the activation of environmental appliances and exercised demonstrable control over her environment for the first time in her life. Data indicated that when an appliance became contingent on a vocalization, the rate that a vocalization was spoken dramatically increased over the baseline rate. The rate that off was vocalized also increased in relation to the number of appliance activations, indicating the subject realized that she had control to turn the different appliances off.

The rate that the subject vocalized each word remained fairly stable once the corresponding appliance was available. The exception to this was the vocalization four which declined once the second

appliance had been available for some time and remained low for the remainder of the study. An explanation for this is that the subject became bored with the repeated viewing of the same classical stories. Data indicated the subject had distinct preferences for the different appliances in the following descending order: videotape of the family, vibration massage pad, radio, and videotape of the classical stories.

Results show that the subject learned how to operate a computer system by voice, responded differentially when different contingencies were in effect, and learned cause-and-effect relationships. The subject possessed definite preferences for the different appliances and liberally exercised the option to select, not select, or de-select the aspects of the environment over which she could exert control. She demonstrated her understanding of, and desire for, control of her environment. She also appeared to turn some appliances on and off for the sheer pleasure of controlling them.

The subject demonstrated an increase in, and maintenance of, positive affective behavior when operating the computer system. Videotapes taken during data collection captured the subject laughing with delight when she realized her impact on her surroundings by operating the computer system. Through the course of this research, caretakers reported the subject made gains in her social interaction. In addition to the computer system enhancing her skills, it is possible that the increase in opportunities for the subject to interact with the staff and the investigator as a result of the research contributed to this gain. It was observed that considerable positive attention was given to her by others when the sessions were not being conducted as a result of her participation in this computer application.

Important findings were obtained on the limitations of the technology used in this application. For research purposes, a human observer recorded and categorized each of the subject's vocalizations and the system's response to them. For each vocalization, the system could (a) ignore it, (b) respond with the "wrong" (unintended) appliance activation, or (c) activate the correct appliance. At the beginning of the research, the system responded correctly on approximately 49% of the subject's vocalizations. By the termination of the research, and with continued experimentation with the Shadow/VET, the system responded correctly on approximately 58% of the subject's vocalizations. It was discovered that the success of the subject's voice training was a very delicate process. Because of the subject's articulation and voice volume limitations, successful training of the system for accurate recognition required that each vocalization be trained with a different number of training counts than the other vocalizations due to the inter-relationship of all counts for all words.

The system also crashed on approximately 14% of the sessions, with system refinement by the Bioengineering programmer and Scott Instruments engineer in an ongoing process. The main sources of the system's problems were the operating system of the computer, the limited memory addressing capability of the 8-bit processor, memory or processor contention with the Shadow/VET, hardware reliability of the Shadow/Cache interface card, electrical noise originating from some of the electrical appliances, and bugs in the newly designed software. It is a strong testimony to the robustness of the subject and the power of the application that significant learning took place in spite of these problems.

CONCLUSIONS

This research demonstrates that of the thousands of persons with the most profound mental retardation and physical handicaps, who typically receive the most minimal of services and are the last to be considered for more normalized living routines, can begin to exercise the basic fundamental rights of freedom of choice and expression through advanced technology. Hidden capabilities can be unmasked and new skills developed. It can also offer to parents, teachers, and therapists optimism that more normalized and rewarding lifestyles are indeed possible for persons with severe handicaps. With the speed with which technology is advancing and costs declining, further research on such technological applications can continue this critical process of extending their capabilities and normalizing their lives.

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7.5

DEVELOPMENT OF A SPEECH RECOGNITION COMMAND SET FOR WHEELCHAIR CONTROL

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ABSTRACT:

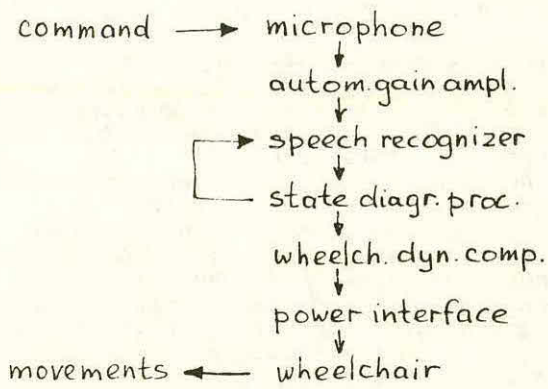
This paper presents an example of the application of methods to study the human factors aspects of speechrecognition. The application deals with the voice control of a wheelchair. Two different experiments were carried out to study the command language (the vocabulary) for wheelchair control.
The first experiment was aiming at the development of a natural command language. The results of this experiment show that it is rather difficult for potential users to define such a commandset.
A second experiment dealt with the confusion between different commands in a command vocabulary and errors invoked by words which were not included in the command set. The results of this experiments show a clear trade of between errors and correct recognized words due environmental noise and the setting of the recognition threshold.

INTRODUCTION:

Although the use of speech recognition by disabled people is not new, it seems that succesful applications are inhibited by several factors. One of these factors is the problem of designing a command set that meets a number of requirements. This paper will deal with the description of the requirements and methods which can be used to design a proper command set. These general applicable ideas and methods are put into practical use for wheelchair control. One of the first examples of a voice controlled wheelchair was designed by Youdin of the New York University Rehabilitation Engineering Center. This prototype was rather voluminous and expensive because of the state of the art at that time of speech recognition hardware and microcomputing products. The product never came into production and the prototype and its designer retired. The design mainly dealt with the speech recognition problem itself and little effort was put into the human factors aspects. A second example comes from France and Germany. The Kempf company sells a voice controlled wheelchair which is based on a specially designed speech recognition unit and microelectronics system. Although this design of Martine Kempf is put to a commercial available product, there seem to be problems with the marketing of this wheelchair. The French government obviously refuses to assist this company in supporting the production and provision. It is quite possible that other institutes have also tried out speechrecognition to wheelchair control. It appears very difficult to get detailed information about these developments. However, it is shown that it is technically feasible to design a voice controlled wheelchair. It is not shown that the use of such a wheelchair gives major advances for larger groups of handicapped people compared with more common control systems. Therefore, we decided to study the human factors aspects of voice control of a wheelchair.

PROBLEM DESCRIPTION:

The voice control of a wheelchair can be described by the following diagram:



The microphone and amplifier were not of special interest in this study: good quality is required with respect to the noise-cancelling characteristics. The speechrecognizer was purchased from Vecsys, France. This device has good characteristics and the possibility to use a template which could be set in real time by the computer. This feature was used implement a state diagram which diminishes the number of commands that has to recognizable at the same time. Due to undesired characteristics of the wheelchair (acceleration, deceleration, unsufficient control of speed) it appeared to be necessary to add a compensation for these undesirable dynamics. This compensation is done by software control; via a trial and error procedure the accelleration was set by a gradually increasing motorsignal. In this way it was also possible to compensate for high zero velocity roll resistance just by giving a temporarily large motor current. The state diagram can be designed such that different control principles can be used:
a) **Steering wheel control:** commands are available that bring the wheelchair in a rotational movement. This movement has to be stopped by a second command.

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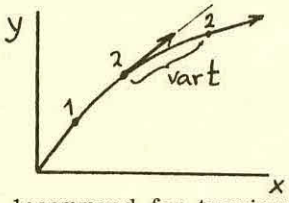
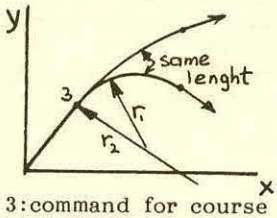
b) **Course control:** a command effects a rotation for a limited time, which gives the final result of a new course. In this case an argument(number) has to be added to the command in order to be able to select different courses.
c) **A combination** of a) and b) is possible.
d) Every **complex course** could be executed under software control. The most interesting one is to make two consecutive turns resulting in a lateral movement and continuing in the earlier direction. Again, there is an additional argument necessary to the command in order to be able to vary the lateral distance. The difference of steering wheel control and course control is that in the latter case less commands are needed, while the commands are more complex. In both cases there is a "noise" in the outcome of a command. This is shown in the next diagram:

a)steering wheel contr.



1:command for turning
2:command for straight ahead driving

b)course contr.



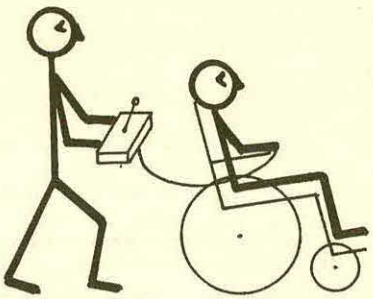
3:command for course deviation

Due to unprecise timing in case a) , the command at point 2 shows a time variance var t, which results in a variance of the resulting course. In the case of course control a command can be affected by noise in wheelchair turning control. This also gives the effect of a variance in the desired direction. Which of the variances is the largest one depends on the skillfulness of the wheelchair driver in case a), and the quality of the wheelchair motor controller in case b).

EXPERIMENTS:

Two different experiments were executed:
Experiment A was set up to discover a "natural" command language for wheelchair control,
Experiment B was the testing of a specific designed commandset on recognition quality. A third experiment is planned. In this experiment a comparison will be made of the different control strategies with respect to their efficiency. The result of this latter experiment will be published elsewhere.

Ad experiment A: The main purpose of this experiment is to discover a command language that could be characterized by "natural easy to use, easy to learn". A simple method was used in this experiment, which is applicable to all kind of other situations where speech recognition has to be implemented. The situation is given in the following figure:

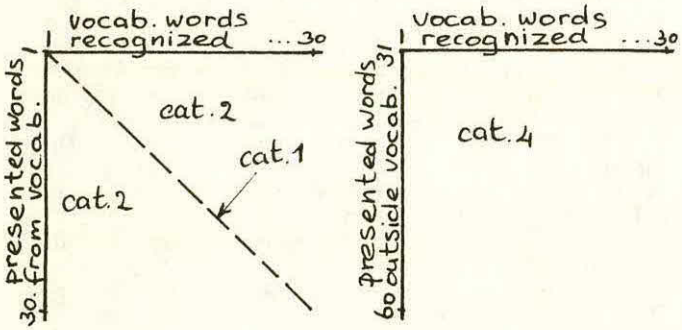


The speech recognizer was simulated by a normal person. This person had the task to interprete the commands of the potential user of a speech recognizing system. In our experiment this subject was sitting in a wheelchair. The subjects were all trained in normal wheelchair driving. A test route was laid out on the floor of a spacious room. This test route was designed such that normal practical situations were covered. The wheelchair subject was asked to drive three times through the testroute by giving spoken commands to the person walking behind the wheelchair. This person actuated a hardware simulated joy-stick. Before the testroute was run, the human speech recognizer was told by the wheelchair subject how to react on his commands, and which commands should be used. All utterances of the driving subject were recorded and analyzed. The result of this experiment can be summarized as follows:
1 It appears rather difficult for a person to find spontaneously command words. The most obvious command is just to ask somebody "to push the wheelchair to the kitchen or to that table". However, the human speechrecognizer is not allowed to use his vision or any other environmental feedback for doing so, because the fact that the technical speechrecognizer did not have either any environmental feedback. Thus the wheelchair driver had to find words which are related to directions, positions and velocity. The words which were finally used were close to the command words which were found by "desk" research.
2 Those command words which were related to corrections were used very frequently: "more right, more foreward, a little bit more, etc.". This could be explained by the fact that course control possibilities were rather poor in the experimental situation; it was not possible to give a specific command for turning a specific number of degrees, which could be excuted with sufficient accuracy.
3 A total of 43 command words were used in this experiment; many of them were combined words with adjectives as said above. The stop command was used very frequently: The wheelchair was stopped to enable the driver to do some mental planning.

Ad experiment B: The major problem in speechrecognition is discriminability of commands. Words spoken into the microphone can be categorized with respect to intended or actual effect as follows:

1 Correctly recognized
2 False recognized, another word is taken from the vocabulary
3 Not recognized while the word is in the vocabulary
4 A word is recognized while it is not in the vocabulary
5 Not recognized, and not in the vocabulary

The first and the last one are correct results. For this experiment a list was composed of command words(vocabulary) for wheelchair control, based on the outcome of experiment A (30 words). This list was extended with a list of similar words which should not be recognized (another 30 words). In the actual experiment, each word was entered in the microphone 5 times in a ramdom order. The words were presented to the speakers by a speech synthesis system. The results were plotted in a matrices(confusion matrix) which presents on the vertical axis the stimulus (the words generated by the speech production system) and on the horizontal axis the words which were recognized. All deviations of the diagonal represents an error, either of category 2 or 4 of the above given list:



This experiment was also repeated at different environmental noise levels. The following table gives the results(Steeneken,1984):

noise dB(A)	0	55	65	75	85
1 recogn.	86.0	88.7	83.3	87.3	82.7
in vocab	(94.7)	(94.7)	(94.0)	(96.7)	(96.0)
2 wrong	0.0	2.0	2.0	0.7	0.0
in vocab	(0.7)	(2.0)	(2.0)	(0.7)	(0.0)
3 not	14.0	9.3	14.7	12.0	17.3
recogn.	(4.7)	(3.3)	(4.0)	(2.7)	(4.0)
4 recogn.	5.3	8.7	5.3	4.0	4.7
out vocab	(34.0)	(37.3)	(32.0)	(28.0)	(26.0)

The figures between the brackets denote a session with a threshold which was set to facilitate recognition. It is seen from this table that the system is rather noise resistent: the recognition rate does not drop very much. The recognition rate is rather low, but this seems not to be very uncommon (although the brochures of the manufactures report rates as high as 99 % or better). This (theoretical) figure could be obtained in a laboratory situation, with well chosen words (not applicable in practice), with a well-trained speaker and with a limited size vocabulary. Another problem is that the number of false recognized words, either from words within or outside the vocabulary, increase very strongly with an increasing noise level. Furthermore, the recognition threshold has a very strong influence on the figures. These latter points will hamper the application of speech recognition very seriously.

DISCUSSION:

The use of the human to simulate a speechrecognizer for investigation of command vocabularities, and the laboratory tests to produce a confusion matrix can be useful for many applications. It is a simple method and a rather inexpensive way to find the bottlenecks in a certain speech recognition application. The confusion matrix can also be set up for speech production systems: which words produced in case of an automatic speech feedback are sensitive to failure in understanding and confusion with other words, either inside or outside the used synthetic speech vocabulary. This is a well-known method for testing speechoutput systems.

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DESIGN OF AN OFF-THE-SHELF KNEE ORTHOSIS USING AVERAGING TECHNIQUES

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ABSTRACT

Three-dimensional leg geometries for fifteen volunteers were determined using plaster casting and computer digitizing. The computer files were averaged on a section by section basis to determine the average leg shape. Shape variations due to leg angle and muscular contraction were examined. Models were made of the 'average leg' from which an off-the-shelf knee orthosis was made.

INTRODUCTION

There are many different knee orthoses available today. The basic configuration is the same for all designs; hinge mechanisms placed at either side of the knee are attached to the leg using a cuff arrangement around the thigh and calf. The hinges themselves have different designs, mostly producing uniaxial or planar motion. The attachments to the leg can be as simple as a tubular elastic sleeve which stretches to fit any individual, or as complex as a rigid framework fabricated as a customized unit for a particular individual. The prices for the various designs cover a wide range, with the assumption that the more costly, customized orthoses give superior performance and comfort.

In previous studies (1,2,3) we measured the three-dimensional kinematics of the knee joint. The results of these investigations have shown that knee motion is similar in all individuals and that there is an envelope of allowable motion that a healthy knee exhibits. We have used this data to design a hinge mechanism which will follow the average neutral motion path of the knee (4).

It is our feeling that the most cost effective design of a knee brace, incorporating these hinges, would be an off-the-shelf knee orthosis. However, to maximize fit (and thereby comfort) the attachment to the leg must use a cuff design that closely matches the three-dimensional shape of the individual leg. Also, to achieve the most benefit from the hinges, they must be accurately positioned on the leg. Mal-placement of the hinges would reduce the benefit derived from using a hinge which is compatible with 3-D knee motion (5).

METHODS

Plaster casts taken of the right and left the legs of fifteen male volunteers to get 3-D negative molds. These long leg casts were applied with the leg in extension, the subject supine, and the muscles relaxed. The subject's foot was placed in a boot which held it in a vertical orientation, providing a consistent rotational orientation of the knee. The casts were marked with an axis system based on palpable boney landmarks, such that the sections could later be placed in the correct rotational orientation, and the location of one section related to the next.

For five individuals, additional casts were taken to investigate the effect of knee angle and muscular contraction on the shape of the leg. With the subject supine, casts were made with the knee at 45 degrees of flexion and at 90 degrees. One thigh cast was made with the quadriceps acting one with the hamstrings acting and a calf cast with the gastrocnemius acting.

The casts were oriented along a central axis for slicing into wafers, perpendicular to this axis. The first slice was made at the knee joint line with successive slices proximal or distal to the joint line at a spacing equal to 2% of the total leg length.

The interior circumference was traced from each slice along with the appropriate axis markings. These tracings were then digitized into the computer as 36 equally spaced points with the first point on the anterior axis line. During digitizing, the coordinates were normalized to the average leg length of 824mm. The sections were numbered from 1 (distal) to 49 (proximal) with 26 at the joint line.

A computer program with graphical display was written to average the legs on a section by section basis. Any data file could be displayed either in whole or in part, from any viewing direction. The program could plot the peripheries only, or a wire frame model with hidden line removal.

As a pre-processing stage, the axes of each leg were redefined with reference to the long axes of the tibia and the transverse axis of the knee (1,2). This was done with reference to anterior points on the shank, and utilizing radiographic data from previous knee specimins. The average valgus angle was computed and the thigh sections of all legs transformed to the average valgus. The above ensured reasonable 'concentric' overlap of equivalent sections between the different legs.

The method of averaging the profiles is shown in figure 1. Radial lines PQ were drawn from the mean geometric center to intersect the profiles at A, B, & C. The mean coordinates of A, B, & C were defined as a point on the mean profile. The procedure was automated, such that an average of any specified number of legs could be obtained simply by keying in the appropriate data file names.

Each section was printed out, actual size, from which a sectionalized model of the average leg was built. This stacked section model wasfilled with plaster, smoothed, and used for vacuum forming of thermoplastic cuffs.

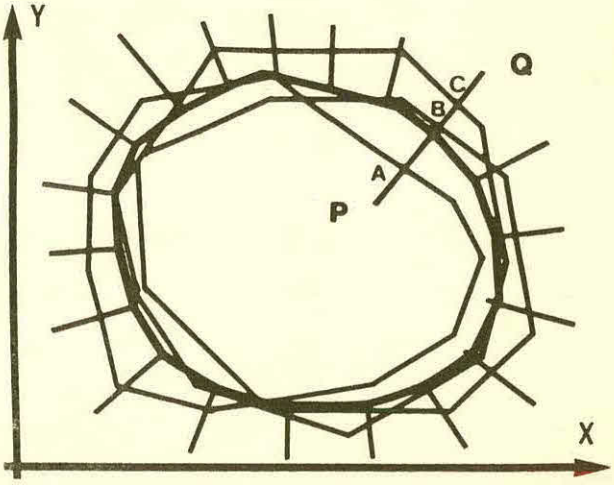

Figure 1. Method used in averaging sections.

RESULTS

When all sections of a given number were superimposed, there was considerable variation. A typical group of profiles on the shank is shown in figure 2, along with the computed mean profile. The mean profiles form sections 1-49 constituted the basic data file for 'the average leg'. The 'average leg' is shown from three different views in figure 3. The axes are important in that they relate the boney landmarks to the external geometry of the leg and are used in the positioning of the orthotic hinges on the cuffs. The model is very realistic even though no smoothing was carried out. The stacked section model of the 'average leg' is shown in figure 4. Different sizes are readily made by scaling up or down. Displays were generated to show the shape changes observed under different conditions. Figure 5 shows the shapes with the leg at 0 and 90 degrees of flexion, with relaxed muscles. The shank changes were small, but at 90 degrees, the thigh narrowed in the A-P view, narrowed in the M-L view just above the knee, but further up the thigh, the width increased. Under quadriceps action, the thigh narrowed uniformly in the A-P view, and expanded uniformly in the M-L view. The shape changes were much reduced just above the knee. Very similar changes were seen when the hamstrings were acting. When the gastrocnemius was acting, there were relatively small changes seen in the A-P and M-L views. The changes were particularly small in the region just below the knee with no change along the front of the tibia. The main changes were bulging of the body of the muscle, and contraction just below the belly of the muscle.

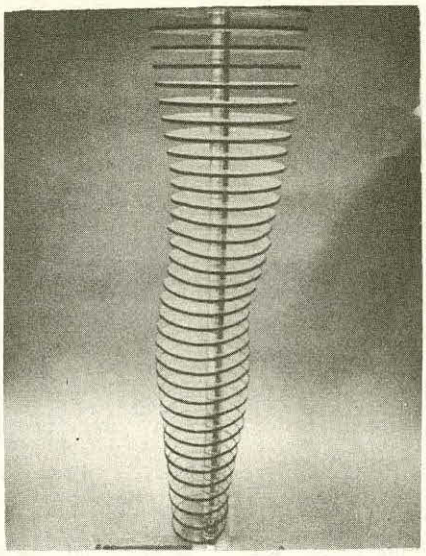

Figure 4. Stacked section model of average leg.

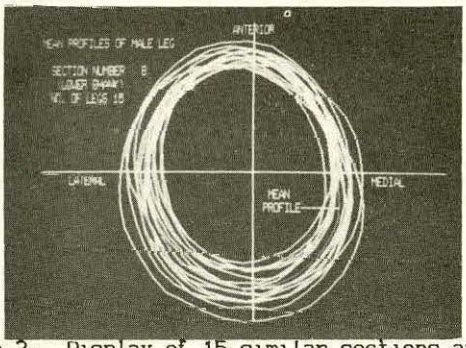

Figure 2. Display of 15 similar sections and thier average.

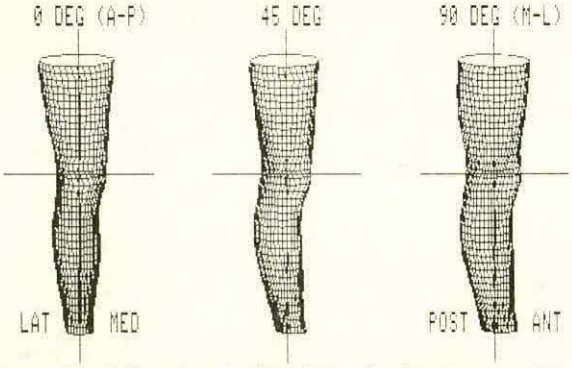

Figure 3. Wire frame display of the average leg from three different views.

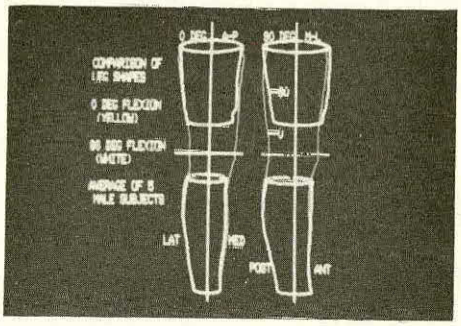

Figure 5. Display of shape change resulting from change in leg angle.

APPLICATION TO LEG BRACE DESIGN

The data we have obtained (though from a small sample of legs at present) has been applied to the design of an off-the-shelf knee orthosis (figure 6). Leg models in several sizes were made, from which a range of sizes of the knee brace were produced. The best fit for a particular individual will depend on the closeness to an available size as well as the difference in leg shape from the average. However, even in the case of custom molded cuffs, the shape changes due to flexion angle and muscular action will result in some degree of mismatch during dynamic motion. The mismatch in fit will have to be taken up by flexibility of the cuffs and by a stretch range in the straps.

We have chosen two rear-opening plastic cuffs for use in this knee brace. The thigh cuff was molded to the average leg with quadriceps acting. It is thought that the relaxed quadriceps will more easily deform to fill this cuff shape then would the flexed muscle fill a cuff molded to the relaxed muscle shape. This cuff geometry should result in better fit and comfort during dynamic activity.

Following the earlier innovative work of others (6,7,8) we are working on a low-cost but rapid and accurate method of determining 3-D leg geometry. This is based on a bank of 3-D digitizers stroked across the leg with software to generate a mathematical surface model. We plan to apply this to expanding the data base, and to customized cuff design and manufacture where applicable.

Figure 6. Knee orthosis based on average knee motion and average leg geometry.

ACKNOWLEDGEMENTS

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8.2

MILEAGE RELATED CHANGES IN THE MECHANICAL PROPERTIES OF RUNNING SHOES

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The change in shock absorption properties of running shoes was evaluated as a function of miles run. Differenct models of running shoes encompassing a wide range in retail price were obtained and mechanically tested.

An approximate 33% difference in the initial shock absorption was observed in the different shoe models. In general, the shoes retained approximately 73% of their initial shock absorption capability after 50 miles of simulated running, and approximately 67% after 100 to 150 miles. Between 250 and 500 miles the shoes retained less than 60% of their initial shock absorption capacity. No differences in shock absorption characteristics were apparant based upon either shoe price or the manufacturer model. The results of shoes tested by the volunteer runners also showed a marked reduction with approximately 70% of initial shock absorption retained at 500 miles.

INTRODUCTION AND BACKGROUND

The dual function of a running shoe is to provide both shock absorption and motion control for the runner. Degenerative changes in bone and cartilage as well as low back pain and stress fractures have been related to impulse loading in the musculoskeletal system. In this respect, the running shoe must be viewed as an important component in both treating and preventing injuries. While it is generally agreed that the shock absorption characteristics of running shoes deteriorate with miles run, very little data exist on the specific rate of this deterioration. It was the objective of this study to quantify the rate of change of shock absorption properties of running shoes as a function of miles run.

MATERIALS AND METHODS

Running shoes were obtained in different models encompassing a wide range in retail price and manufacturers (Table 1). Approximately 40 models were studied including many of the current midsole design concepts. The shoes were mounted in a specially designed test fixture and loaded to simulate the repeated heel strikes of running. The shoes were cycled from 0 to 150 kilograms (0 to 330 lbs.) at a frequency of 2.5 hertz. The shock absorption of the shoes were evaluated at intervals to the equivalent of 500 miles run. In addition, the forefoot of the shoes were tested in a similar fashion.

TABLE 1
Summary of Shoe Models Tested

Manufacturer	Model
Adidas USA, Inc. Rutherfor, NJ	Boston Marathon Oregon Seattle Zx 500
Asics Tiger Corp. Santa Ana, CA	X-Caliber GT Alliance
Avia Athletic Footware Portland, OR	RT 680
Brooks Shoes, Inc. Rockford, MI	Chariot Graphlex
Converse, Inc. Wilmington, MA	Laser
Etonic, Inc. Brockton, MA	Quasar
Hyde Athletic Co. (Saucony) Cambridge, MA	Dixon Jazz Jazz Plus Quest
Keds, Corp. Cambridge, MA	Conquest
Muzik	Muzik
New Balance Athletic Shoes Brighton, MA	990 700 770
Nike, Inc. Beaverton, OR	Daybreak Odyssey Pegasus
PUMA USA, Inc. Framingham, MA	Easy Rider

To determine the reproducibility of measurement and the variability within shoe lots, multiple shoes of the same model were tested. Additionally, the recovery of shock absorption capability of the shoes which may be masked by the continuous machine loading rather than the intermittent nature of actual running was assessed. These shoes were cycled and allowed to recover after testing for periods of 24 and 48 hours and then retested. In order to correlate the machine simulated running with actual running shoes were tested after having been worn by volunteer runners during normal training. All shoes worn by volunteers were tested at comparable mileage intervals to the machine simulated running.

RESULTS

The shoes were found to have an approximate 33% difference in initial heel strike shock absorption capability (Table 2). However, after machine testing was begun no trends or differences in absorption characteristics were apparent based upon either the shoe price, manufacturer or a particular manufacturer's model. Generally the shoes retained about 73% of their initial shock absorption after the first 50 miles of running and retained approximately 67% of the initial absorption at 100 to 150 miles. Between 250 and 300 miles the shoes retained about 58% of their initial shock absorbing capability and at this point the losses appeared to level off.

TABLE 2
Shock Absorption Characteristics as a Function of Milage for all Shoes Tested

Miles	Shock Absorption (Newton-meters) Mean ± SD	% Initial Shock Absorption Mean ± SD
0	10.5 ± 1.4	100 ± 0.0
5	9.0 ± 1.0	86.2 ± 5.6
10	8.8 ± 1.0	83.9 ± 7.5
25	8.1 ± 0.8	78.0 ± 5.2
50	7.4 ± 0.5	73.0 ± 4.6
75	7.1 ± 0.5	70.6 ± 4.6
100	6.9 ± 0.5	67.8 ± 6.0
125	6.8 ± 0.4	66.7 ± 6.7
150	6.6 ± 0.4	65.8 ± 6.7
200	6.4 ± 0.4	62.3 ± 5.4
250	6.1 ± 0.5	59.6 ± 7.5
300	5.8 ± 0.4	57.2 ± 7.5
500	5.5 ± 0.5	53.2 ± 6.1

The testing of multiple shoes of the same model indicated that the results from the machine load were reproducible and that good quality control exists within the manufacturers. The maximum deviation between shoes at any interval was about 10 percent with most differences in the range of 2-5 percent.

The recovery characteristics of shoes tested allowed to recover demonstrated essentially no change in the measured values. The maximum differences observed were less than 3% indicating that the shoes did not return to their more shock absorbent condition (Table 3).

TABLE 3
Recovery Characteristics as a Function of Time for Two Different Shoe Models*

Miles	0 hr	24 hr	72 hr	Mean ± SD
Shoe X				
5	10.8	10.9	10.7	10.8 ± 0.1
10	10.2	10.3	10.5	10.3 ± 0.2
25	9.4	9.5	9.5	9.5 ± 0.1
50	8.9	8.9	9.0	8.9 ± 0.1
100	7.1	7.1	7.2	7.1 ± 0.1
Shoe Y				
5	8.8	8.7	8.7	8.7 ± 0.1
10	8.3	8.5	8.4	8.4 ± 0.2
25	7.7	7.5	7.6	7.6 ± 0.1
50	7.4	7.3	7.4	7.4 ± 0.1
100	7.0	6.9	6.9	6.9 ± 0.1

* Shock absorption (Newton-meters).

The shock absorption lost by actual running was not as great as the machine simulated running. The machine simulated running produced about 25 percent greater reduction in shock absorption as comparing to actual running.

DISCUSSION

All shoes tested demonstrated very little difference in the characteristics of the energy absorbed at heel strike. This would suggest that choosing a shoe based solely upon price or initial shock absorption may be misleading. However, other aspects of shoe performance such as motion control and stability, may be reflected in the price of the shoe and must be considered when selecting a specific model.

The less severe deterioration of the shock absorption in the in vivo loaded shoes as compared to those machine loaded is probably the result of a number of factors. The machine simulated did not follow the normal gait pattern of the human runner resulting in the concentration of forces within the heel portion of the shoe. Also, the surfaces encountered in normal running vary from that of concrete to grass and dirt, in contrast, the test machine surface would have to be considered to be that of an extremely hard running surface. Finally, the normal shock absorbing properties of both the hard and soft tissues would tend to decrease the shock loading of the shoes.

CONCLUSION

Although no evaluations were performed on the motion control and stability of the various shoes tested, it was apparent from the various designs that much work has been done in this aspect of shoe design. However, the new design innovations and combinations of midsole materials apparently do very little to enhance the shock absorption properties of the shoes. Thus, if proper long term protection is to be provided it would appear that more research needs to be done in improving the midsole material shock absorbing characteristics of running shoes.

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INTEGRATION OF ORTHOSES WITH FUNCTIONAL NEUROMUSCULAR STIMULATION IN PARAPLEGIC SUBJECTS

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ABSTRACT

Six paraplegic subjects are walking with functional neuromuscular stimulation (FNS) delivered to percutaneous electrodes implanted in up to thirty-two muscles of the pelvis and lower extremity. Because of this extensive muscle use and computer-controlled programming of the stimulation, bracing has been reduced to a minimum. An ankle-foot-orthosis was designed to control and protect the foot; it was used by four subjects with satisfactory results.

INTRODUCTION

Continued advancement of research in FNS points the way to new levels of independence for paraplegic individuals. In particular cases, such research has produced systems allowing functional ambulation (8). However, these systems are a combination of FNS and orthosis. While FNS provides initiation of movement of body segments, the bracing stabilizes and protects bones and joints. Bracing of a joint is required if stimulation is incomplete or absent to those muscles which supply torque to that joint. Since FNS can restore function to levels approaching normal, it is logical to strive for reduction of bracing and maximization of controlled (closed loop) FNS. Replacing braces with electrical stimulation is becoming feasible as computer control over muscles improves and as the number of muscles under this control increases. The latter is possible with intramuscular electrodes. This paper describes the lower-limb bracing currently in use with internal FNS at the Cleveland VA Medical Center; as well as the rationale for its use and design.

Since the value of braces to functional ambulation for the paraplegic person remains controversial and because the wheelchair is the primary means of locomotion, it is necessary to define the goals of FNS-induced walking.

We know that energy expenditure is a significant factor affecting the acceptance of any system of locomotion. It has been reported that the energy cost of ambulation in terms of kcal/kg/min for wheelchair users is approximately equal to the energy cost of normal walking (4). In contrast the energy cost for the paraplegic ambulating with braces, in terms of kcal/kg/m, can be as high as nine times that of the normal (3). In addition, brace ambulation is reduced to one-half the normal speed. Our goal is to reduce energy consumption to a near normal level by stimulating muscles in the lower extremity, pelvis and trunk. Erect posture of the trunk aids in alleviating the upper extremities of their role in weight support. The work load becomes distributed over a larger muscle mass when the lower extremities are stimulated with FNS. This has a direct effect on the overall efficiency of the system.

Paraplegic individuals traditionally have been taught to ambulate with swing-to or swing-through gaits. The development of the Reciprocating Gait Orthosis (RGO) and the Hip Guidance Orthosis (HGO) (1) brought about a change in thought in three ways. First, it is more natural to alternate flexion and extension of the hips; second, body weight should be supported through the lower extremities at all times; third, posture should approximate the normal. In other words, the head, spine and pelvis should remain relatively vertical for standing and walking. While the RGO accomplishes these three things, it also has severe limitations which force the user to walk mechanically. The knee joints must remain locked and the ankles rigidly fixed. Because the brace extends from the foot to the xiphoid process (THKAFO), transverse rotation of the extremities and trunk is limited. Lehneis (6) noted the importance of transverse rotation in the design of the Spiral Helix Below-Knee Orthosis. He wrote: "Ideally, orthoses should provide joint motions which are as similiar as possible to those of the normal extremity." The goal here is to produce a natural gait with FNS along with the minimum amount of bracing. While some bracing may be required, mechanical joints should be aligned close to the anatomical center of rotation so that they will not have an adverse effect on motions of the extremities.

To be functional outside the home, the paraplegic person must negotiate stairs, curbs and ramps in addition to ambulating on level ground. To overcome any of these obstacles the body must change joint angles and adapt to the height of the step or curb and to the inclination of the ramp. For example, the hip and knee must flex and the foot dorsiflex while a lifting force raises the body.

Each subject can use up to thirty-two electrodes (from a total of 40-70, about half of which are redundant) which have been implanted in the pelvis and lower extremities . The muscles innervated about the ankle are the gastroc-soleus complex and tibialis anterior. The primary functions of these muscles are plantar and dorsiflexion, respectively. Presently, no attempt has been made to actively control the foot and ankle in the medio-lateral plane. In addition, the inherent instability of the talo-crural joint in plantarflexion,

subtalar joint, and the tarsal joints distal to this would make it extremely difficult to guard against injury without the aid of an orthosis. Even on level ground some subjects exhibit severe laxity in these joints with inversion of the foot. This becomes more apparent with plantarflexion due to the inverting force component of the medial gastrocnemius. We must also assume that the patient will encounter irregular walking surfaces outside the home. For these reasons we have chosen to use ankle-foot-orthoses bilaterally.

METHOD

The orthosis is custom-fit from a cast (negative) taken of the subject's leg. The process of fabrication begins with modification of the mold (positive). The stainless steel joint is positioned medially at ankle center and the upper and lower segments are shaped to follow the contours of the mold. The orthosis is then laminated and trimmed after curing.

The design of these braces would not have been possible without the use of composite materials. Specifically, the wall thickness of the plastic had to be very thin, yet be very strong and rigid. Unidirectional carbon fiber was chosen for its desirable characteristics. Carbon fibers are silky, black, small diameter filaments made up of millions of long chains of carbon atoms. They have a high strength and modulus and a low density. Strength and stiffness can be tailored by orienting the fibers with respect to the lines of stress (7). Maximum strength is achieved with fibers arranged parallel to these lines. Fiberglass is used as well and is usually interpositioned between layers of carbon to geometrically increase rigidity. Acrylic resin is used as the matrix. Heating the plastic will allow some reforming for adjustments.

The orthosis is basically a medial strut (single upright) overlaying the medial head of the gastrocnemius which extends from three-quarters of an inch below the level of the neck of the fibula to the stainless steel joint head at the medial malleolus. A two-inch Velcro strap encircles the leg and secures the orthosis proximally. The joint head then continues distal and is anchored into a UCBL insert (2). The insert is modified in the standard way except the posterior portion is completely cut away. This would normally wrap around the calcaneus. Also, the insert is made full foot to support three-sixteenth inch Sorbothane sheeting that extends from the heel to the toes. A shoe one-half to one size larger than the regularly worn shoe will accommodate the added bulk without cramping the foot.

Important design considerations were to make the orthosis light weight, strong, flexible in some areas while rigid in others, cosmetically acceptable and easy to put on and take off. More specific design criteria follow.

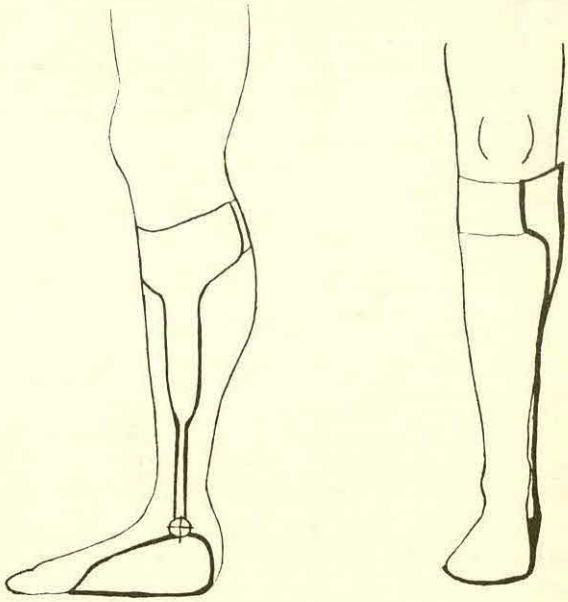

FIGURE 1. Medial and Anterior Views of AFO

The ankle joint is a single, medially placed, free-motion joint which is aligned anatomically according to Inman and Isman (5). The upper and lower halves are embedded in composite materials eliminating screws and rivets (Figure 1). Eventually, this joint will be outfitted with an electrogoniometer to provide the FNS controller with information about the position of the ankle.

The posterior portion of the foot insert is cut away because it was felt that it served no purpose and without it the heel could rest comfortably against the heel of the shoe. Also, the shoe material is softer and more forgiving than plastic and would lessen the chance of a pressure area developing over the calcaneus. Along the medial and lateral trim the plastic has been made flexible to avoid having the edge of the plastic cut into the skin as the foot slightly shifts as the subject walks. The flexible trim also allows the foot to swell without adverse affects. From the metatarsal heads distally the plastic is flexible to allow the foot to bear weight in plantarflexion. The remaining aspect of the insert is rigid and provides the necessary support and protection to the foot.

The Sorbothane insert is used to dissipate energy by absorbing shear and compressive forces inorder to protect the insensate foot. In the future, pressure transducers are to be housed in this insert at the metatarsal heads and heel.

RESULTS

Four subjects have been fitted with the ankle-foot-orthosis in the last six months. They wear the orthoses when using FNS for

walking. They walk on the average four to five days per week and range between 500 and 1500 feet per day. No pressure sores have arisen from the orthoses to date. Donning them adds very little to the time to put a shoe on. All subjects state they feel the brace is cosmetically acceptable. One brace was remade due to inadequate clearance of the joint from the leg.

DISCUSSION

We will continue to evaluate the orthotic needs of subjects using FNS as research progresses. Presently, we feel that ankle-foot-orthoses are indicated for these individuals. The material used in fabrication will allow adaptation in design. In the future, a locking mechanism may be developed for the ankle as well as integration of electromechanical devices for feedback control.

On one end of a continuum paraplegic individuals can ambulate solely with braces. On the other end, they may rely on FNS alone. Our conclusion is to develop a system of ambulation that is primarily FNS with minimum bracing needed for safe and effective walking.

ACKNOWLEDGEMENTS

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COCONTRACTION TO SELECT PROPORTIONAL MYOELECTRIC CONTROL MODES

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A strategy has been developed to permit an amputee to select more than one function for proportional myoelectric control with two muscles. Cocontraction provides the basis for function selection, allowing the prosthesis to shift from elbow operation to hand control and vice versa. Return to the original function may be configured to occur during total relaxation of both muscles or by subsequent cocontraction. The system was developed as an alternative to conventional cable operated terminal devices. This new alternative enables terminal device control over a broader range of body motion.

INTRODUCTION

An increasing number of above-elbow amputees are being fit with myoelectric elbows in place of body powered prostheses. Typical proportional myoelectric elbows are the Liberty Mutual Elbow and Utah Artificial Arm. While these myoelectric elbows offer enhanced utility over body power, the same level of enhancement is more difficult to achieve in the terminal device. The Utah Arm includes an option whereby myoelectric control can be passed to the hand whenever the elbow stops moving, but some amputees experience difficulty in learning the rapid cocontraction required to resume elbow operation.

The terminal device options for independent control of the terminal device of a myoelectric elbow are:

1. Body powered hook operated by Bowden cable.
2. Switch controlled electric hand triggered
3. Switch controlled electric gripper triggered by a small body motion.

These options share a common control source. Upper body motion is translated into cable pull for a hook and into strap motion for a switch controlled electric hand or gripper. Both methods are limited when the limb is in certain positions where the cable or strap system is either too loose or too tight. The hook is further limited for amputees who cannot generate sufficient body motion to operate it.

There are many amputees who would benefit from a way to eliminate body motion and to control the terminal device with the same two myo- electric sites that control the elbow. What is needed is a selection scheme which will permit shifting to hand control and back at will. A good scheme is intuitive and requires little concentration. It should be difficult to make the shift inadvertently. We report here a scheme for shifting control which is both reliable and easy to operate.

Cocontraction Is A Normal Muscle Action

Cocontraction is a normal response of the antagonist muscle groups in the arm. For instance, just prior to hand contact with an object during grasping, there is an increase in the cocontraction level of the arm's antagonist muscle groups. This cocontraction stiffens the joints. Such a rise in stiff- ness is a normal part of human muscle response which improves stability. Remnants of this cocontraction are often observed in amputees during terminal device manipulation.

Controlling An Elbow Myoelectrically

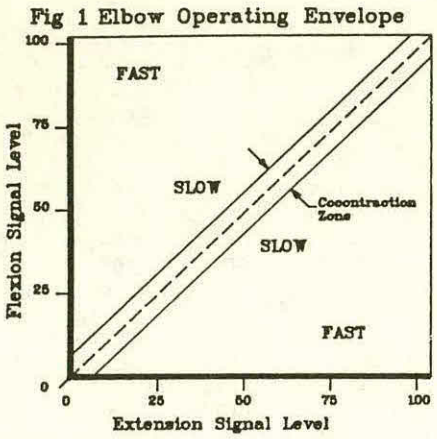
In observations of experienced myoelectric elbow wearers, we find that the transition from extension to flexion results in a smooth drop in the triceps signal accompanied by a corresponding rise in the biceps level. The midpoint of the transition passes through a region of low level cocontraction. This cocontraction during the transition also occurs with higher level shoulder disarticulation amputees.

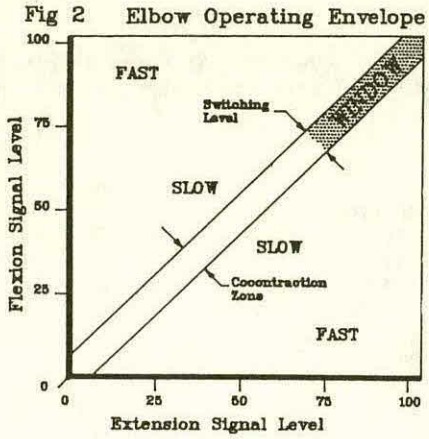
THE EXPERIMENTAL SETUP

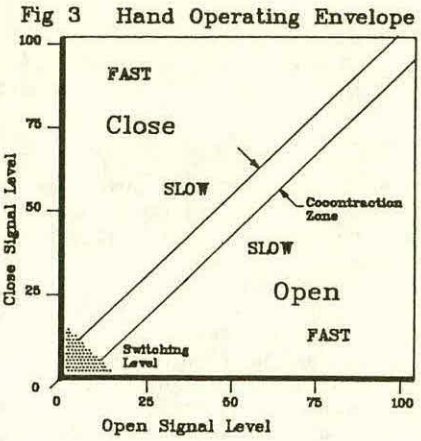
The Liberty Mutual Elbow is a myoelectric prosthesis with proportional speed control. The control signals are obtained from electrodes at two muscle sites. These myoelectrodes amplify and filter the signals for the following stages. Variable gain amplifiers and filters follow and provide a range of adjustment for patient and muscle site variations. The gains are typically set so that an amputee perceives that an "equal effort" is required to achieve the maximum signal on each of the two channels. Once the two gains have been adjusted, the signals can be con- sidered to be normalized to the same maximum value.
The normalized signals are rectified to produce DC signals of opposite polarity for circuits in the following stages. Figure 1 shows one way to think about these two signals. Here they are plotted with flexion on the Y-axis and extension on the X-axis. The elbow circuits generate the absolute value of the net signal which can be thought of as the distance from the dashed diagonal line. This "distance" controls speed. An independent circuit controls direction. No motion is permitted when the net signal lies in the cocontraction zone on either side of the diagonal.

Selecting With Cocontraction

Figure 2 shows in used for selection. The width of the cocontraction zone is controlled by a circuit which compares the difference between the two myosignals to a reference. Differences below the reference are "cocontraction". The width of the zone is adjusted by changing the height of the reference. To avoid inadvertent operation of all low levels of cocontraction must be rejected, because they are routinely present when amputees control the elbow. Both myosignals are summed and compared to a reference value. This circuit







area. It is important that the switching level in Figure 2 be set close enough to the upper right corner of the plot. Otherwise, the normal cocontraction in going from flexion to extension will activate the shift.

Returning Operation To The Elbow

Two strategies are available for reverting to elbow operation. Figure 3 shows an area of total relaxation in the lower left quadrant. For many is a good trigger for state subjects relaxation change. However, the subject needs to maintain some level of contraction of at least one muscle or the shift will occur. To keep the change from being too easy to make, a time delay is helpful. The shift then occurs after a set period of total relaxation. For subjects that find it difficult to tell whether they are relaxed or not, the alternate strategy is to use the original trigger a second time to revert to elbow operation. This requires remembering which mode one is in, however, one need only be good at generating one trigger signal.

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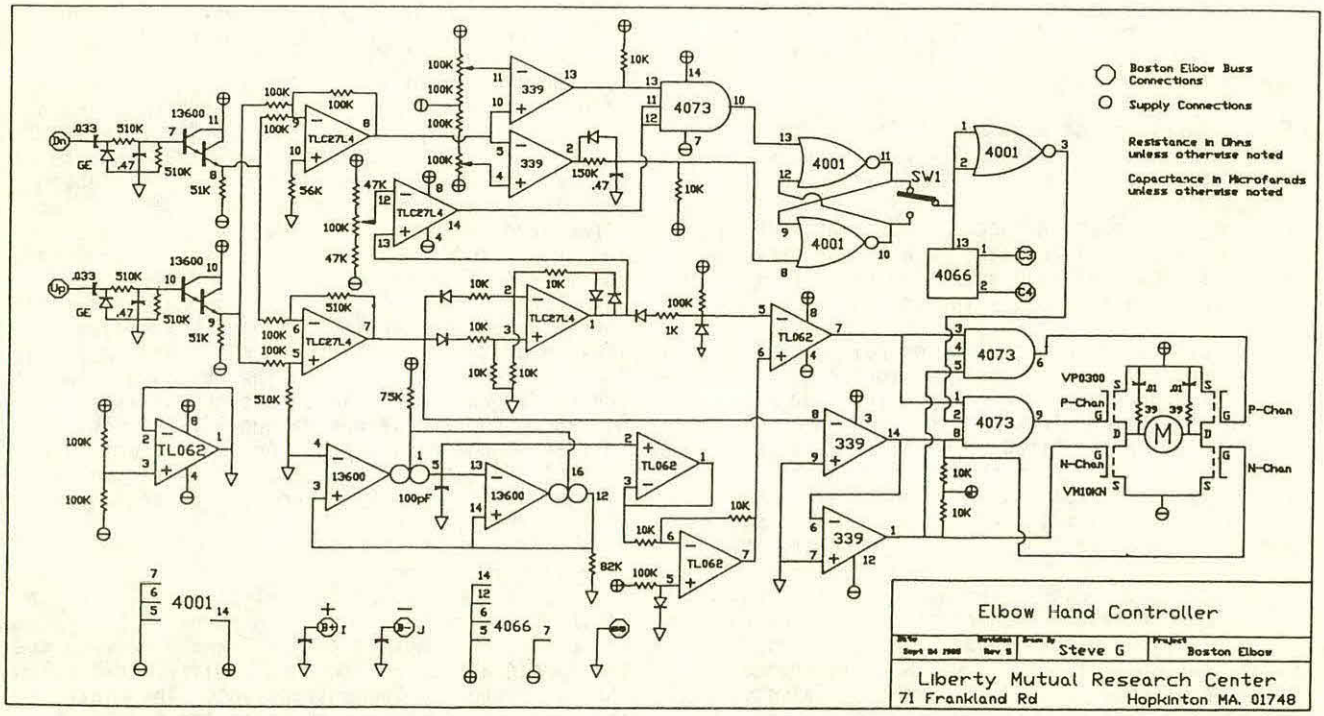

Fig. 4. Circuit used to test the concepts discussed in the paper. Not shown is the circuitry of the elbow control and power boards. Use this diagram for ideas. There are severe restraints due to the interaction of the predesigned control and power boards. Contact the author.

SPECIFICATION FOR A NEW MULTIFUNCTIONAL HAND PROSTHESIS

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ABSTRACT

A project of a multifunctional hand prosthesis based on a new pattern of prehension is presented together with an emphasis on certain functional characteristics. A computer aided three-dimensional interactive applications program has been utilised firstly to verify the geometry and the kinematic behaviour of the proposed model, secondly, as a conceptual and a designing tool.

INTRODUCTION

A project of a multifunctional hand prosthesis has been undertaken in an attempt to overcome some of the functional prehensile weaknesses associated with conventional types of prostheses. For this task, our approach is mainly based on the exploitation of a new line of motion for the thumb when restricted to a single active degree of freedom. Obviously, this restriction is necessarily a compromise over the natural model. However, we have shown in an ergonomic analysis of prehensile activities with daily used objects that: despite its five degrees of freedom, the natural thumb uses a prefered plane of motion. The plane found intersects the plane of flexion of the middle finger with an angle ranging from 45 to 55 degrees (1). For mechanical considerations, an angle of 45 degrees has been proposed for the new hand.

In conventional prostheses, the "pinch" or the opposition of the thumb with the first and second finger has traditionally been obtained from their coplanar motions. The fact that the same pattern of prehension is also used with multi-functional prostheses may indicate that it represents an acceptable compromise. An exception to this trend is the hand developped by E.W. Davies and al from the Orthopaedic Bio-Engineering unit, Edinburg(2). The prosthesis produced by this group is called the OBEU Hand. This hand is a body powered monofunctional hand in which the thumb moves about an axis inclined at angle of about 60 degrees to the first and second finger axis. Comparing it with prostheses currently available, the group have reported improvements in cosmesis and function.

Our decision to develop a new hand prosthesis has resulted from the ergonomical analysis and previous clinical experimentations with a hand orthosis utilizing the proposed pattern of prehension(3). These two approaches have shown that a multi-functional hand prosthesis based on the proposed pattern should offer the following advantages over the traditional approach:

- improves the quality, the stability and the cosmesis of the grip function;
- minimizes the use of body and arm compensatory movements; during the hand positioning and utilization phase of the object;
- improves visibility, particularly when picking up an object;
- allows a better orientation for most of the objects held for use.

GENERAL SPECIFICATIONS

The general specifications are based, and shall meet those proposed by H.W. Kay and M. Rakic(4). Some specific requirements about the projected multifunctional hand prosthesis are listed as follow:

1. The palm and fingers segments shall resemble the human model in configuration and proportions.
2. The fingers and thumb shall be actively powered by a single electric motor located in the palm.
3. Maximal opening between the thumb and the second finger tips shall not be smaller than 9 cm.
4. The motor-gearbox unit shall be of the self-locking type and free of objectionable noises.
5. The total weight of hand and glove should not be higher than 500g.

FUNCTIONAL CHARACTERISTICS AND THE USE OF CAD TECHNIQUES

The adoption of a new line of action for the thumb cannot be done simply by grafting a new thumb on a conventional hand prosthesis. The reason stems mainly from the fact, that the objects will be approached and grasped differently. For example with a conventional prosthesis, a cylindrical glass will be aligned along an axis perpendicular to the fingers axes, as with the pattern of prehension proposed here; the glass will be inclined within the palmar arch. As a consequence such hand design requires a different conceptual approach particularly regarding:

a) the geometry of the palm and the joints location and orientation;
b) the structural and the kinematic behaviour of each finger segment.

Furthermore, designing techniques using standard mathematical approach have proved arduous. Fortunately, we have at our disposal a powerful computer graphics aided three-dimensional interactive applications program (CATIA)(5) which in spite of some limitations has proved to be an efficient conceptual tool. The following is a brief description of some of the application possibilities offered by the program. In the graphical mode, it permits to create models, visualize and analyse several geometrical elements in the plane or in space. In the two animations modes, mechanisms or Robots, it permits to visualize 2D and 3D articulated movements and to execute instant calculations on important physical parameters.

The above mentioned techniques have been utilized here to verify the geometry of the proposed model and to a large extend to determine its overhall kinematic behaviour. The following represents the proposed functional characteristics.

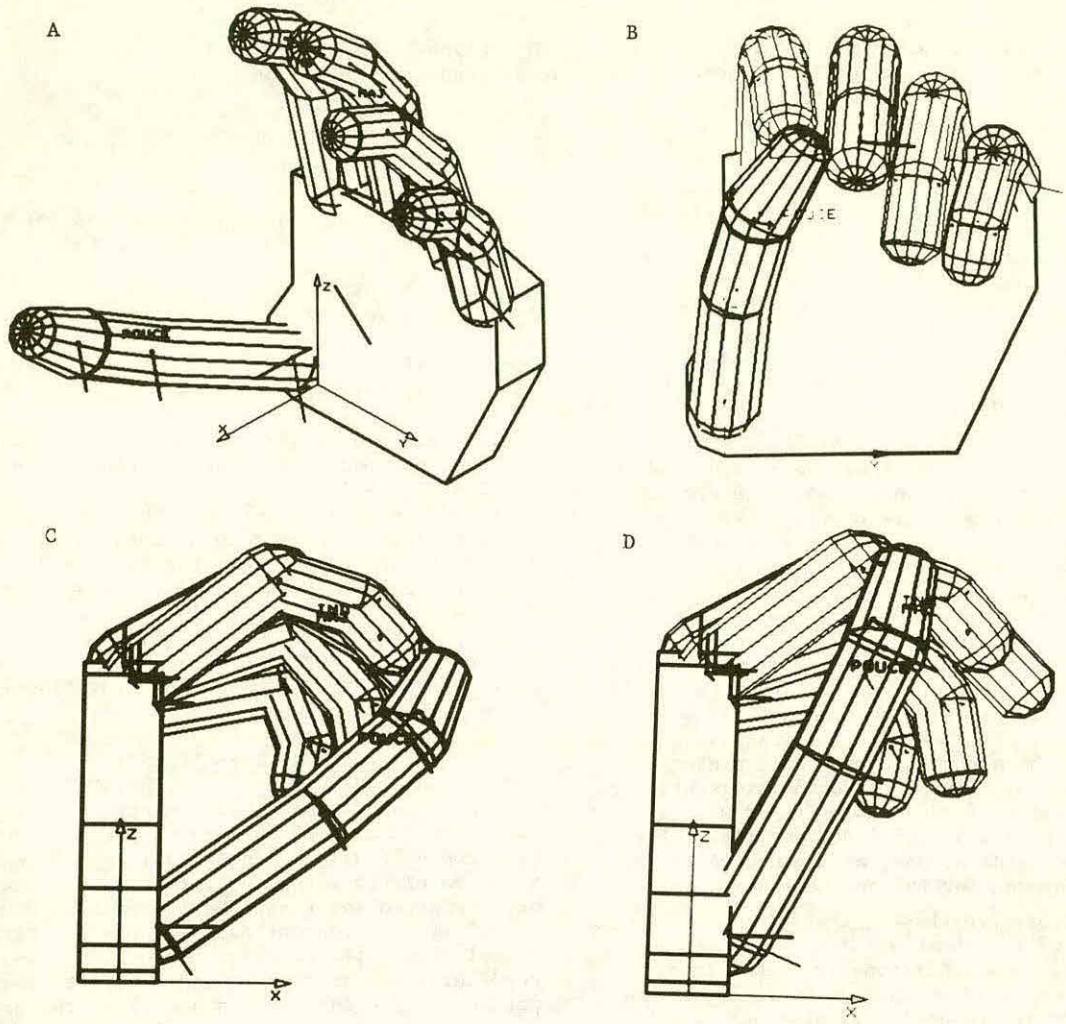

Figure 1
CATIA three-dimensional representation of the proposed prosthesis.
A. Hand open in resting posture
B.C. Frontal and lateral view of the tridigital prehension pattern.
C. Lateral view of the lateral prehension pattern.

The palm should be anthropomorphous in appearance with a certain degree of resilience at some pre-selected points.

The fingers

- The metacarpo-phalangeal (MP) joints will describe as in the natural hand a domed curve in both the transversal and longitudinal plane of the palm.
- The four fingers will flex about the MP joints and the proximal interphalangeal (IP) joints. The distal IP joints will be bended at an angle of 30 degrees.
- The rate of flexion of each digit will increase when going from the first to the fourth. Combined with the above joint geometry, this will give a tridigital pinch with the thumb opposing the tip and the lateral side of the first and second fingers respectively (see fig. 1b and c). This will also contribute to decrease the functional ergonomical interference usually associated with the third and fourth digits of the conventional hand prosthesis.
- The fingers velocity should be such that the tridigital pinch is executed in a maximum of 0.8 second.
- The fingers segments should be activated by a mechanism that will allow independant adaptability of the two active phalanxs.
- The fingers should flex when pushed by an external force.
- The fingers tips should spread apart during extension.

The thumb

With conventional hand prosthesis the thumb has been generally mobilized about the MP joint. With the proposed pattern of prehension it will be actively flexed about the carpometacarpal and the IP joints. The MP joint will be bended at an angle of 10 degrees. A third but passive axis of rotation should permit the lateral and possibly the fist pattern of prehension.

Finally, a gripping force of 45N has been specified for the tridigital pinch. This may be

difficult to meet. However, we believe that a stable grip can be attained with a substantial lower force.

CONCLUSION

A real hand substitute is behond the reach of the present technology, and any attempt to provide a solution calls for new compromises. Choosing the best compromises remains a prerequisite for any successfull solution. This task and many others may benefit from the use of CAD techniques. Such techniques will not replace the creative processes of the designer, but will enable him to cope with move complex problems.

ACKNOWLEDGEMENT

This is a joint project of l'Ecole Polytechnique de Montréal and fo the Research Centre of the Institut de réadaptation de Montréal. The support has been provided by the Institut de recherche en santé et sécurité au travail du Québec , grant no. RS-85-16.

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LOW COST TRANSDUCERS FOR JOINT POSITION MEASUREMENT ON A PROSTHETIC ARM

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ABSTRACT

This paper describes the development of several low cost transducers which can be used to measure joint position, velocity, and acceleration of prosthetic limb components.

INTRODUCTION

Technology has brought us a long way from the days of the wooden arm with a hook on the end. Today, we have motorized arms with joints which are similar to those of natural arms. However, the modern day prosthetic arm has no capacity for conveying any judgement of movement back to the limb controller. Consider the task of using a joystick operated robot arm to pick up an electrical plug off the floor and to insert it into a wall receptacle. This is analogous to using a present day prosthetic limb. That is, every day tasks are often difficult or frustrating for a prosthesis recipient.

This project involved the development of low cost transducers which can be used to measure joint position, and it's derivatives, velocity and acceleration, on a prosthetic arm [1]. The signals from these transducers can then be made available to limb controllers such as the microprocessor based limb controller described in Hortensius et al [2] which is capable of making use of these signals.

IMPLEMENTATION

The transducers were fitted to a four function prosthetic limb (hand open/close, wrist rotation, elbow flexion/extension, and humeral rotation).

The hand is a standard OTTO BOCK hand where finger movement is controlled by gears. Therefore, by measuring the rotation of the gears, the opening of the hand between the thumb and index fingers can be found. A standard multiturn potentiometer is used in this measurement. A bracket which is fastened with a screw on the prosthetic limb is used to hold the potentiometer. The shaft of the potentiometer is inserted into an aluminum tube which is attached to a screw at the center of a gear controlling the finger movement. The exact placement of the potentiometer can be seen in Figure 1.

For rotational measurements at both the wrist and the humerus, several different transducers were tested. Both the wrist and the humeral rotation were implemented using a standard OTTO BOCK wrist.

One transducer that was used was a Hall effect device. This particular Hall effect device is essentially a switch that is activated by the presence or absence of a magnetic field. Magnets were mounted on the fixed portion of the wrist or humerus and the Hall effect device was attached to the rotating part of the arm (either the lower arm for humeral measurement, or the hand for wrist measurement) as shown in Figure 2 for the humeral measurement. As rotation occurs the Hall effect device passes over the magnets causing it to switch on and off, resulting in a pulse train corresponding to the direction and speed of rotation.

Another device which is similar in operation to the Hall effect transducer is the reed switch which is mounted like the Hall effect transducer (see Figure 2). As the reed switch passes over the magnets the switch closes and opens yielding the same pulse train as the Hall effect device.

The final device used to measure rotation at the wrist and humerus was a source and detector assembly (SDA). This device is a light source and a light detector mounted in the same assembly. A gap is provided between the two devices allowing the light to be blocked at will. For rotation measurement, an aluminum ring with holes cut into it was attached to the rotating portion of the arm. The SDA was attached to the fixed part of the arm in such a way that the aluminum ring was placed into the gap (see Figure 3). Then, as rotation occurs, the ring turns and the holes in the ring block or allow light to strike the detector at regular intervals. The result is a pulse train similar to the Hall effect and reed switch transducers.

The prosthetic elbow is constructed using a New York elbow. When the forearm is raised or lowered the angle between the forearm and the humerus is changed and by properly mounting a slide potentiometer these angular changes can be detected. The potentiometer is mounted as shown in Figure 4.

DISCUSSION

The hand transducer provides the continuous output of Figure 5. The actual angle of opening can be calculated from the distance of the spread between the thumb and the index finger. This transducer worked very well, is very

reliable, and operates without affecting either the cosmetic appeal of the hand or it's mechanical operation.

The rotational measurement transducers yielded a pulse train such as that shown in Figure 6. The effective resolution of the transducer could be doubled by counting pulse edges rather just pulses. At present, the resolution of these three transducers is about 20 degrees but this is solely a function of the number of magnets. For increased accuracy a larger number of magnets should be mounted to the arm. Both the Hall effect and SDA transducers operated very reliably creating very few problems once they were operational but the reed switch has a major reliability problem. The reed switch consists of two metal reeds in a glass tube. This tube is easily broken, thereby creating a reliability problem.

The elbow slide potentiometer produces the continuous output of Figure 7. While this curve is somewhat nonlinear, the majority of the range of motion of the joint is in the linear region of the curve. The major problem lies in the starting threshold of the transducer. At 60 degrees this value is much too large. The starting angle is a function of the straight nature of the slide potentiometer. If a curved slide potentiometer were used then the curve could exaggerate the movement in the potentiometer at smaller angles, thereby reducing the starting threshold.

While the resolution of the analog transducers is much better than the present digital transducers, this does directly imply that the analog transducers are significantly better. For example, these transducers were designed for use with microprocessor based limb controllers. Since the digital transducers are already in the correct form for use in the microcomputer a certain amount of translational effort over analog transducers is saved. Total cost for all the transducers used is under 50 dollars.

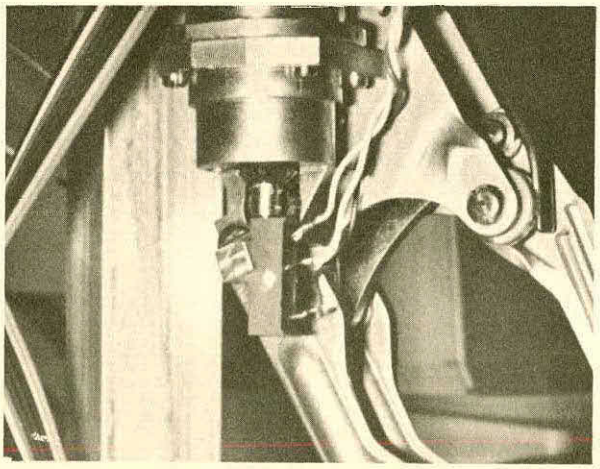

Fig. 1 Hand Transducer.

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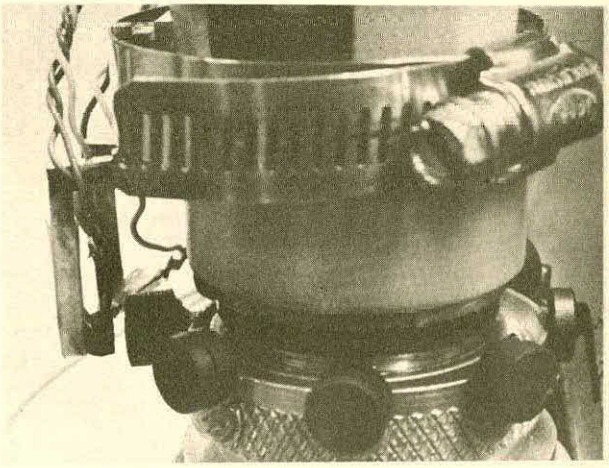

Fig. 2 Hall Effect and Reed Switch Rotational Transducers.

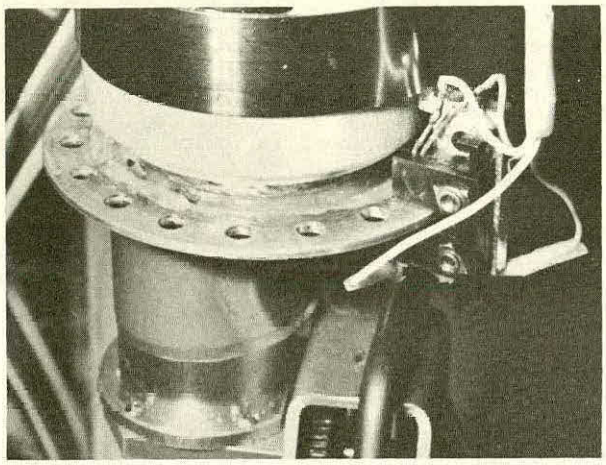

Fig. 3 SDA Rotational Transducer.

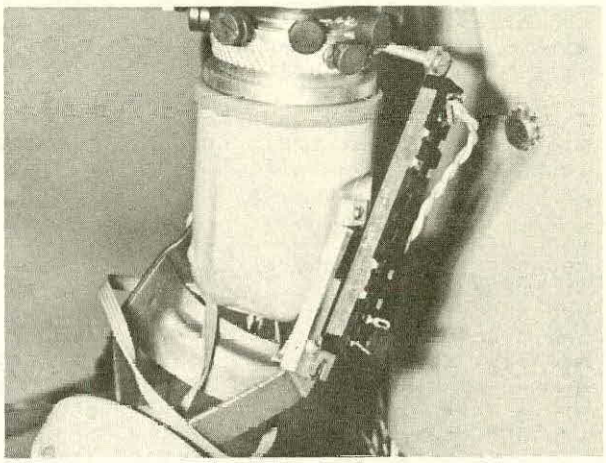

Fig. 4 Elbow Transducer.

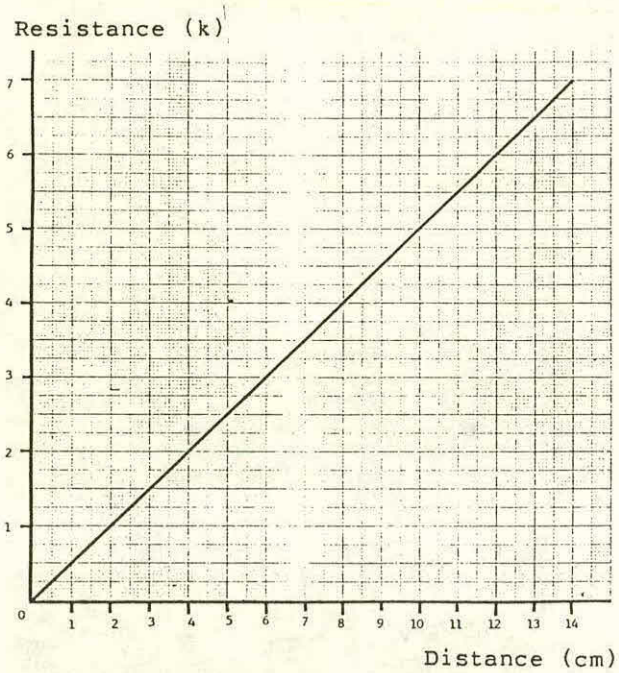

Fig. 5 Hand Transducer Output.

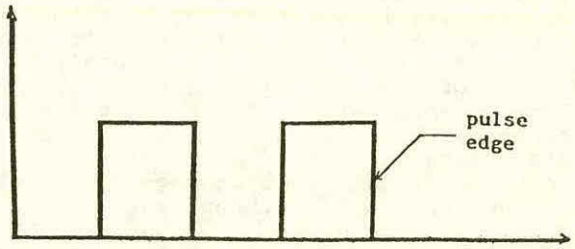

Fig. 6 Rotational Transducer Output.

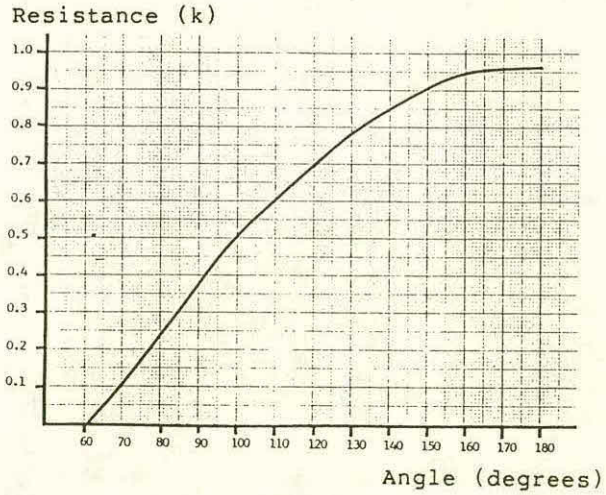

Fig. 7 Elbow Transducer Output.

Design, Development, and Evaluation of a New Body-Powered Arm Prosthesis

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ABSTRACT

Amputees, prosthetists, and rehabilitation professionals have indicated a need for an improved design of body-powered arm prostheses. The objective of our project is to design, develop, and evaluate a new body-powered arm prosthesis to meet this need. Our approach has been to develop and evaluate several systems which translate body movement into prosthesis activation. From these systems, the Tricep Power Capture (TPC) arm prosthesis was selected for further development. We have developed the TPC Prosthesis through engineering drawings, models, and analysis. Detail drawings have been released and prototype manufacturing is under way. We will begin testing the working prototype with amputees in April 1986.

INTRODUCTION

Prosthesis Development Associates (PDA) is a team of three Master's students in the Mechanical Engineering Department at Stanford University. We are currently involved in a nine month project sponsored by Children's Hospital at Stanford. The project goal is to design, develop and evaluate a body-powered arm prosthesis for below-elbow amputees.

In this paper we present a brief background of body-powered arm prostheses followed by the results of our design process. We state our goals in terms of a functional specification and present a new design which meets these goals. Finally, our manufacturing and testing plans are presented.

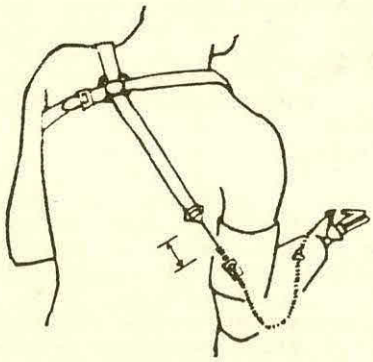

Figure 1. Standard Body-Powered Arm Prosthesis

BACKGROUND

The design of a body-powered arm prosthesis has not changed significantly since developments in the 1950's. The prostheses still employ aircraft technology using a shoulder harness and cable system for power and control.

It is the consensus of several leading prosthetists in the United States that there is a need for improved design of body-powered arm prostheses ([1,2,4]). To quantify this need, PDA investigated statistics on the U.S. amputee population.

There are 100,000 upper limb amputees in the United States. Of these amputees, 45,000 wear body-powered prostheses ([3]). These systems are very reliable and provide a range of functions that help the amputee to live independently. Only 5,000 people have chosen to wear externally (battery) powered prostheses. Most amputees feel the externally powered prostheses do not provide the functionality, reliability, or feedback capabilities of the body-powered prostheses.

The alarming figure, however, is that 50,000 amputees do not wear a prosthesis. Typically, this is because they feel that existing arm prostheses are uncomfortable, ugly or too expensive. The fact that half the amputee population chooses not to wear the existing prosthesis clearly indicates the need for improved design.

DESIGN PROCESS

Functional Specification

To identify and prioritize our design goals, PDA interviewed a number of amputees and rehabilitation professionals. Their responses, coupled with information from our literature and patent searches, were the basis for the following functional specification.

Essential Features. It is essential that:

1. The system is completely body-powered with no external power source with the exception of a very small long-life battery to operate a control system (if needed).
2. There is minimal restriction of body movement in the process of powering the prosthesis.
3. A grip force of at least five pounds be generated at the terminal device.
4. The system is 99 percent reliable; it must be operable ninety-nine out of one hundred days.
5. The response at the terminal device occurs within 0.1 seconds of actuation.

6. The sensory feedback capabilities of the system are better than or maintained at the same level as the present cable system. Sensory feedback is required to indicate the force applied at the terminal device and the amount of opening of the terminal device.
7. The system cost less than $2,000.00.
8. The weight of the power system is less than one pound.

Desirable Features. It is desirable that:

1. The prosthesis can be donned/doffed within 15 seconds.
2. The system is comfortable.
3. The system is aesthetically pleasing.
4. The basic operation can be learned in one month.
5. The system does not soil or interfere with clothing.
6. One force is capable of two separate functions such as rotating the wrist and opening the terminal device.
7. The maximum gripping force of the terminal device is variable between two and eight pounds.

Design Decision

During the autumn quarter, 1985, PDA generated many design possibilities. Each design was evaluated on the basis of the functional specification with weight factors representing the relative importance of each feature. The Tricep Power Capture arm prosthesis was found to have the highest ranking among the design alternatives. Is was, therefore, selected for further development.

TRICEP POWER CAPTURE PROSTHESIS

System Overview

The existing body-powered arm prosthesis achieves excellent functionality by providing independent movement of the terminal device and the other joints of the amputated arm. Independent movement is achieved, however, by powering the terminal device from a remote location (the opposite shoulder). The harness essentially inhibits the freedom of a large portion of the upper body in powering the hand.

Similar function is achieved without a harness in the TPC prothesis by providing coupled and uncoupled modes of operation. In the uncoupled mode, the user moves the elbow freely and no movement occurs at the terminal device. When activation of the terminal device is desired, the user taps the nudge switch; this latches the single axis hinge at a fixed angle. Tricep flexion and terminal device movement are now coupled. Hence, by extending his arm the user opens the terminal device. The nudge switch is tapped a second time to release the hinge allowing uncoupled movement of the elbow. This process is shown in Figure 2.

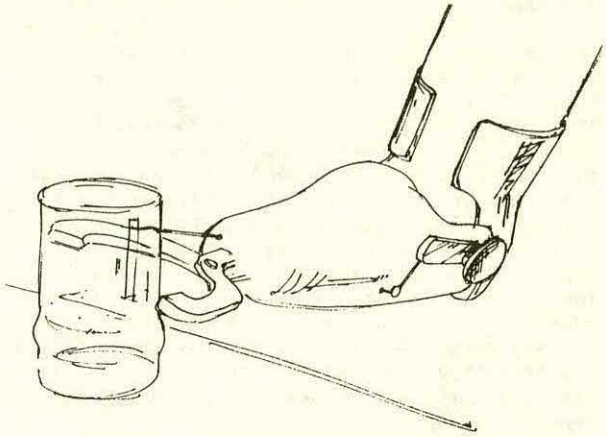

Figure 2. Tricep Power Capture Arm Prosthesis.

Manufacturing

In addition to the overall design concept, three of the seven sub-systems are custom designed. The first custom sub-system is the tricep cuff which suspends the prosthesis at the epicondyles of the humerus. This part will be vacuum formed over a model of the amputee's elbow. The single axis hinge is the second custom sub-system. It is designed to transmit power from tricep flexion to the control cable. This part is easily manufactured in a standard machine shop. Finally, the nudge switch was designed to latch and unlatch the hinge. This part is also easily manufactured.

Detail drawings of all non-standard parts as well as a system layout have been released to the Hosmar Dorrance Corporation for manufacturing. We expect to have a working prototype to begin testing by April 15, 1986.

Test Procedure

The two major objectives of our test procedure are to test and optimize design parameters and to compare the functionality and acceptability of the TPC prothesis to the existing prothesis. The design parameters we will test are the location of the nudge switch, the length of the hinge strap, and the effectiveness of the epicondyle suspension system. We will compare the functionality of the TPC design to the existing design using standard training exercises. Acceptability will be determined by direct amputee feedback. We are fortunate to have several below-elbow amputees who will assist in this testing process.

CONCLUSION

The Tricep Power Capture Prosthesis responds to the amputees request for improved design while

retaining the functionality of the existing body-powered arm prosthesis. Detail drawings have been released for manufacturing, and amputee testing will begin in April, 1986. The TCP Prosthesis final product will be presented at Stanford in June, 1986.

ACKNOWLEDGEMENT

We wish to thank both Maurice LeBlanc of the Children's Hospital at Stanford, and Felix Zajac of Stanford's Mechanical Engineering Design Division for their excellent advice and outstanding support of this project. We also wish to thank Idd Delp for illustrating the Tricep Power Capture Prosthesis.

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TECHNIQUES OF WEIGHT REDUCTION OF POLYPROPYLENE ANKLE FOOT ORTHOSIS

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ABSTRACT

Three techniques were utilized to analyze an existing ankle foot orthosis (AFO): mathematical modeling, finite element modeling, and experimental testing using strain gages. The data from the three techniques were compared and a modified AFO design incorporating variable AFO thickness resulted. An optimized, practical, lighter weight ankle foot orthosis was thus developed.

INTRODUCTION

Ankle Foot Orthoses (AFO's) are utilized by patients suffering from a wide range of deformities and diseases including cerebral palsy, muscular dystrophy, polio, and other muscle disorders. AFO's provide muscle control at the ankle and toe joints of the leg about the transverse, subtalar and verticle axes.

Inversion, eversion, dorsiflexion and plantar flexion are all disorders corrected by the AFO. A patient suffering from inversion walks on the outer portion of the foot. There is a medial deviation of the metatarsals and the foot is rotated along the subtalar axis. Eversion is the rotation of the foot along the subtalar axis in the opposite direction and involves a lateral deviation of the metatarsals causing the patient to walk on the inner portion of the foot. Dorsiflexion is due to paralysis of the gastrosolus calf muscles which elevates the toes causing most of the weight to be placed on the heel of the foot during gait. Plantar flexion is due to weak dorsiflexion muscles and results in a greater weight distribution on the back of the leg and the foot tends to remain in the plantar position.

Generally, there are three classes of AFO's prescribed to correct these problems. They are the Solid Ankle AFO, the Semi-Rigid AFO and the Posterior Leaf Spring (PLS) AFO. This report analyzes the load distribution during gait of existing PLS AFO's and then utilizes this information to develop an optimum light-weight ankle foot orthosis.

METHODOLOGY

Initially a list of ideal characteristics for an optimum AFO was developed. The brace must provide the rigidity needed for muscle control yet, the flexibility required for the lower leg throughout the gait cycle. The brace must be designed so that it can be readily manufactured and the overall cost must be kept to a minimum. Appearance and comfort are also important characteristics of an optimum AFO. However, the most important characteristic is the weight of the brace. The weight of the AFO is significant because any extra weight applied to an already weak leg will hinder, rather than facilitate, the patients ability to walk.

Reduction of the weight was achieved by changing the geometrical shape of the AFO and analyzing the loading conditions on the AFO. Three techniques were used to establish the new geometrical design parameters ultimately resulting in weight reduction. They are:

1. Mathematical Model
2. Finite Element Model
3. Experimental Model

The Mathematical model was developed to predict the design constraints for the overall AFO. Due to the fact that the final shape of the AFO is determined by the orthotist, specifically for each patient, the new design was limited to changing the geometry of the brace by varying the cross-sectional thickness. From data obtained at the Newington Children's Hospital Gait Lab, in Newington, CT it was found that the maximum load on the brace during walking occurs at the toe off position. To calculate the bending stress, the variable cross-section of the AFO from the heel to the strap had to be considered. This was taken into account by dividing the AFO into ten one inch segments vertically along the back of the AFO. Assuming each segment was of constant cross-section, the bending stresses on the brace were calculated for each segment.

Finite element analysis was used to determine maximum stresses by a computerized method. Finite elements, joined together by nodes, were used to approximate the shape of the AFO. Loads and constraints were applied to the model from predetermined experiments utilizing maximum deflection. The model was then submitted to a commercial program (ANSYS) and the stresses on the brace were computed.

The particular program used was the PC Linear ANSYS: this assumes that the stress is linearly proportional to the strain. The output was then extracted from the program and presented in a stress contour plot which gave a visual understanding of how the stresses are distributed throughout the AFO.

The experimental model consisted of two parts: 1) determination of the magnitude of stresses and 2) their directions in the AFO as a result of normal use by a patient. (The normal use

consists of a patient who uses the AFO while walking, sitting, bending, etc.) Brittle lacquer, which cracks at certain strain levels when loaded, was used to determine the directions of the stresses. Specifically, the lacquer used cracked at levels of strains greater than 500 micro-inches/inch. The directions of these stresses are located perpendicular to the cracks formed on the AFO.

The magnitude of the stresses during normal use were then determined using strain gage data. Strain gages of various types were mounted along directions perpendicular to these cracks at each one inch segment. A strain gage indicator and a strip chart recorder were used to document these results, and a plot of the strain level versus the distance along the vertical axis was developed as typically shown in Figure 1.

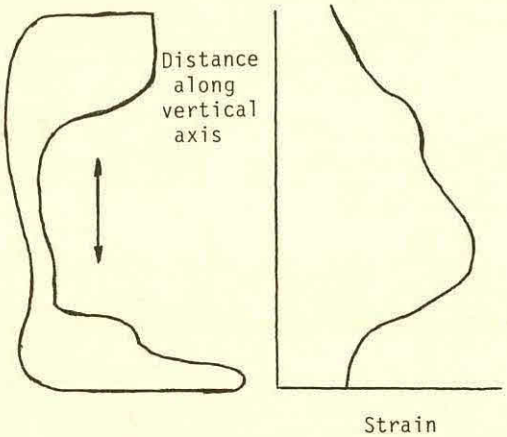

Figure 1
Strain Distribution along
Vertical Length Axis

RESULTS AND CONCLUSIONS

Presently the orthotic devices are vacuum formed using thermoplastic polypropylene sheet; hence the brace is relatively uniform in thickness. The overall goal of this project was to reduce the weight yet maintain the functional requirements of the PLS AFO. Regions of high and low stress on the AFO were defined utilizing the data obtained from the three models. This information was implemented to achieve an AFO with a uniform stress distribution during normal use. The experimental AFO was then surface cut in areas of low stress using a machining process to obtain a variable thickness AFO. This yielded a functional AFO with a weight reduction of approximately twenty-seven percent as compared to a monimal uniform thickness AFO.

To be implemented on a larger scale by orthotists, the current sheet forming technology could be maintained by adding a secondary milling operation to the manufacturing technique. An alternative to sheet forming would be to use CAD/CAM controlled equipment in which the operator could program the computer for a patient's specific geometry data and computerized milling could be used to form the entire AFO. Problems arise in establishing the three-dimensional coordinates that define the irregular geometry of the AFO.

Another alternative for establishing a variable thickness AFO involved using composite materials and a method of hand lay-up. The hand lay-up involves placing pre-pregnated layers of the composite material over a mold of the patient's leg and using more layers to vary the thickness in high stress areas. The feasability of implementing these alternative manufacturing methods has yet to be investigated.

LITERATURE SEARCH

The following literature search was done at the onset of the project to obtain a general understanding of orthotics and the biomechanics of the leg. The literature search confirmed that this type of analysis had not previously been documented.

The Lower Limb Orthotics Manual (1) provided general background information of orthotics and the medical implications involved. Masafumi and Yonemoto (2) utilize pressure sensitive film to analyze the pressure distribution along the bottom of the foot during gait while Hatze (3) Yamashita and Katoh (4), and Schnider and Chau (5) discuss the ground reaction forces involved during walking. Force versus deflection data of ground reaction forces during gait is presented through the use of a mechanical model by Siegler, Seliktar and Hyman (6). The kinematics of the gait cycle is discussed by Boccardi, Pedolti, Rodano and Santambrogio (7) and by Zarrugh, Sommer and Miller (8). Acceleration data of the entire body during gait is presented by Kaurento and Hellen (9) while the dynamics at heel strike is discussed by Simon, Paul, Mansour, Munro, Aberrutny and Rodin (10). Huiskes and Chao (11) discuss the development of the biomedical applications of the Finite Element Modeling. Proctor and Paul (12) and Winter (13) outline the movement of the ankle joint while Mena, Mansour and Simon (14) discuss the effect of movements occurring at this joint during the swing phase of gait.

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REHABILITATION ENGINEERING ROUNDS AT THE UNIVERSITY OF MICHIGAN

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ABSTRACT

"Rehabilitation Engineering Rounds" has been established as a vehicle for reviewing current clinical cases and new referrals. It provides a time for discussion, education of students and staff, and administrative issues. Discussions are related to patients, their therapies and possible solutions to their problems. Case summaries from a typical engineering round are presented.

INTRODUCTION

The Rehabilitation Engineering (RE) Division at the University of Michigan is within the Department of Physical Medicine and Rehabilitation (PM&R). RE provides clinical services to both inpatients and outpatients on a referral basis. A majority of the referrals come from three PM&R inpatient services: General Adult, Pediatric, and Spinal Cord Injury. The Spinal Cord Injury Service is a part of a federally funded Interdepartmental Spinal Cord Injury Program at the University of Michigan. In addition to these sources, referrals are received from throughout the University of Michigan Hospital System as well as other hospitals, outside health professionals, school systems, and the community. The Division currently has five full time staff, five part time staff, and another 5-10 students per term working on clinical service, research projects, course projects, and independent studies.

RE also has an active research and education program. Current research areas include functional electrical stimulation, computerized systems for the handicapped, robotics, and others. Undergraduate and graduate students come to the division from the Departments of Bioengineering, Electrical Engineering and Computer Science, Mechanical Engineering and Applied Mechanics, Engineering Science, Psychology, etc.

A team approach is used at our center so that the rehabilitation engineers work closely with other health care professionals and PM&R staff which include physicians, psychologists, physical therapists, occupational therapists, speech language pathologists, nurses, social workers, vocational counselors, and teachers. This structure provides a model setting for the clinical practice of rehabilitation engineering.

Once each week RE staff meet for "Rehabilitation Engineering Rounds". The purpose of this meeting is to review current cases, provide a chance for discussion, serve as an educational tool for our students, and handle administrative issues. The purpose of this paper is to provide 10 brief case summaries from Rehabilitation Engineering Rounds for a typical week. It is our intention that this provide insight into rehabilitation engineering activities at our center and aid others in providing structured rehabilitation engineering services.

CASE HISTORIES

Case 1: Inpatient, Spinal Cord Injury Service

P.A. is a seventeen-year-old young lady with a spinal cord injury at the C1-C2 level. This is an incomplete injury and she has some motor output spared. She was injured on December 31, 1984 and was transferred to the University of Michigan Medical Center in February of 1985. She has always had fair to good control of her left ankle, including plantar and dorsiflexion and medial and lateral rotation. She also has fairly good strength in her left quadriceps. Recently, RE has modified a Rolls Arrow electric wheelchair so that P.A. can drive the chair with her foot. Forward and reverse motions are controlled with plantar and dorsiflexion of the foot; left and right motions are controlled with lateral and medial rotation of her left ankle. Her left foot is strapped into this foot controller.

At this time, P.A. has driven this system on three separate occasions. One night she went to downtown Ann Arbor to see a movie. She has been able to control the wheelchair remarkably well. A panic switch must still be included in case P.A. loses control. With the panic switch, she can turn the chair off at any time. We have also been experimenting with the speed controls on the chair to set it up for the best possible response related to the positioning of her foot. P.A. has no sensory feedback from her lower extremities. The foot control must, therefore, be mounted so that she can see it. As mentioned, P.A. has had very good control of the chair; however, she has been very tentative about driving it. She is nervous, especially around other people. I am confident she will be able to overcome her fears and drive the chair with the foot control. -DJK

Case 2: Inpatient, Adult Rehabilitation Service

E.K. is a fifty-year-old man with C5-C6 quadriplegia from an episode of polio at age 15. He requires full mechanical ventilation with some oxygen assist. He controls the joystick of his Rolls electric wheelchair and the recliner switch with his left hand. The recliner's toggle switch is mounted near the joystick of the wheelchair. RE has added brackets for an E-sized oxygen cylinder and fittings so that this oxygen can supply both a talking tracheostomy (1) system and the oxygen (30%) needed for ventilation. This system was mounted so that it does not interfere with the reclining mechanism of the chair. The talking tracheostomy branch of the pneumatic circuit

includes a solenoid valve so that the air supply can be turned on and off by E.K. He actuates the solenoid valve with a "versa"[1] switch and electronic latching relay. Each activation of the switch toggles the solenoid valve from "on" to "off". The ventilator requires a constant flow of two liters of oxygen per minute and the talking tracheostomy requires 5 liters per minute. The oxygen cylinder needs to be replaced on a daily basis (indicating considerable use by the patient). -DJK

1 A "versa" switch is a custom designed switch developed and used by the RE Divison.

Case 3: Inpatient, Adult Rehabilitation Service

V.B. is a seventy-five-year-old lady with a spinal cord injury at the C5-C6 level. She will be discharged to her home with 24 hour nursing care. Her house is quite large, and as she will be using a manual wheelchair, she will require an alarm system to notify her nurse if she needs assistance. A small portable alarm was manufactured in RE and given to her prior to her discharge. This alarm consists of a toggle switch, a piezo electric buzzer, and a 9 volt battery all packaged in one small box that can be mounted either on her wheelchair or her bed. It is expected that the high pitched tone of the piezo buzzer can be heard throughout the house. Follow up on this case will be needed. -RLK

Case 4: New Inpatient, Pediatric Rehabilitation

J.C. is a seventeen-year-old young man with a spinal cord injury at the C5-C6 level. He has been injured for approximately five years and was admitted to the Pediatric Rehabilitation Service to assess his ability to live independently. J.C. has been through several rehabilitation services throughout the state. He came to our program from a nursing home in the Upper Peninsula. He would like to live on an independent basis in an apartment in the Ann Arbor area. He is planning to take a high school graduation equivalency examination and start college classes next fall. He has not attended school since eighth grade. Currently, he is a ward of the state and his legal guardian lives in the town of Mesquite, Michigan in the Upper Peninsula. He will be seen by RE to assess his potential for using environmental control systems and living independently in an apartment. Additionally, we plan to introduce him to the use of a computer and work towards the goal of attaining independence for him at a computer work station. Unfortunately, the chance of obtaining funding on technological devices for his use is poor for the immediate future so that extensive training is not warranted at this time. I will discuss this issue with social work and vocational rehabilitation. -RLK

Case 5: Outpatient Referral, Dr. JAL and Michigan Vocational Rehabilitation

P.C. is a young man with a primary diagnosis of arthrogryposis. This is a progressive disease which affects the joints. For him, it has affected only the upper extremity joints. It affects the distal joints greater than the proximal. Consequently, he has limited to zero range of motion in his fingers, wrist, and elbow. His shoulder range of motion is fair. P.C. will be graduating soon in the field of Industrial and Operations Engineering from the University of Michigan and will be looking for work. He is interested in the design of a computer work station that is efficient as well as comfortable for him. An appointment is scheduled for him next at which time an update on this patient will be presented. -RLK

Case 6: Outpatient Referral, Rehabilitation Psychology, Dr. NLK

D.M. is a sixty-year-old lady who lives with a personal care attendant in her home in Ann Arbor. She suffered a traumatic brain injury about eighteen months ago and underwent rehabilitation here in the University of Michigan PM&R Department. The rehabilitation team hopes that she can live more independently. RE has received a referral to evaluate the possibility of adding some safety devices in her home for when the care attendant is not present. Specifically, the psychologist and occupational therapists are quite concerned that Ms. Mayerson will not be able to respond appropriately should a stressful situation occur. They are interested in devices such as a timer on the stove so that it will only run for a certain amount of time, a fire alarm system that will automatically call an emergency number when it is activated, an alarm system that would make a telephone call when Ms. Mayerson falls, and possibly a burglar alarm system. A visit to her home is planned in the near future. Discussions with the occupational therapists and psychologist regarding her needs will follow the home site visit. -RLK

Case 7: Inpatient, Pediatric Rehabilitation

W.C. is an eighteen-month-old young boy who was in an automobile accident and suffered a spinal cord injury at the C4-C5 level. He is currently dependent on a ventilator but the medical team hopes that he can be weaned in the near future. He is being seen in conjunction with both the Speech Pathology and Occupational Therapy Divisions. A simple environmental control unit (ECU) has been set up for him using a BSR Scanning System. We plan to change to a custom ECU that was made in RE several years ago for a different patient (2). This system is much simpler and has three channels plus an "off" and operates on a single switch. Each channel turns on automatically when it is selected. I also plan to work more with Speech on some simple communication ideas. RE has recently developed a game that can be activated with a single switch to be used with very young children, and we will try that first with W.C. -RLK

Case 8: Inpatient, Spinal Cord Injury Service

M.G. is a new spinal cord injury patient at the C5-C6 level. This past week, we assisted in the preparation of an electric wheelchair for him

to drive. RE moved a joystick from the left side to the right side of the chair. It is planned to introduce an environmental control system and install one in his room so that he can use it on a daily basis. -DJK

Case 9: Inpatient, Pulmonary Intensive Care

A referral was received requesting that RE set up a temporary communication system for K.M. He is a seventy-five-year-old man who had a tracheotomy with the resulting temporary loss of his voice. The medical team feels that he will eventually get his voice back; therefore, only a temporary system is necessary. When I responded to the referral, K.M.'s son was in the hospital room with a Commodore 64 computer. He was hoping to have his father learn how to use this system. I discussed the options of different communication systems with his son and explained that RE would be able to provide one on a short term basis. At this particular time, K.M. was not alert; therefore, I did not spend any time with him to determine an appropriate system. -RLK

Case 10: Inpatient, Adult Rehabilitation

K.G. is a young man on the Rehabilitation Service for an evaluation of his needs. He has a unique form of muscular dystrophy which usually affects only the facial muscles (FSH muscular dystrophy). K.G. has a very severe form of this muscular dystrophy as it has also affected his extremities. He is still able to ambulate on a limited basis but very slowly. He is being evaluated for electric mobility, adaptations to his home to make him independent, and his needs at work. K.G works as a computer programmer. We have discussed alternate communication systems that could assist him in programming. -MAD

DISCUSSION

The cases presented here reflect the activities of the RE Division for one week's time and do not necessarily represent the mix of services rendered throughout the year. The full range of services provided include mobility, seating and postural support, communication, manipulation aids, environmental control and other applications of technology to problems of the handicapped. A strong effort is made to use commercially available equipment and minimize custom development in these areas.

In addition to Rehabilitation Engineering Rounds, RE staff regularly participate in chartrounds, staffings, and family meetings for inpatients on the PM&R service. These team meetings, involving PM&R staff and other health care professionals, assist the various disciplines in the coordination of therapies provide a forum for problem solving, and are concerned with all aspects of the patient's welfare. As needed, team evaluations of outpatients are held and can include family members, health care professionals, teachers and counselors.

Rehabilitation Engineering Rounds has proved to be a viable means for coordinating patient services which stimulates discussion and leads to effective development of solutions to our patients' problems. At the same time, it has provided a vehicle to help organize the week's activities and a formal mechanism for delegating responsibilities. These meetings help ensure quality care for patients by more efficiently utilizing the RE Division's clinical time and expertise

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REHABILITATION ENGINEERING SERVICE DELIVERY; TWO CASE HISTORIES REGARDING LOW BACK PAIN

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ABSTRACT

Rehabilitation Technology Services (RTS) is a private for-profit company affiliated with the Vermont Rehabilitation Engineering Center (VREC). RTS provides a wide range of engineering services. Since the VREC's research priority is low back pain, many of RTS's service activities involve low back pain in the workplace. This paper presents a short description of RTS's services and two case histories. The first history involves a woman with M.S. and a fractured L5 vertebra who required seating that would support her lumbar spine. A commercially available office chair was modified and installed in her Amigo wheelchair. The second history involves modifying a cart used to transport computer tapes. Problems were encountered in moving the cart weighing up to 444 pounds up and down steep ramps. An Amigo wheelchair drive head and steering handle were used to modify the cart so as to make it powered.

INTRODUCTION

The provision of rehabilitation technology services has come into sharp focus during the last few years. More and more people are interested in insuring the availability of services to those people who can benefit most from it. Rehabilitation Technology Services (RTS) is a private for-profit company started in 1984 that is now affiliated with the Vermont Rehabilitation Engineering Center. RTS was begun in response to the need to provide quality rehabilitation engineering services to those people who could benefit from it through state agencies, hospitals, industry and educational institutions. Areas of service include, but have not necessarily been limited to, worksite evaluations and modifications, wheelchair/mobility evaluations and modifications, seating/postural systems, communication/computer systems and architectural accessibility.
These services have been provided on basically a freelance basis. They are funded through a variety of mechanisms and agencies, including:

- State Vocational Rehabilitation agencies
- Private rehabilitation agencies
- Insurance (Medical, Worker's Compensation, Liability)
- Medicare/Medicaid
- Special Education
- Industry
- Private pay

A few contracts have been developed to provide these types of services on a regular basis through various agencies, i.e. an acute medical rehabilitation facility, a job placement service and a therapy resource agency.

Through the affiliation with the Vermont Rehabilitation Engineering Center, addressing the problem of low back pain in the workplace has become an important part of RTS's activities. Low back pain is mankind's most common musculoskeletal complaint, the single greatest source of compensation payments, and is the second most common cause of work loss. As such, it represents an area where rehabilitation engineering services can make a dramatic impact.

Two case histories have been chosen to demonstrate the benefits of rehabilitation engineering in providing adaptations to the workplace for people with low back pain.

CASE HISTORY I

Janet is a 42 year old woman with multiple sclerosis who has been working for a local telephone company for 23 years. Due to a fall in the summer of 1984 she suffered a fracture of her fifth vertebra. The normal course of treatment for a fracture of this type would have been about 3 months of bedrest. However, bedrest was contraindicated due to her multiple sclerosis. Her doctor felt she would have lost a great deal of strength had she been confined to bed for that long a period of time. Therefore, Janet continued to work but experienced a great deal of back pain. Tylenol III was originally prescribed as a pain killer but soon proved to be ineffective. Intra-muscular injections of Talwin (a strong narcotic) were then prescribed. Janet had reached the point of needing two injections per day or 14 per week. She usually worked four days per week.

Janet works in the engineering department of the telephone company. Her job requires her to work at a desk all day. She uses an Amigo wheelchair for mobility. Because of difficulties in transferring, she utilizes the seat of the Amigo at her desk. Because the molded fiberglass seat of the Amigo offers little support to the lumbar spine it was suggested that Janet use a chair that provided more support.

A number of office chairs are available that provide good lumbar support while performing tasks at a desk. It was recommended that Janet purchase a Hermann Miller office chair. The problem for the rehabilitation engineer was how to interface the office chair with the Amigo wheelchair.

PROBLEM SOLUTION

The office chair was removed from its base. The molded fiberglass seat was removed from its pedestal on the Amigo. The plate atop the pedestal was big enough to accommodate the office chair. A spacer was fabricated out of maple wood

(Figure 1) that tilted the chair such that the seat pan was horizontal. The spacer had to be big enough for the bottom of the armrests of the chair to clear the pedestal plate. Holes were then drilled in the spacer and the pedestal plate to match those of the chair. Four 1/4-20 bolts were used to fasten the chair and spacer to the pedestal.

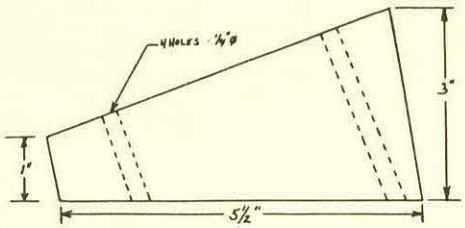

FIGURE 1. Side view of spacer

The completed chair was delivered in September, 1985 about two months after the design was finalized. Since using the chair Janet has had perfect attendance at work for the first time since beginning work 23 years ago. Her pain medication has been reduced from 14 injections of Talwin per week to 5-6 per week and continues to be reduced. The only difficulty with the adapted chair has been due to the increase in height of the chair. The use of the spacer and the greater thickness of the office chair seat pan made the seat too high for Janet to reach the floor with her feet. She now uses a footrest under her desk to compensate for the additional height.

Although the Hermann Miller chair retails for almost $600, Janet was able to buy a used one from an office furniture supplier for about half price. The total bill for the rehabilitation engineering services came to $150 which included an initial evaluation, fabrication time and installation. Thus the total cost of the adaptation was approximately $450. Janet was later reimbursed by her medical insurance carrier for the chair and modifications.

CASE HISTORY II

A large insurance company in Massachusetts uses "Tape Seal Trucks" to transport computer tapes from one department to another. The Tape Seal Truck is a utility type push cart. The cart weighs 124 pounds empty. When loaded with 126 tapes, each weighing 2½ pounds, the loaded cart weighs a total of 444 pounds.

When the insurance company first contacted RTS, two older women were responsible for moving tapes with the carts. Their jobs involved moving tapes around the Computer Systems Department of the company as well as moving backup tapes from the Systems Department to a vault in another part of the building. Because the wiring and cables for the computer are run beneath an elevated floor, there are a number of ramps that must be negotiated in order to move the computer carts into and out of the Systems Department. These ramps ranged from 3 to 8 feet in length and from 10° to 14° in inclination. As shown in Figure 2, this means 107 pounds of force must be applied to the cart just to keep it stationary on the ramp. Somewhat more force must be applied to push it up the ramp and overcome friction. Besides merely pushing the cart up the ramp it must be maneuvered around tight corners and over floor switches to open electrically powered doors. One of the women lost control of a cart pushing it up a ramp and was pushed backwards into a wall. She suffered a back injury which kept her out of work for a few days. The company then decided to seek an alternative to the hand carts in order to make it easier for their present employees to handle the cart and to prevent any more accidents which might prove to be more serious than the one they had already experienced. The rehabilitation coordinator contacted RTS to investigate the possibilities of modifying the worksite.

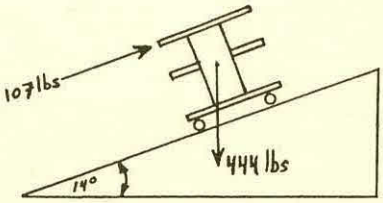

FIGURE 2. Force needed to hold cart stationary

PROBLEM SOLUTION

The original idea for modifying the workplace was to purchase a powered cart that could transport the tapes. The manufacturer of the original cart was called but had no knowledge of a commercially available powered cart. A more extensive review of commercially available equipment revealed that the smallest powered materials handling cart was either a powered lift truck or a powered pallet truck with a capacity of not less than 2,000 pounds.

It was then decided to modify the present tape cart by attaching a power unit to it. An Amigo wheelchair drive head and steering handle were used in order to avoid designing and building the power unit from scratch. The cart was modified by building two supporting frames out of 1 1/2" X 1 1/2" X 1/8" angle iron. A special bushing was fabricated to support the steering handle in the upper frame. The bushing normally used to support the drive head on a wheelchair was purchased from Amigo and welded to the lower frame. The frames were attached to the cart such that the steering handle and drive head were 6" from the back of the cart. The battery was mounted in its case to the center of the lowest shelf. (Figure 3)

The cart is operated the same way an Amigo wheelchair is operated. Steering the cart is similiar to steering a pallet truck, i.e. turning the handle to the right steers the cart to the left. The battery charger is used in the same way as the wheelchair. An "owner's manual" was supplied with the cart which included operating instructions as well as warranties and sources of components used in the modification.

While building the cart an interesting problem arose that needed to be addressed. As the cart was meant to move magnetic computer tapes there

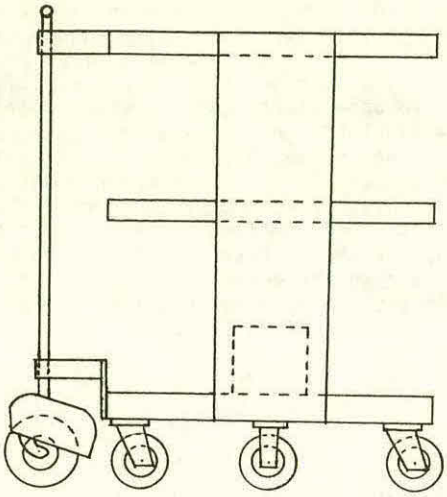

FIGURE 3. Side view of modified cart

was some concern that the magnetics used in the drive motor might affect the data on the tapes. Research was done to determine what effect, if any, the motor might have on the tapes. It was found that the magnets used in the motor were not large enough or close enough to the tapes to cause any loss of data. This information became part of the owner's manual.

The cost of the materials and components for the modification came to $806. Fabrication labor costs came to $250.

The cart has proven to be extremely effective in transporting the tapes over long distances and up and down ramps. There is some discussion about purchasing more powered carts for transporting tapes as well as other uses.

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A TEAM APPROACH TO SEATING

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ABSTRACT

This paper describes a successful seating team, the roles and accountability of its members, and how they contribute to the seating team as a whole. The way in which this team develops its programme, problem solves in the clinical setting and assures that the quality of the devices it produces is discussed.

INTRODUCTION

A team, as defined by Richard Beckhard(1), is a group with a specific task or tasks, the accomplishment of which requires interdependent or collaborative efforts of its members. At the Hugh MacMillan Medical Centre, the seating team has evolved over the last 16 years to fulfill this definition. This multidisciplinary clinical team examines the client's orthopaedic and lifestyle needs to determine the various details of the seating device. A prescription is generated as a result of these decisions. Any deviation from this prescription, not initially anticipated in clinic, must be approved by the team before it is amended. Our team approach to seating is described in this paper.

THE CLINICAL TEAM

The Seating Programme began in 1970 with 2 employees, who designed and constructed 22 special seating devices. The programme has evolved over the past 16 years to one with a staff of 2 therapists, 1 clinician and 6 technicians who complete up to 950 work orders in one year. Through experience and a highly motivated staff, this service clinic has developed successfully into a full cost recovery programme.

The clinical team consists of the patient, the caregiver or parent, an orthopaedic surgeon, seating therapist and technician. Others who may be involved are the client's teacher, speech pathologist, and physical or occupational therapist.

Pre-clinic screening of attendees involves using a standardized form that collects information regarding client problems and needs, together with environmental, communication and transportation considerations. This information is presented to the doctor and technician by the therapist. The client and attendees are then seen in clinic.

The doctor, an orthopaedic surgeon, who brings expertise regarding deformity and contracture, physically examines the client. If needed, x-rays are taken. Inoperable contractures are assessed with the view to seating the client as comfortably and functionally as possible.

The therapist assesses the motor age of the client, pathological reflexes, the degree, if any, of tissue trauma and the level of functional independence.

The technician measures the client, recording all pertinent data relative to the types of seating components, special instructions, and equipment to be included.

PROBLEM SOLVING IN THE CLINICAL SETTING

Following the clinical assessment, the team discusses ways in which the gathered information can be translated into the production of a mobile seating unit which best suits the client's needs. This is accomplished by prioritizing these needs.

Considerations include:

Security - Once the client is made to feel safe without the need to brace or protect himself, he can focus his attention on the world around him.

Comfort - This is determined by examining how long a client is likely to sit in his chair each day; if a periodic change of chair position is required; how much soft tissue is present; and by physical manipulation and positioning, how to accomplish muscular relaxation. In this way we can determine what is required in order to make the seating as comfortable as possible, and enhance user acceptance of the device.

Orthopedics - Considerations of contractures, their effect on seating prescriptions and any possible delays due to surgical correction are the responsibility of the orthopedic surgeon.

Function - Careful examination of client/caregiver needs in order of importance and appropriateness influence decision making of the team. Requirements for the client include that it be cosmetically pleasing. Considerations for the caregiver include feeding, client transferring, and size of seating unit. For the therapists, task accomplishments and independence are important. For the teachers and parents, communication, learning aid accommodation, peer group and sibling interaction, and behaviour management are of importance.

From this, an estimate of cost is prepared and presented. The client and parent/caregiver are consulted on what, if any, referrals will be made on their behalf. They are informed about what mobile seating unit will be prescribed and an estimate of how much this will cost. They are then directed to Patient Services which will explore funding resources.

Inter-professional behavior asks that members act together by placing the needs of the team ahead of those of individual professions(2). Accepting that they respect both themselves and others as having the knowledge, understanding, skills and equal rights to participate in the problem-solving process, the clinical team successfully makes its decisions.

SERVICE DELIVERY PROCESS

When the appropriate seating device is ready, the client and parent/caregiver are called for a fitting session with the seating therapist and technician. Again, others who may attend are the client's teacher, occupational and physical therapists, social workers and anyone who may have valuable input into the client's quality of life. At these sessions, clients sometimes do not display typical behaviour patterns. The question of comfort over extended periods of time is again often discussed.

If any doubts exist concerning the seating device's suitability, the client is encouraged to take the modified but unfinished seating device home for a trial period of up to two weeks. This is a very valuable step in the successful seating process. If more modifications are required, a follow-up session is scheduled. Before final covering of the device, the seating therapist is responsible for ensuring that it's function and purpose are the best that can be achieved.

TEAM DEVELOPMENT AND ACCOUNTABILITY

Administration encourages team development by providing time outside the work context so that roles can be explored, negotiated, and defined.

The clinical team meets for an hour every week to discuss client referrals, research and development, new seating products, and ideas and related philosophies.

A second technical team functions within the seating programme and is comprised of seating therapists and technicians. This team has the responsibility to ensure the quality and safety of the seating device, and that it fulfills the requirements of the prescription. Each technician is responsible for the quality of his work. The shop supervisor will then check the seating device for workmanship and safety prior to the therapist checking the seat against the original prescription. Once the seat is covered, it is again checked by the deputy coordinator of the programme. A quality assurance form is filled in and signed for each new seating unit completed to attest to the fact that the client or caregiver has been instructed on the safety and function of the device and/or a maintenance manual has been issued and explained.

More technically oriented, this group meets twice a month for 1 hour to discuss production problems and solutions, new manufacturing techniques, new products and training schedules.

The team is headed by a coordinator who is responsible to the Director of Therapies for clinical aspects of the programme, and to the Director of Rehabilitation Engineering for the technical and financial aspects. The orthopaedic consultants are accountable to the Medical Director of the Centre. Team members' roles and responsibilities over the years have been developed under the guidance of the Rehabilitation Engineering administration group.

CONCLUSION

Effectiveness implies that the greatest part of the energy of the group is focussed on accomplishing the tasks of the group(3). We strive to produce a functional, safe and comfortable product which is as cosmetically pleasing as possible, and which meets the special needs of each client and his family.

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A SERVICE DELIVERY MODEL FOR THE PROVISION OF SPECIALIZED SEATING

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ABSTRACT

Since 1980, Canadian Posture and Seating Centre (CPSC) has been exploring the needs, methods and problems involved in the provision of specialized seating. The importance of carry-over of services and information, from the specialized seating team to those involved with the clients ongoing care, has been recognized. CPSC has directed its efforts toward assisting in the development of expertise within the local service provision team.

New clinics have been started in communities where specialized seating services were not previously available. This has been accomplished through the development of a training program designed to facilitate the start-up or expansion of postural seating clinics. Various service delivery models have been explored. This paper is a brief outline of the models investigated and adopted by CPSC and a discussion of how these have been applied to the seating clinic start up process.

SEATING TEAM STAFFING

A frequently used model for seating clinic service delivery is a multi-disciplinary team consisting of a doctor, preferably an orthopeadic or physical medicine specialist, knowledgeable in the field of specialized seating; a therapist, either a physiotherapist or occupational therapist; and a seating technician. To date no formal training requirements have been specified for a "seating technician". In some settings this individual is an orthotist.

The physician is responsible for the medical referral for specialized seating and for indicating factors which might influence the design and fabrication of the seating system. The therapist assesses the needs of the individual and works with the technician to develop the design specification. The technician is also responsible for the fabrication and/or assembly and installation of the new equipment. For existing equipment, the therapist and technician must identify the need for changes to the device and refer the client back to the physician when required.

CPSC has adopted this model with the following provisio: It has been found that few physicians caring for adult and geriatric populations have had previous exposure to specialized seating. Many adulcs and elderly persons needing specialized seating are not under the care of a specialist. Consequently it becomes the responsibility of the seating clinic therapist and technician to inform and educate the client and his/her family, the physician and other care providers about the relevant aspects of specialized seating and to seek referral for further medical consultation if needed. Attendance of direct care providers and family members at the assessment and equipment selection sessions is encouraged. If the physician is unable to attend, the assessment and equipment specification information should be forwarded to the physician and written approval requested. This practice has been effective in providing local care providers with the information and tools needed for decision making.

Most importantly, the seating team must be a closely cooperating group whose members possess the skills needed to do a thorough neuromotor, orthopeadic, and ADL assessment as well as evaluate and work with available technical options. During the five day training program conducted by CPSC a structured procedure which will facilitate group decision making is practiced in a hands on, clinical environment.

ESSENTIAL COMPONENTS OF CLINICAL SERVICE

Dr. David Symmington, in a lecture on technology transfer, (London, Ontario, 1984), outlined eight aspects of service delivery to be considered when providing adaptive equipment and devices for disabled individuals. The model is particularly relevant to the provision of specialized seating.

The eight aspects of service delivery suggested by Dr. Symmington include:

1) Education
2) Equipment selection
3) Sourcing of Funding
4) Assessment, Equipment Specification and Fitting
5) Instruction for Equipment Use
6) Equipment Adjustment
7) Equipment Maintenance
8) Follow-up

1) Education

Education includes both the ongoing training and continued upgrading of the seating team and the education of both primary and secondary users.

The CPSC start-up program is done in the local community. The program includes an in-service presentation which provides an opportunity for physicians, therapists, nursing staff, funding agencies and family members to become informed about the new seating service offered in their community.

The training sessions for the seating team include both classroom instruction as well as supervised clinical experience. A training manual is supplied to each team as a training guide. The manual also suggests references for further reading.

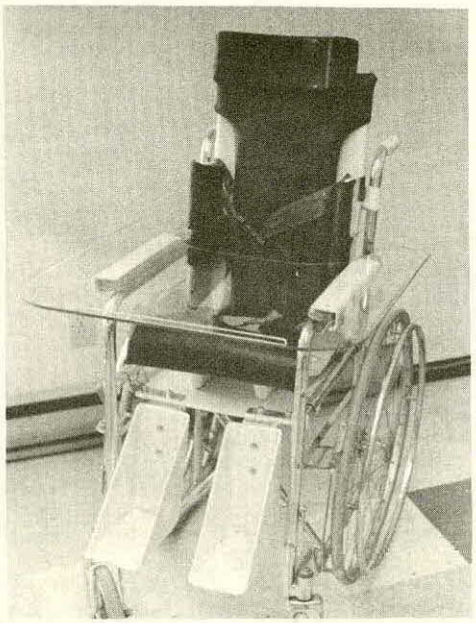
Much of the training program consists of supervised clinical experience. Up to eight clients are assessed for specialized seating needs. Assessment and

equipment specification forms are completed. Through this process the team becomes familiar with various problem solving approaches and a variety of seating components.

The need for continuing education is emphasized during the start-up program. Information sharing, through conferences and informal, ongoing networking between those involved in specialized seating is suggested. CPSC has established a library of literature on current approaches, technical options and research in the area of seating and positioning.

2) Equipment Selection

Because each client's seating needs are individual, CPSC recommends that the seating team have a variety of seating devices available for trial when doing an assessment. At the time of clinic start-up, CPSC provides a selection of modular components, moulded seating raw materials and a range of accessories and other postural aids. With this battery of equipment, the clinic can solve many seating problems. During the start-up program, the seating team becomes familiar with appropriate applications for modular and moulded types of seating by using the equipment to mock-up complete systems for several clients. Attention is also given to the merits of other seating products which may provide optimal solutions for certain seating problems. In general, familiarity with a large assortment of technical options from various sources will expand the capabilities of the seating team.



New seating teams use standard modular and/or molded components to mock-up complete seating systems for eight clients during the training program.

3) Sourcing of Funding

The importance of establishing a funding base is emphasized at the time of clinic start-up. It is recommended that the clinic become directly involved in pursuing funding for specialized seating. This can be done by forming an association with a funding "clearing house" such as the March of Dimes or by including a person with experience in locating funding, such as a social worker, to act as a consultant to the team. If the clinic is providing services to institutions where there is a social services staff, often they will agree to assist in finding the necessary funding. The inservice presentation during the first day of the training program provides an opportunity to bring all funding sources together for an introduction to the seating clinic program.

Standardized data collection forms for assessment, equipment specification, and physician referral, developed by CPSC are helpful when pursuing funding. These forms are used to document all necessary information including assessment findings, equipment specification details, itemized costing of the equipment, and a complete record of personnel recommending the special seating device.

4) Assessment, Equipment Specifications and Fitting

Assessment, equipment specification and fitting are the most visible activities of the seating team. During the training program, up to eight clients are assessed and if appropriate, various seating equipment alternatives are tried. CPSC provides instruction in the use of the standardized assessment form. The assessment includes consideration of neuromotor and orthopeadic status, activities of daily living such as self care, home and work environment factors, and finally an exploration of the individual's seating status and any specialized seating used, past or present. From this information, specific problems and goals can be identified. It is important at the time of assessment that the client, care providers and members of the seating team all understand and accept any compromises expected when a solution is implemented.

Using a battery of in-house seating equipment, trial seating solutions can be mocked up. When the team is satisfied that an optimal solution has been found, the exact parts required can be detailed on the standardized equipment specification form. Relevant details about the chair(s) to which the insert is to be interfaced are also noted on this form.

In some cases, where funding is in place, equipment can be assembled and installed at the time of the assessment. This can save time and cost when travel involves long distances. Ordinarily however, the individual is seen a second time for fitting the seating device.

5) Instruction For Equipment Use

Once a seating system has been provided, the team must instruct the client and/or care givers to use the seat effectively. This includes information on break-in and maximum usage periods, instruction in how to do transfers, and how to properly position the individual in the seat. It is also important to explain how the insert is placed into and removed from the chair. Any adjustability built into the seating system, for example altering the recline angle or the amount of head support, should also be demonstrated.

Instructions for seating equipment use are reviewed during the clinic training program. It is emphasized that specialized seating is only a part of a total positioning program to be considered for the individual.

6) Adjustment

Posture is a dynamic activity which can be greatly influenced by growth and development, medication, the presence of equipment, etc.. As these influences and/or the individuals environment changes, the seat will require adjustment. During the start-up program it is recommended that a system be developed to recall seats for periodic review to ensure that necessary adjustments are made. The recall period will vary depending on degree of disability, level of care available to the client and complexity of the seating system.

CPSC staff are available to the seating team to recommend necessary adjustments. The training program is spread over a period of several months. The first three days of the training program are completed consecutively. The final two days are scheduled at the request of the seating team to provide problem solving support as adjustments and revisions are required.

7) Maintenance

Because most seating systems are used daily, periodic repair and replacement of parts will be required. CPSC asks that any damaged equipment be returned to the Kitchener plant for inspection by engineering personnel as part of an ongoing product upgrading program.

8) Follow Up

To determine the effectiveness of the seating system prescribed by the team, a method of obtaining client feed back is required. It is recommended that initially the team work with clients who can be followed easily. This will allow the team an opportunity to see results, both positive and negative, during early stages of clinic activities. A simple, standardized follow up form is included in the clinic manual for use by the team.

IN CLOSING

The model described above suggests a structure and procedure for providing specialized seating. The clinic start-up and operation process, like any process, continues to grow and change with experience -- ours and that of others.

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Software Design Methodology for Microcomputer Based Aids: Practical Considerations in Implementation

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ABSTRACT - The advent of microcomputer based technical aids has precipitated a need for increasingly complex and extensible software. A software design methodology is presented that supports the construction of software that is both portable and modular. Given the commercial availability and clinical use of many different microcomputers, it is desirable to standardize the hardware development environment in the laboratory and prepare software applications that will execute on one or more other target machines. The authors discuss the selection of a structured programming language that is portable to different hardware environments and conducive to the construction of independent program modules. An implementation of the design methodology in the cross development of a microcomputer-based communication system is described.

INTRODUCTION:

The microcomputer is currently a popular alternative to dedicated rehabilitative devices. A microcomputer-based aid typically consists of commercially available peripherals and software that provides communication and/or control functions [Vanderheiden, 1981]. In the development of such an aid, a substantial amount of project time is often spent in the design and construction of software central to the defined application.

The application software can exploit the flexible capabilities of the microcomputer host if it is well designed. When adequate time is allocated to the design phase of software development, programmer productivity is increased and the resulting programs are both maintainable and extensible [Myers, 1978].

The staff at the Rehabilitation Engineering Center has adopted a development model to be used by resident project teams engaged in software design and program generation. This model consists of five components:

1. Standardization - Utilization of the same hardware and software tools in all development projects [Jaffe, 1983].

2. Portability - Development of software that will execute in various hardware environments.

3. Local Modularity - Design and coding of program modules or routines which may be utilized by various applications programs without alteration and repeated compilation.

4. Global Modularity - Specification of a design strategy that produces programs suitable for integration in different application environments.

5. Cross Development - Construction of programs on a powerful host computer for a clinically intended target machine.

These components have been implemented in a laboratory-standard development environment and a structured software design methodology. The core of this environment is the IBM PC® family of microcomputers and the **C** programming language. Thoughtful design of structured programs has contributed to the goal of maximizing portability and global and local modularity.

DEVELOPMENT ENVIRONMENT:

Hardware and Supporting Software - Internal compatibility in the laboratory can be increased when the same hardware and software tools are used by all staff members. The potential benefits may include:

1. Ease of hardware maintenance and expansion

2. Operating system familiarity

3. Supporting software familiarity (eg., editors, compilers, utilities)

4. Project compatibility and increased interaction among staff

Performance capabilities and availability of supporting software have made the IBM PC family of microcomputers the choice of the Rehabilitation Engineering Center. This computer has been successfully used in the areas of communication, computer access, and robotics [Demasco, 1986; Demasco & Minneman, 1986; Minneman & Pham, 1986].

Programming Language - The **C** programming language is currently being used for nearly all software development in the laboratory. **C** is a language that facilitates portability [Grand, 1984; Brodie, 1985]. It has a well deserved reputation for producing object code that is both fast in execution and compact in size [Roberts, 1983]. A structured language such as **C** offers few obstacles to the coding of independent program modules and extensible, stand-alone programs. In addition, **C** provides facilities that diminish the need for low-level assembly language functions.

Cross Development - A common strategy used in commercial software production is cross development. By utilizing a more powerful host computer to compile program source code, an increase in programming efficiency can be realized. Additionally, hardware independent modules are written and debugged in the host environment. It is advantageous to isolate the hardware dependent modules and maximize the utilization of the host's native environment. This strategy is both efficient and maximizes portability

of the resulting application. The potential time savings is illustrated in Table 1. Three programs of differing size were benchmarked for compile times in 4 development environments (cross compilation times include serial downloading). It can be seen that it is most efficent to maximize development in the PC/AT native mode (90% of the development for CONFIG was done on the AT).

Computer: Environment:	PC AT (native)	PC (native)	PC (cross)	Apple (native)
NULL (36 bytes)	0:19	0:40	1:57	3:03
SIEVE (492 bytes)	0:24	0:49	3:07	4:14
CONFIG (14k bytes)	0:50	2:00	4:06	11:27

Table 1. Compile times (min:sec) for various development environments and source files (size in bytes)

This strategy has been applied in the development of software for the Apple II. **C** programs were edited and compiled on an IBM PC and downloaded to the Apple. Computer Innovations C86™ was used for native code development on the IBM AT. The Aztec C II© cross compiler by Manx Software Systems generated executable code for the Apple. One of the unique features of the Aztec compiler is the choice of machine-code or pseudo-code models. The pseudo-code model produces programs that are more compact but slower to execute than those generated by the machine code model.

APPLICATION OF DESIGN METHODOLOGY:

Description of Application - A micro-computer based communication system for non-speaking individuals was developed along the lines of the design methodology set forth by the authors. The Apple II was selected as the target machine for the system due to its prevailing use in clinical settings. A low-cost, commercially available, touch-sensitive tablet called the Powerpad provided the means to directly select programmable vocabulary items from a 12 x 12 inch active area.

Several features may be adjusted by a clinician or third party prior to a communication session. The active area of the tablet can be configured for any number of keys in a square matrix arrangement. The response time of the device to contact with the active area may also be adjusted to the desired "sensitivity". Vocabulary items selected for an individual user may be stored in a file and loaded into the Powerpad at run time. One or more output devices (eg., speech synthesizers, printers) can be interfaced to the system. Each of these capabilities lends a measure of customization to the communication system and may increase its utility as an assessment tool.

Environment Implementation - Software to support the system was built in the defined IBM PC/AT microcomputer environment and cross compiled for downloading to the Apple II target machine. Code generation and debugging was performed in the IBM AT native compiler environment for all machine-independent modules. Programs were then cross-compiled using the IBM PC. Once downloaded to the Apple, the programs were ready for execution without further compilation in the Apple native environment.

An initial aim of the project team was to minimize total executable code size to well within the limited memory of the Apple without contributing significantly to the execution time of routines responsible for real-time control of the hardware components. These run-time functions did in fact require the speed enabled only by the machine-code compiler model.

The necessary savings in memory space was found by using the pseudo-code model to compile the configuration programs of the clinician interface. The consequent slower execution speed of these programs did not adversely affect their operations, which are less time-critical than those of the run-time driver. Executable code size was reduced by approximately fifty percent, which was adequately compact.

Local Modularity - Development pursued the goal of maximizing the portability of all programs in the package. Employment of the **C** programming language contributed significantly to this end. It also enhanced the authors' abilities to construct independent modules each performing a well-defined function. These modules are extensible in the sense that they can be separately compiled and then linked to any applications program which requires the same run-time control functions.

Global Modularity - Modularity in program design has also produced adaptable, portable programs. The configuration programs that were written for this project are general purpose by design. Information access and modification is driven by a sequence of menu selections. Data files can be composed of menu selections appropriate to any application. It is then easy to integrate the programs in other environments without change to the source code.

The system configuration program created for this project enables a clinician or third party to specify the values of device operations parameters (eg., printer slot number = 4). In addition to its inclusion in any rehabilitation system package, this program can be used as a project team utility. One such application might be a "help" facility for programmers and program users.

Similarly, a vocabulary configuration program provides the functions necessary to select and organize a set of vocabulary items for an individual user. While this program is primarily used to affect the content and placement of language units in a square matrix, it can be employed as a simple editor for any data file composed of a list of characters, words, and/or phrases.

The configuration programs described give the clinician or concerned third party some limited means to customize the communication system for each individual user. However, since these programs are currently designed to accept standard keyboard input, the individual who will use the system has little measure of control of the adjustable parameters. Ideally, if the configuration programs were modified to also accept Powerpad key input, the system would be user-programmable to the same extent that it is currently clinician-programmable. The project team plans to implement this extension.

CONCLUSION:

The authors are working toward refinement and further application of the presented software design methodology. Several C interpreters have recently become available for the IBM PC. Although interpreters execute code slowly, they provide an interactive and efficient programming environment. The REC is currently incorporating a C interpreter in its development strategy. In addition, cross development is being extended to the Radio Shack Model 100/200 computer for the design of portable communication aids.

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THE DESIGN PROCESS VIEWED AS A TECHNOLOGY

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ABSTRACT

Analogy is made between the process of selection of treatments and medication in medicine and the design of technical aids in rehabilitation engineering. Both have problems in "compliance." For engineers such problems are evident in lack of use of the technical aids by clients for whom they were prescribed. More effective aids may be developed with a more conscious application of the methodology of the design/planning process when it is viewed as a technology in its own right. Several aspects are described of what has been a more effective planning process as used with persons with disabilities in management of health problems. Particular emphasis is on the criteria for measurement of patient/client participation in the planning process leading to the goal statement.

STATEMENT OF THE PROBLEM

The practice of engineering and medicine have certain commonalities. They are both applied sciences in that the role of practitioners is to improve the human condition rather than merely study it. The goal of both practitioners is to increase the "fit" between the person with disabilities and his or her environment by the use of technology. The rehab. engineer is concerned with the design of appropriate technical aids; the physician with the selection of appropriate treatment programs and medications. There may also be analogies in the methods both use for the identification of the technologies to be used. Indeed the term "prescription" is sometimes used by both sets of practitioners. These similarities may also extend to the problems which exist in both fields arising from the ineffectiveness of the selection process.

One problem mentioned in both fields is the lack of "compliance" in use of the medications suggested or the items designed. Engineers are concerned with the degree to which their technical aids remain unused. The effectiveness of both professions may be enhanced to the degree that this selection process may be made more effective. Techniques in use with health problems of persons with spinal cord injury may be helpful to engineers in dealing with analogous issues in their own work.

A medication may be inappropriate on several counts. It may be the wrong type of drug. It may be used in too low a dosage or too high when applied in some routine manner. A drug may be selected for a particular disease or set of symptoms and approved for certain indications and not others. Much less attention is paid to the process by which any individual drug may be made to be efficacious in the individual patient. This same problem may exist in other fields where practitioners design remedies for problems not quite fitting the person.

The selection process is viewed as an ongoing planning process. In each instance, a plan is generated consisting of a goal and technical aids or medications as means for accomplishing those goals. This plan, both the goals and the means, is then subject to ongoing evaluation and further modification. The method by which the plan is generated is here viewed as a technology in its own right; and thus subject to review just as the technical aids must be evaluated in terms of their effectiveness in meeting the goals.

It is the premise of this presentation that a conscious awareness of the parameters of a more effective planning methodology can support the entire design activity. Awareness of the process by which the plan is made is crucial in ensuring it is appropriate in meeting the needs of the client and thus be one which is more likely to be carried out.

DEVELOPMENT OF AN APPROPRIATE PLAN

In developing a plan, the first task is to define the problem to be alleviated. In medicine, the patient is asked to state his or her complaint. These are generally stated in terms of symptoms such as pain or cough. For those with spinal cord injury, one relatively frequent complaint deals with muscle spasms. The problem for the person is not necessarily the spasms per se but the functional consequences of the spasms. For one person, it may be a concern about being thrown out of his wheelchair during the day; for another, kicking their bed partner at night. Some have only occasional difficulties; others have more constant problems. Such more precise delineation of the difficulties can arise only out of a complete exploration with the patient since they must be defined in the context of his or her own life.

Given a more adequate delineation, it is then more likely that the goal to be established will deal with what is really troublesome to the patient. This process of problem identification is of course an ongoing one. The client becomes better able to state his problems as he has experience in doing so.

The technique in use for more adequate problem identification is a brainstorming procedure in which judgment as to the problem is suspended at least through three stages of exploration. Only after having explored the question as to concerns these several times is it useful to actually select the problem that will serve as the context for the goal statement.

For example, a person with left-sided weakness when asked as to her problem first stated it to be as "a weak left arm." When asked as to the difficulties the weak arm was causing her in her life, her answer was: "I can't take care of my children."

When asked, now for the third time, as to what she was unable to do for her children that particularly concerned her, her answer was "not being able to make them their meals." As an example,she later mentioned things such as making them a sandwich when they needed one. The design of an appropriate technical aid would now be enhanced once the patient has defined her problem in these more functional terms. These several levels of exploration were necessary in order to define the problem to be solved with a technical aid just as the proper application of medication for the treatment of spasms depends on when that problem occurs.

After exploration of the problem has gone on in this model, the actual selection of the "best" statement of the problem or the highest priority problem can now proceed. For some patients, the cycle of three-fold exploration and then selection may require repetition before an adequate delineation has occurred. One criterion in use for evaluation of the "adequacy" of the problem definition process is patient agreement and confirmation as to the relevance of the problem when it is fed back by the interviewer - a "check-out" process.

The identification of the goal to be achieved can now proceed. Once again exploration of several alternative goals is useful prior to the selection of the highest priorty goal or the "best" statement. Once again, it is necessary to have the patient describe the goal in functional terms that are meaningful to the patient. If the goal is couched in those terms, the patient may then become a major participant in evaluating the degree the goal is accomplished.

For example, in the management of the functional problems arising out of the spasms, one patient stated his goals to be:"not to fall out of my chair in the early morning each morning when being washed." Still another person stated his goal to be "not injuring myself on the radiator when transferring to do bowel care every other night." Still another was "to be able to drive my electric chair without hitting anyone during the day despite my spasms." The goal in each instance was not to do away with the spasms per se but the functional consequences which could be defined only with involvement of the patient in terms of his own life.

In addition to enabling the patient to explore and select in terms of his own life, still another aspect of the planning/design process is the degree to which the goal is "specific." A three point criterion is sought to include answers to the questions as to:WHAT is to be accomplished,WHEN or WHERE (the conditions) and HOW MUCH (to what degree). The goals described above in each instance met these criteria although couched in terms arising out of the patient's own experience.

The content sought by use of the planning questions is not new. What is important is the relatively self-conscious manner is which there is adequate exploration of the problem and specification of the goal. The third measure of the planning process is the maximization of the degree to which the answers to the planning questions arise from the patient vis a vis the professional serving as the interviewer.

A MEASURE OF PATIENT PARTICIPATION

A scale has been developed to enhance the degree of awareness as to how one may enable the patient to participate. At the highest level of "independence," the patient both asks himself the questions as well as provides the answers. At the more common level of "free choice," the professional may ask the questions as to problem and goal, but the client provides the answers as exemplified by the woman described earlier who had difficulty with her left arm. When necessary in order to meet the goal of a specific statement, the professional may then move to the next level of "multiple choice." The patient has merely to choose from several options provided by the interviewer in order to find answers to the planning questions, but is still making a major contribution to the decision-making. Movement to the next level of "concurrence," marks a significant shift. At this stage, the patient has merely to state "yes" or "no" to the decision already made since but one answer is recommended. Many decisions for the selection of medications and technical aids are made at this level of participation at time inappropriately so. Even less appropriate would be use of the stage when the client is told what to think and has merely to acquiesce - to be "complaint" and perhaps even more often to be "non-compliant."

The criteria for maximization of the degree of participation has been 1) for initiation of the questioning at the highest stage of participation, 2) to go down the scale only one step at a time and only when it is necessary to meet the need for specificity, 3) to return to the higher stages of participation as soon as possible.

CONCLUSION

Awareness of several technical aspects of the planning process may make the design of technical aids more effective as such awareness has made the choice of medication more effective. The goal of the planning process has been to define a specific goal in an area of concern to the patient with the maximal degree of patient participation. Input is welcome from engineers as to how their ways of specifying the technical aids they design may help in specifying even more the procedures used in this technology of the design process itself.

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A SERVICE DELIVERY PRACTICE CONSIDERED

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ABSTRACT

In the context of rehabilitation engineering service delivery, one seemingly widespread practice is examined. The practice involves customization of rehabilitative aids when equally appropriate applications of commercial rehabilitation technology are available. The appropriateness of this practice is questioned. Reasons for the persistence of the practice are introduced and implications analyzed. Finally, recommendations are made to minimize the practice so as to maximize the benefits of rehabilitation engineering and rehabilitation technology for persons with disabilities.

INTRODUCTION

As increasing attention is given to rehabilitation engineering service delivery issues, a serious question arises as to the appropriateness of one seemingly widespread rehabilitation engineering practice. The practice in question involves fabricating assistive devices for clients whose functional needs can be met most effectively by using either off-the-shelf or slightly modified commercially available technology.

From both a service delivery and an economic standpoint, this practice represents inefficient and costly use of valuable rehabilitation engineering capabilities. The client receives customized engineering services when less engineering-intensive applications of commercial technology are appropriate. Consequently, other individuals, who do need those engineering services, may not receive them. Equally important, fabricating a device for one person, despite the existence of an appropriate commercial alternative, effectively raises the unit cost of that commercial product to all other persons with similar needs, making it more difficult for them to benefit from available rehabilitation technology.

The practice of engineering a rehabilitation device from scratch, when an off-the-shelf or slightly modified commercial alternative exists, therefore, can be described as an over readiness to fabricate. If this is the case, three associated questions arise: Why does the practice persist? What are the implications of that practice? What might be done to minimize the practice?

MISCONCEPTIONS WHICH MOTIVATE UNNECESSARY FABRICATION

There are two principal situations where rehabilitation technologists customize rehabilitation aids when similar devices are available commercially. The technologist either is unaware of the commercial alternative, or is aware of the alternative, but chooses to fabricate despite that knowledge.

Unawareness is essentially a problem of failures in marketing or information flow and, although a very important problem, is outside the scope of this paper. Conversely, situations where technicians are aware of commercial options, yet continue to fabricate, indicate an underlying problem of technician misconceptions regarding commercial rehabilitation technology. These misconceptions help to explain why the over propensity to fabricate persists.

Three common misconceptions warrant particular consideration. Those are a misunderstanding regarding the costs of commercial technology, a misconception as to the benefits of commercial technology, and a failure to differentiate between a client's functional versus financial needs.

Costs of Commercial Technology

The first misconception is that commercial rehabilitation products are excessively expensive. It is commonly assumed that the price of a commercial product includes significant, and inappropriate, profits to the manufacturer and distributor of that product. Experiences of staff at the Electronic Industries Foundation Rehabilitation Engineering Center (EIF/REC), when dealing closely with commercial companies over the past several years, indicate that this assumption is usually inaccurate and unfair.

Providing specialized rehabilitation products commercially is a very costly undertaking. In addition to incurring costs common to the course of business for any industry (e.g., overhead, start-up, production, distribution and marketing costs), commercial providers of rehabilitation technology incur costs unique to their special needs markets. For example, for most companies in commercial rehabilitation technology markets, the costs of marketing a product can be unusually high, sometimes higher than all the costs of producing the device. This is because companies must market to a number of target groups (e.g., third party payers, prescribing clinicians/technicians, and the end users of the product), many of which are small and hard to reach, precluding the use of more cost-effective, mass marketing techniques.

Not only does commercial provision of rehabilitative devices entail relatively high total costs, unit costs are also high because the volume of devices demanded is small. Compared to most consumer product markets, there are relatively fewer people who benefit from, and who pay for specialized products offered through rehabilitation product markets. Each person,

therefore, must pay for a relatively larger portion of the total costs of the product.

Benefits of Commercial Technology

Closely related to the first misconception, regarding the costs of commercial products, is a failure by many technologists to recognize the total benefits of commercial availability. Often service providers consider only the commercial device itself, and react to the expense of procuring that device by fabricating something similar. What those providers fail to realize, however, is that a consumer of a commercial device actually receives much more than just a device for that price. The client, in effect, receives a significantly different "product" when procuring a commercial device versus obtaining a customized one. Closer examination reveals important differences between the two.

The commercial device, itself, can be significantly different from a customized one. From an engineering standpoint, the former products tend to be more refined than the latter. From prototype development through the final production phase, commercial devices typically receive more iterative engineering than could be duplicated through customization, given service delivery time and financial constraints.

Beyond engineering comparisons, the person who procures a commercial rehabilitation aid receives various assurances as to the operation of that product, which rarely are replicated through fabrication. The commercial device typically has been extensively evaluated as to safety and efficacy of use. Field tests also are often conducted to determine the product's appropriateness to address needs shared by numerous potential users. A commercial product typically carries a warranty against malfunction for a specified period. If malfunction occurs, modular replacement parts may be available. Further, commercial companies routinely carry comprehensive liability insurance to protect against the risk of injury to the product user. This, of course, benefits the company, but it also, again, provides important assurances to the product user.

Perhaps the most important attribute of a commercial rehabilitation product vis a vis a customized one, however, lies in its availability. Commercial companies market and distribute rehabilitation technology, and the associated benefits outlined above, to a number of disabled individuals in a fairly extensive service area. Conversely, rehabilitation engineering service organizations provide valuable technology to persons with relatively individualistic needs in much smaller geographical areas. Each approach to rehabilitation technology service delivery has comparative advantages over the other. When a piece of technology can help a relatively large group of people, commercial availability offers some valuable advantages to customization.

Wider availability of a particular type of equipment can be very important to many disabled individuals. It can minimize dependence upon availability of customization capabilities. This can free individuals to relocate geographically, or to travel with less concern about equipment failure. Wider availability also means that persons in areas which do not have adequate rehabilitation engineering services can benefit as much from rehabilitation technology as can others in areas which offer more extensive service delivery options.

Commercial rehabilitation devices, therefore, are not completely interchangeable with customized devices. One cannot replicate all of the attributes of a commercial aid simply by trying to duplicate its product design.

Functional Need Versus Financial Need

A final misconception shared by many technologists who fabricate unnecessarily is a failure to differentiate between a client's functional need and that client's financial need. When a commercial device is appropriate for one's functional need, fabricating a similar device is really an attempt to address that person's financial need.

There are better ways to help people acquire useful commercial rehabilitation technology when they have trouble affording it. Some of those approaches will be outlined in a subsequent section of this paper.

Not only are there better approaches to addressing financial need, but addressing that need through fabrication is inappropriate and costly. An over propensity to fabricate has associated cost implications which need to be clarified.

IMPLICATIONS OF UNNECESSARY FABRICATION

An over propensity to fabricate can have three adverse results.

The first result of unnecessary fabrication is the impact it has on the individual who is receiving the customized service. As described earlier, the client actually can receive a significantly different benefit package when obtaining a customized device versus procuring a commercial one. Because of such attributes as engineering characteristics, maintenance and repair assurances, and availability considerations, the two types of equipment are rarely directly interchangeable.

A second result from an over propensity to fabricate relates to the opportunity cost of misusing the rehabilitation engineering capabilities. Presumably there is unmet need for rehabilitation engineering services in any program's service area. That is, there are individuals with functional needs for which access to the fabrication skills of a qualified rehabilitation technician is the only way that they will benefit from rehabilitation technology. When that technician's time, and other limited service delivery resources, are used to

address needs for which commercial options exist, those resources are being channeled away from uses for which they are critical, toward uses for which they are not.

The third result of unnecessary fabrication is the impact it has on the unit cost of commercial products. When a technician customizes an aid for a client who can be served appropriately through procurement of a commercial alternative, potential demand for the commercial product is lowered. Most of the total costs of providing that product commercially, however, remain the same. It is the unit cost, and hence the retail price of providing the product, which is affected. Price rises due to decreased demand. Any other individuals, with disabilities similar to those of the client, who are seeking to address their functional needs through acquisition of commercial technology, must pay a higher price for that technology as a result of unnecessary fabrication. Ironically, by addressing one person's financial needs inappropriately through fabrication, that action can exacerbate the financial needs of many other individuals.

STRATEGIES TO MINIMIZE UNNECESSARY FABRICATION

Given the motivations for and the implications of unnecessary fabrication, three strategies are suggested to guide rehabilitation engineering service delivery. The first strategy, of which this paper is one attempt, is to eliminate the misconceptions which are motivating technicians to customize devices, despite being aware of commercial alternatives.

A second approach is to establish service delivery policies at an organizational level, which recognize the respective advantages of commercial technology and of customized technology. Project IMPART, a service delivery model operationalized for a period in the late 1970's, instituted one such administrative policy. The Project established a hierarchical technology selection process to guide service delivery efforts, internally. A generalized version of that selection hierarchy is as follows:

- utilize commercially available aids and devices, "off-the-shelf", whenever they meet an individual's needs appropriately. This alternative includes use of both specialized rehabilitation products, as well as mass market equipment, when applicable.
- modify commercially available aids and devices when necessary.
- fabricate an aid or device, from "scratch", only when the previous two options have been exhausted.

A final strategy, which should be used in conjunction with the other two, would be to seek ways, other than fabrication, to address a client's financial needs, when that is indeed what they are. That might be achieved internally, if a service delivery program has the capability to provide financial as well as engineering services. More frequently, it will require referring the client to third party payment agencies, local philanthropic groups or other financial service providers.

To facilitate referral, efforts are being taken to improve financing options for individuals seeking to acquire commercial rehabilitation technology. A number of innovative service delivery programs around the country have instituted no or low interest revolving loan funds to help residents of their respective service areas pay for needed devices. EIF/REC, and other organizations are studying these model financial service delivery programs, hoping to replicate some of them in other regions. EIF/REC also is looking to expand financing alternatives for consumers of commercial rehabilitation technology through other means. The Center has organized a national task force to consider ways in which third party payment agencies can become more responsive to the growing needs for rehabilitation technology. EIF/REC is also investigating ways to increase access to more conventional, commercial financing sources, recognizing that there may be a number of potential consumers who might qualify for and appreciate that alternative.

SUMMARY

The rehabilitation engineering practice of fabricating aids for which commercial alternatives exist tends to persist because of provider unawareness. Some technicians are unaware of commercially available technological options. Others may be aware of the commercial alternatives, but continue to fabricate similar devices because of failure to recognize either the implications of that practice, or that their actions might be motivated by certain misconceptions about commercial technology. There seems to be a need, therefore, to address unnecessary fabrication practices by expanding awareness: awareness of the problem of unnecessary fabrication, the costs of its persistence, and strategies to minimize the practice.

ACKNOWLEDGEMENTS

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SERVICE DELIVERY AT THE UNIVERSITY OF TENNESSEE REHABILITATION ENGINEERING PROGRAM

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Rehabilitation Engineering Program, Memphis, Tennessee

ABSTRACT

The following article describes the major elements of the Client Services section of the University of Tennessee Rehabilitation Engineering Program. The expertise of the program is the provision of specialized custom seating systems as a preliminary to in depth evaluation for other solutions, such as independent mobility, or communication devices.

INTRODUCTION

In order to shorten this background summary the following comments will be restricted to the provision of seating systems. This however, does not reflect the Service and Development efforts in the direction of "total rehabilitation" by the provision of communication and mobility aids and aids for independent living.

The Client Services section of the University of Tennessee Rehabilitation Engineering Program (UTREP) has been evolving over the last 12 years.

In the early years service was provided as a part of the Research and Development effort at no charge to the recipient (client). These services were the basis for the R & D carried out.

It had become apparent that amongst the population served, (mainly children with CP), the primary need was for special seating. Once function had been maximized, it was then more appropriate to carry out further evaluations for communication and/or independent mobility where desirable. The devices indicated were then provided, or developed and provided, where resources permitted.

In the early years the population served was relatively small and it was possible to create "from scratch" devices to meet the needs of the client, without too much regard for time taken.

It was quickly determined that this process was less than efficient and the Modular Plastic Insert (MPI) seating system was developed with a simulator system to aid prescription. At one time this comprised about 60% of the devices fitted, and also became commercially available on a widespread basis- **this was a key technical development in the provision of services on a cost effective basis.**

By early 1982 the seating devices provided consisted of MPI, Foam and Plywood, Vacuum Formed Shells and Foam in Place systems.

TECHNIQUES

We currently have available 5 different seating systems and multiple accessories (total 64) which allow us to provide for clients with a wide range of disabilities. The techniques are MPI, Foam and Plywood, Foam in Place, Bead System and Shapeable Matrix.

These "off the shelf" techniques can often be combined to produce innovative solutions to client's problems and enable us to take into account growth, and other changes which might occur, to ensure that the device has a useful life.

Associated with each of these techniques is a record of the cost and the time it will take for one technician to complete.

TEAM PRESCRIPTION

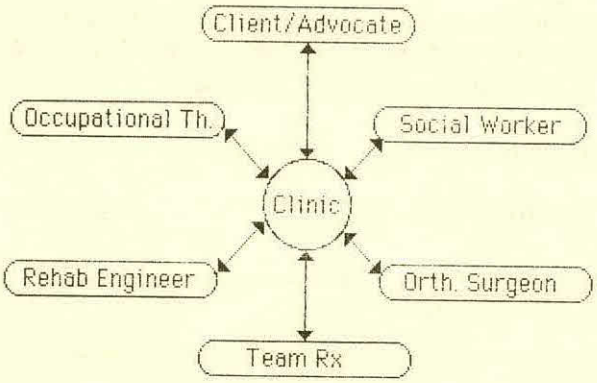
One of the most powerful tools we have is our Clinic Team, at which an Orthopedic Surgeon (Team Leader), Therapist, Rehabilitation Engineer, Social Worker and the Client and/or their advocate participate in the prescription process.

The various UTREP staff members are involved in evaluations and simulations in the morning and during the afternoon Clinic, the Therapist presents his/her evaluation and recommendation, the Social Worker presents any relevant information and the Engineer is there as a technical advisor to ensure compatibility between the goals for the client and the technical resources available.

It is this process which simultaneously combines all the available information, and from which the recommendations emerge.

We **do not offer** "I wish I could help solutions". All the staff are sympathetic to individual clients needs but if we are not confident that a specific combination of the systems and accessories will work then we will not be able to help that individual. (Occasionally it is possible for the R & D Program to "pick up" such an individual who may represent a group for which there is no commercially available or UTREP solution- in the hope that this will lead to the development of a new device).

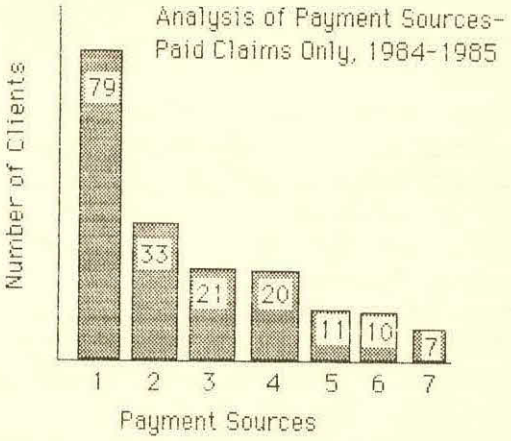
Decisions must be the product of the clinic:- if we are to provide a device we must define what it is, how much it will cost and how many technician hours will it take to deliver. Other services may include evaluations in the home, school or workplace, evaluations for communication devices and follow up.



FUNDING

The concept of fee for service is simple- putting it into practice on the scale of recovering $300,000.00/year on devices which most insurance companies and third party payers have never heard of is very complicated, time consuming and expensive. A staff of three have a primary responsibility of determining that funding is in place for a particular individual. This involves provision of correct information to the third party payers, billing and scheduling. See chart for information on funding sources. Without the development of this expertise it is doubtful that there would be a significant Service Program at UTREP.

The process of obtaining funding is largely educational, as most of the third party payers have never heard of the devices we deliver. In order to help this process along we have developed consistent terminology in Therapy and Doctor's notes, consistent pricing policies, brochures with pictures describing the devices to be provided (or manufacturers brochures are included) and slide presentations for seminars.



1) Insurance/Family/Medicare/Champus
2) State Medicaid
3) Rehabilitation Centers/Residential Facilities
4) Family, includes some donations
5) "Special Friends"- paid total cost
6) "Special Friends"- paid part of costs
7) Crippled Children's Program, Voc Rehab

The process of obtaining funding is largely educational, as most of the third party payers have never heard of the devices we deliver. In order to help this process along, we have developed consistent terminology in Therapy and Doctor's notes, consistent pricing policies, brochures with pictures describing the devices to be provided (or manufacturers brochures are included) and slide presentations for seminars.

Experience has shown us that even with the best of efforts, it is sometimes impossible to find funding for individuals of limited means. In this case, it is sometimes possible to find funding through the mechanism of a charity we have developed called **"Special Friends"**. Individual needs are brought to the attention of local philanthropic groups, with an accompanying professionally produced tape/slide presentation, in order to obtain sponsorship for that person. After provision of the device(s), photographs are sent to the sponsoring organization and visits to our Service Program are also encouraged.

WELL TRAINED STAFF

It is vital that staff are comfortable and well trained in the various techniques- the introduction of a new

technique is done an a controlled basis under the supervision of an engineer until a level of competence is achieved.

INSTRUCTIONAL SEMINARS

At least once a year, UTREP hosts an instructional seminar. These have recently concentrated on the "hands on" approach with "real clients" being fitted. Service Program staff serve as Group Leaders, since they have the up to the minute experience of dealing with the needs of their clients. This is a time of much exchange of information on an informal as well as formal basis, and is a learning experience for the UTREP staff.

GENERAL

A combination of factors has lead to our survival, which as individual items have equal importance, but it must be stressed that the combination of these factors into a **team product** is what makes things work.

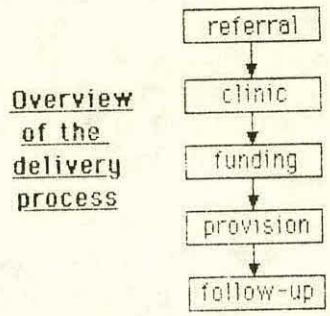
The availability of components is essential to the prompt and efficient provision of complete devices- the identification of reliable sources of supply of components and an accompanying computerized inventory control program have been major steps forward. Associated with the inventory control program has been the development of a computerized accounting system and the continuing development of a client data base.

A computer bulletin board has been established in the hope that persons involved in all aspects of the provision of "technology for disabled individuals" will pool and share information. For more information on this contact Doug Hobson at the address below.

WHERE WE ARE NOW

In 1985 the UTREP Service Program provided 190 seating systems with a total value of $300,000.00. Staffing for most of this time consisted of 6 people:- 2 technicians, 1 administrative, 1 clerical, 1 therapist and 1 engineer.

While this has been our most succesful year to date, in financial terms, we are not complacent and recognise room for improvement in several areas and will be concentrating on consolidation at what will be our maximum capacity with our expanded staff level of 9.



CONCLUSIONS

Our experience has shown us that **it is possible to run a viable Service Program for the provision of Rehabilitation Technology** For further information please contact Doug Hobson.

FOOTNOTE

The UTREP has been directed since its inception by:

Doug Hobson, Technical Director and Dr. Robert Tooms, Medical Director

Current staff at UTREP Service Program:-

Joe Abraham	Workshop Supervisor
Hope Erwin	Administrative Assistant
Myra Kessel	Deputy Director
Bobby Medford	Technician
Billy Millican	Technician
Lynn Monahan	Occupational Therapist
Nigel Shapcott	Director
Brenda Smith	Secretary
Susan Taylor	Supervisor, Occupational Therapy

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NATURE AND PROPORTION OF PHYSICAL IMPAIRMENTS AMONG INDIANA'S FARM OPERATORS

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ABSTRACT

This study was conducted to determine: (1) the proportion of Indiana's active farm operators who have physical handicaps; (2) the types of physical handicaps which might hinder active farm operators in completing essential farm-related tasks; and (3) what these farm operators perceive as significant barriers to their operation of agricultural equipment and completion of other essential farm-related tasks. To accomplish these goals a random sample of 500 farm operators were surveyed. The survey revealed that approximately two-thirds of Indiana's farm operators suffer from one or more physical impairments.

INTRODUCTION

In 1984 approximately 290,000 farm and ranch residents, nationwide, were involved in accidents that resulted in disabling injuries (Accident Facts, 1985 Edition). A summary of 31 state farm accident surveys, completed by the National Safety Council, revealed that almost two-thirds (65.3 percent) of reported farm injuries were considered severe (broken bones, cuts requiring treatment, sprained back, etc.) with an additional 1.9 percent resulting in permanent disability (Hanford, Burke, Fletcher, Hoskin & Miller, 1982, pp. 7). National Safety Council data shows that agriculture is now the second most hazardous of all occupations with respect to the total number of accidental fatalities and disabling injuries (Accident Facts, 1985 Edition).

Physically handicapped farm operators face barriers which make completion of essential farm-related tasks difficult or even impossible. These barriers are especially burdensome with respect to operation and maintenance of agricultural equipment. According to tabulation of farm accident surveys in 28 states by the National Safety Council (October, 1981), 1.2 percent of the farm accident victims required rehabilitation. Only 16 percent of the rehabilitated farmers chose to return to farming as an occupation (National Safety Council, October 1981).

THE NEED FOR RESEARCH

A review of literature including research conducted by state and federal agencies and major universities suggest that even though there is substantial information available on the types of disabilities affecting the general population, there is a lack of information concerning the prevalence of physical handicaps among the farm population. The need for research in this area if supported by several sources such as former Secretary of Agriculture Bergland in his 1977 Annual Report to Congress which stated, "Relevant data (concerning the rural physically handicapped) is not available; there is no current ongoing system for data collection on the characteristics or number of handicapped individuals ..." (Knebel, 1977, pp. 59).

Not only is there little current data concerning the nature and prevalence of physical handicaps among the farm population, but the data that is available lacks consistency. Data from one study to another varies depending on the definition of disability and the age of the population surveyed. For instance, a National Center for Health Statistics study (Series 10, No. 124, 1978, pp. 31) revealed that 15.2 percent of the farmers suffered from arthritis, while a Michigan study (Baker & Wilkinson, 1974, p. 32) found that 27 percent of Michigan's farmers were impaired by arthritis.

In addition to increasing the amount of information available concerning the types of physical handicaps affecting the farm population, there is a need for more investigations into the job-related problems encountered by physically handicapped farmers. On-site interviews and communications with physically handicapped farmers indicate that there are few resources available to assist them in modifying essential farm-related tasks to help them in performing them.

Handicapped farmers who need their farm operation modified so that they can continue to perform essential farm-related tasks must often rely on their own ingenuity and local craftsmen to make the necessary modifications. Equipment manufacturers are often hesitant to modify farm equipment for the physically handicapped because cost and possible product liability problems. According to Field and Preiss (1980), cases have been encountered in which local farm implement dealers have refused to make modifications to farm tractors for use by physically handicapped farmers because of questions relating to liability.

PROCEDURES

The Indiana Agricultural Statistician's office was discovered to possess the most comprehensive listing of farm operators in Indiana (total listing of 88,000 farms). This listing provided the data base from which a sampling of farm operators was drawn.

To insure that the sample was not composed of an overproportionately large group of small or large acreage farms the stratified random sampling procedure was used. Five hundred Indiana farm operators were selected, at random, for inclusion in the study.

Before administering the questionnaire to the randomly selected sample of active farm operators, a pre-test of the questionnaire was conducted to insure that the questions would be clear to the readers. Students enrolled in Purdue University's Agricultural Short

Course were chosen to pre-test the questionnaire (they came from all areas of the state and were actively engaged in farming). The pre-test revealed that several questions were unclear and thus did not provide consistent, useful data. Changes were made and the revised questionnaire was again pre-tested by a second group of students.

Following the development of the questionnaire, it, along with a cover letter, was mailed to the survey sample. Follow-up attempts to reach the non-respondents included: (1) post card reminder mailed approximately two weeks after the questionnaire was distributed, (2) phone contact with a random sample of non-responders, and (3) mailing of the questionnaire a second time.

Two-hundred and seventeen (217) farm operators completed and returned the questionnaire, of which 196 were deemed complete enough to be included in the study.

ANALYSIS OF THE DATA

The Statistical Package for the Social Sciences (SPSS) subprograms, "Frequencies" and "Crosstabs" were used to analyze the data. Relationships among the various variables were tested for statistical significance using the Chi Square statistic. Phi, Cramer's V, Tau b and Tau c were used to measure the strength of the associations.

RESULTS OF THE STUDY

Physical Impairment Characteristics of the Farm Operators

Of the 196 farm operators analyzed, 60, or 30.6 percent indicated that they were affected by a musculoskeletal impairment. The most common musculoskeletal impairment was arthritis, reported by 20.9 percent of the farm operators. In addition, 9.2 percent of the farm operators mentioned having chronic back problems.

Hearing impairments affected 25.0 percent of the farm operators. Approximately twenty (19.9) percent of the farm operators indicated that they were hard of hearing.

Other hearing impairments cited were: use hearing aid (1.5 percent), deaf in one ear (2.0 percent), and deaf in both ears (0.5 percent).

Twenty-four (24.0) percent of the farm operators revealed that they suffered from a cardiovascular impairment, as it was possessed by 12.2 percent of the farm operators. Other commonly cited cardiovascular impairments were: prior heart attack (5.1 percent), and hypertension (4.6 percent).

Nearly twenty-two (21.9) percent of the farm operators disclosed that they possessed a respiratory impairment. The most common respiratory impairment, allergy other than hay fever, was cited by 9.2 percent of the farm operators. Hay fever was cited by 4.6 percent, asthma by 3.1 percent, chronic bronchitis and emphysema each were mentioned by 2.6 percent of the farm operators, while 1.5 percent indicated that they suffered from occupational lung disease.

Vision impairments were encountered by 7.1 percent of the farm operators. Vision impairments cited included: cataracts (1.5 percent), night vision problems (2.0 percent), color blindness (1.0 percent), glaucoma (1.0 percent), and total vision loss in one eye (0.5 percent). Interestingly enough, 52.6 percent of the farm operators indicated that they wore corrective lenses (eye glasses, contacts, etc.). (Corrective lense wearers were not considered physically impaired.)

Amputations were cited by 5.1 percent of the farm operators. Amputations affecting the left side of the body were cited more frequently than amputations on the body's right side.

Two and six-tenth (2.6) percent of the farm operators cited a congenital malformation. One farm operator mentioned that he was missing his right hand.

Neurological impairments were mentioned by 2.0 percent of the farm operators. One farm operators indicated that he was affected by neuritis.

Other impairments cited by the farm operators, in addition to the major impairment categories previously mentioned, included: ulcers (4.10 percent), diabetes (5.1 percent), gout (2.6 percent), chronic skin disease (1.5 percent), gall bladder disease (1.0 percent), kidney disease (1.0 percent), and liver disease (0.5 percent).

In summary, of the 196 farm operators surveyed, 131 or 66.8 percent indicated that they were affected by at least one physical impairment. Several farm operators listed more than one physical impairment.

It was also observed that physically impaired farm operators were older than the physically non-impaired farm operators, averaging 53.6 years of age as opposed to a 46.5 year average for physically non-impaired operators. And physically impaired farm operators were more likely to be overweight than were physically non-impaired operators, with 50.4 percent of them overweight compared to 36.9 percent of the physically non-impaired farmers.

Effects of Physical Impairments upon the Farm Operators

Thirty-four (34) farm operators, 17.3 percent, indicated that there were agricultural-related tasks on their farms that they were no longer able to perform because of their physical impairments.

Of the 196 farm operators, 38, or 19.4 percent, said that because of their physical impairment they were hindered or limited in their ability to perform necessary farm-related tasks.

In addition, 9 farm operators, or 4.6 percent, indicated that they had made modifications to their farm tractors, implements or buildings which have helped them overcome a physical disability.

Thirty-eight (38) farm operators (19.4 percent) stated that they required assistance from a neighbor, employee, or family member to perform some necessary tasks in their farm operation. Family members were most often relied upon for assistance as 14.3 percent of the physically impaired farm operators depended upon family members for assistance. Neighbors assisted 3.1 percent of the physically impaired farmers and non-family employees provided help for 4.6 percent of the physically impaired farm operators.

Twelve (12) farm operators, or 9.2 percent of the physically impaired farm operators, indicated that their physical impairment was the result of a farm accident.

Twenty-eight (28), or 21.4 percent, of the physically impaired farm operators revealed that they were willing to share their experiences with other farmers who face similar barriers in their farm operation.

There was a moderately strong positive correlation between cardiovascular impairment and "essential farm-related task(s) prevented". In addition, "essential farm-related task(s) prevented" and "assistance required to perform necessary farm task(s)" also exhibited a very strong positive correlation; that is, operators prevented from performing essential farm-related tasks were quite likely to require assistance from others.

CONCLUSIONS

Physically impaired farm operators are significantly older than the physically non-impaired farm operators. Even though there is very little difference between the average height and weight of both types of operators, physically impaired farm operators were more likely to be overweight than the physically non-impaired farm operators.

The most common physical impairments among farm operators are: (1) musculoskeletal, (2) hearing, (3) cardiovascular, and (4) respiratory. Overweight farm operators experience a higher incidence of physical impairments than non-overweight farm operators, especially in the cardiovascular and musculoskeletal categories.

Overweight farm operators, due to a physical impairment: (1) experience greater difficulty performing essential farm-related tasks; and (2) were more likely to require assistance from family members, neighbors, or non-family employees than non-overweight physically impaired farm operators. Farm operators affected by cardiovascular impairments experience greater difficulty completing essential farm-related tasks than their non-impaired counterparts.

The study found that physically impaired farm operators who are unable to perform certain essential farm-related tasks must rely on family, neighbors, and non-family employees for assistance in completing the tasks.

Physically impaired farmers who were unable to perform essential tasks around their farms indicated a high degree of willingness to assist other farmers who may be in a similar position.

DISCUSSION

Contact with physically impaired farmers, through farm visits and at workshops for handicapped farmers, revealed that there is a strong desire on the farmers' part to remain active in their farm operations. Physically impaired farmers may never be completely independent farm operators, but with assistance they can take an active part in their farm operations. Without help from industry and society in general, many farmers who become handicapped will be forced to give up farming. Not only do the physically impaired farmers suffer, but society suffers the loss of productive farmers. Our society has developed to the point of placing men in space and on the moon and therefore should have no reservations or excuses for not supporting the means to keep physically impaired farmers active in the production of food and fiber.

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10.5

PROSTHETIC AND WORKSITE MODIFICATIONS FOR FARMERS WITH UPPER EXTREMITY AMPUTATIONS

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ABSTRACT

Upper extremity amputations have historically been a serious problem for agricultural workers. The "farmer's hook" is an all too frequent sight at many farm meetings. Over the years, the lack of financial and technical resources and isolation from comprehensive rehabilitative services have resulted in many farmers with upper extremity amputations making modifications to their prosthetic devices or worksites in order to complete essential farm tasks. This paper, based on interviews and surveys of active farmers using upper limb prosthetic devices, will discuss frequent problems experienced by this group and some of the solutions they have developed to overcome these problems.

INTRODUCTION

Even though there is not specific data on the number of farmers with upper extremity amputations, the scope of the problem is partially reflected in several sources of data. According to the National Safety Council, agriculture was the second most dangerous occupation in the United States in 1984[1]. Farmers are more than four times more likely to experience a disabling injury than the average American worker[2]. The National Center for Health Statistics reports that a higher proportion of farmers are missing entire fingers or toes due to amputations than the non-farm population (14.8/1000 versus 8.8/1000)[3]. A recent Purdue study on the nature and proportion of physical impairments among Indiana farm operators indicated that 5.1 percent of the operators surveyed had experienced an amputation[4]. Records at Purdue also revealed that for a 25 year period, when corn pickers were the predominate type of harvester in the state, over 100 farmers per year lost a hand or arm due to corn picker accidents alone. The corn picker has now been replaced by augers and round balers as leading causes of hand amputations.

The Breaking New Ground project at Purdue University, established to assist farmers with physical handicaps, interviewed 20 farmers with arm amputations to obtain information on specific problems they were experiencing in completing essential farm-related tasks and solutions that they have developed to overcome these problems. The farmers were from 11 states and included 17 who were primarily grain farmers and 3 who operated dairy farms.

NATURE OF AMPUTATIONS

Of the farmers interviewed, 10 farmers had below-the-elbow amputations (7 right arm and 3 left arm). Five had experienced an above-the-elbow amputation (2 left arm and 3 right arm). One farmer had an above-the-shoulder left arm amputation and the remaining four had amputations of both arms.

TERMINAL DEVICES CURRENTLY BEING USED

Traditionally, the most frequently prescribed terminal device for farmers has been the "Standard" or Dorrance Hook. Many have referred to it as the "farmer's hook". Of those interviewed, 17 were using the Dorrance Hook and two were using the "Prehensile Hand". One farmer was not using any prosthetic device. The Dorrance Hook operates on the principle that the hand remains closed when the muscles are relaxed and open when making use of the muscles. The "Prehensile Hand", invented by Robert Radocy of Boulder, Colorado, relies on the wearer's muscle use for its gripping strength – it is open when the muscles are relaxed and closed when the muscles and mind coordination cause it to close.

COMMON PROSTHETIC PROBLEMS

The most common problems that the surveyed farmers experienced were in regard to obtaining their prosthesis, learning how to use it to perform various farm tasks, and servicing the device when failure occurred. Isolation from comprehensive rehabilitation services and service personnel was a major problem for most. In some cases, the farmer was several hundred miles from the nearest rehabilitation hospital. As a result, many relied on local craftsmen and their own ingenuity to make the necessary repairs and modifications.

With respect to specific technical problems, those mentioned most frequently, along with suggested solutions, are as follows:

1. Cable Breakage

   The problem most frequently expressed by farmers using either a Dorrance Hook or the Prehensile Hand was frequent cable breakage. To overcome this problem, three farmers with below-the-elbow amputations replaced the cable with nylon rope, which slides through metal loops riveted to the socket and tricep cuff. This arrangement does require an adjustment period to get use to using a rope instead of a cable (Figure 1). Heavier 1/8" cable has also been successful in some instances.

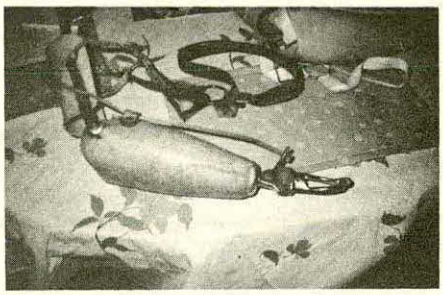

Figure 1

2. Harness Comfort

Another frequently expressed problem with the prosthetic device was with the harness itself. In regard to comfort, seven farmers stated that the harness caused irritation particularly under the arm. Other complaints were related to reaching limitations and the need for a quick release harness, for when the terminal device became caught on something and the wearer needed to let go. One farmer stated that a cow's tail had once gotten caught between the cable and the prosthesis. He reported that he made several revolutions around the cow before he was able to free himself. Fortunately, he was only a little shaken.

To overcome both of the above mentioned problems, Beverly Dailey, of Stapleton, Nebraska, uses a very comfortable quick-release harness, which was constructed with the help of a local saddle maker and her husband. The harness was made out of machine washable Dacron webbing, sheepskin material for underarm comfort, velcro, and part of a bib overall buckle for front attachment and quick release. The harness strap extends behind the back and under the arm, slides through the buckle, is pulled to the desired tension, then folds back and is fastened using velcro material. To quickly release the prosthesis, Beverly simply pulls up on the end strap (Figure 2).

Figure 2

3. Exposure to Cold

The need for a heated or insulated socket has also been indicated by most farmers with below-elbow amputations working in cold weather. For amputees who usually experience a decreased blood flow and decreased sensation in their stump, the problem is amplified. Frostbite to an amputee's stump can occur easily[5].

One farmer reported using a heating pad to wrap his stump in and rewarm it following exposure. Other methods used to cope with the problem included: an electric hairdryer to warm up the prosthetic device; knee-length nylons pulled on before the stump sock to keep perspiration away from the skin; and a tube sock pulled over the top of the prosthesis. One farmer stated that he tried an electric (battery operated) hunting sock, but found that the stump perspired more, resulting in his stump becoming colder.

Work done by Eickman and Mathsen at the University of North Dakota found that the bulk of the heat loss from the prosthesis occurred around the upper-socket region[6]. One of their recommendations was the use of an insulated band around the top of the prosthesis to limit heat loss around the outer polyester cover. Another suggestion, made by Chris Hake of American Orthopedics of Lafayette, Indiana, is if the farmer had two prostheses, he might wish to consider having one made that would use a five-ply stump sock for use in cold weather.

4. Limitations of Terminal Device

Gripping door knobs or other smooth objects with some terminal devices can sometimes be difficult, due to the slippage that can occur. Some farmers have wrapped electrical tape or rubber hose material around the jaws to provide better traction. Many jaws on standard hooks are serrated on the inside. Some jaws are coated with Neoprene, a tough rubber-like material. Chisels that have larger diameter shafts can be difficult to grip with the terminal device. Rubber hose material around the chisels can provide a better gripping surface.

The knife holder on some standard hooks has been found to catch on clothing, therefore five of the interviewed farmers had removed it. One farmer stated that he sharpened the knife holder to cut twine on bales of hay. He also stated that it works well for cutting open bags of feed, silage, and milk replacer. He warned that if it is too sharp it will cut too many things. "A little dull and quick pull works best."

During hay season, handling bales of hay can be difficult for a farmer using a prosthetic device. Rick Banker of Black Creek, Wisconsin, uses a hay hook that was fabricated by a local machinist. To use the hook, Rick unscrews his terminal device and screws the hay hook into the threaded shaft on his socket (Figure 3).

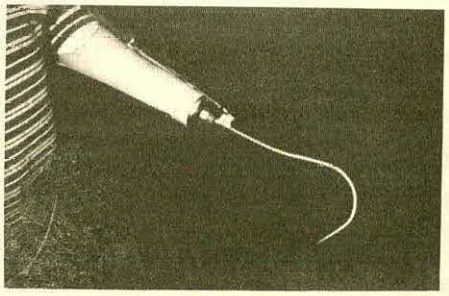

Figure 3

Strength and quality of terminal devices was also raised as a concern. Re-fabrication of certain parts was done to enable them to withstand the more abusive uses found on the farm. One interviewed farmer stated that the wrist unit continued to strip out. Therefore, he had taken the unit to a local machine shop where the machinist had fabricated a new one out of stainless steel. He felt that the stainless steel material was much more durable than the original material.

MODIFICATIONS TO TOOLS AND WORKSITES

In many cases, the terminal device currently being used was not sufficient to enable the prosthetic user to complete essential tasks. Therefore, modifications were made to tools, equipment and buildings as needed. Those mentioned included the following:

1. Modifications to Hand Tools

   Many farmers have expressed some difficulty in using hand tools to complete work-related tasks. To overcome this problem, several modifications have been made.

   Richard Juergens, adaptive equipment consultant, Des Moines, Iowa, designed and fabricated a wrench adapter for one farmer. This adapter fits into a standard Dorrance Hook terminal device (Figure 4). A magnetic nail holder and ratchet wrench will also fit into this adapter.

   Other hand tools that were found useful by farmers using prosthetic devices included various locking pliers, ratchet screwdrivers, the "One-Touch Wrench", "Adjustable Jaw Locking End Wrench", and one-handed grease guns.

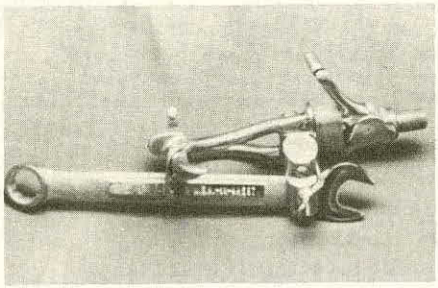

Figure 4

2. Modified Milking Equipment

   Dairy farmers using a prosthesis reported difficulty grasping the claw of a milker unit with the prosthesis while using the good hand to attach the teat cups. Surge makes available a flat C-shaped handle that can be welded to the air divider of the milker unit. This handle can be grasped with the terminal device. For milking parlors, the Surge Randel Milker Support allows the operator to attach the teat cups using only one hand. The milker support arm holds the claw and can be adjusted with one hand at any position under the udder.

3. Modified Vehicle Controls

   The many controls on tractors and combines are very difficult to grasp with a prosthetic terminal device. In some instances, these controls can be relocated so they can be easily reached with the operator's good hand. The Department of Agricultural Engineering at the University of Nebraska-Lincoln relocated the hydraulic control for the hydraulic lift arms on Larry Streff's tractor. This allows Larry to actuate the control using his good left hand rather than with his right prosthesis (Figure 5).

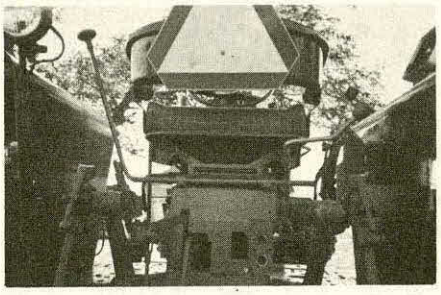

Figure 5

Lever extensions can also be used on existing controls to allow for easier grasping. For example, Lavern Truby of Randolph, Nebraska, has made a longer lever for the throttle on his tractor to make grasping it easier (Figure 6).

Figure 6

Lee Kayhart, a bi-lateral arm amputee, uses a steering ring mounted on the tractor steering wheel. The ring swivels 360° and can be easily grasped with the prosthetic hook (Figure 7).

Figure 7

Murray Bedel from Lebret, Saskatchewan, Canada, also designed and constructed a steering assist ring which can be easily connected and disconnected from any steering wheel. The adjustment shaft is constructed with one piece of square steel tubing and a solid piece of barstock, which slides in and out of the tubing. A spring is placed on the inside of the tubing to provide the needed tension against the steering wheel to hold the steering ring in place.

Merv Copeland from Tomahawk, Alberta, Canada, uses a leg brace extension to operate a hydraulic control on his combine.

4. Building Modifications

Opening and closing gates can sometimes be difficult for farmers who have use of only one arm. Mike Meierhenry, Hoskins, Nebraska, uses a one-handed gate latch, constructed of channel iron attached to a stationary post. The channel iron acts as a gate stop to allow the gate to swing in either direction. To open the gate, Mike raises one or the other gate stop using only one hand.

Climbing grain bins can also be very difficult for farmers with the use of only one arm. Larry Streff uses a back support loop around the ladder of his grain bin. Larry rests his back against the loop while he uses his good hand to open the bin lid (Figure 8). Grain bin stairs are also very useful in overcoming this problem. The user can maintain better balance climbing steps than he can climbing a vertical ladder.

Figure 8

OTHER RESOURCES

The modifications discussed in this paper are only a few of the many modifications being used by farmers with arm amputations who wish to remain active in their operations.

The Breaking New Ground Project at Purdue has recently published a resource manual titled, *Agricultural Tools, Equipment, Machinery and Buildings for Farmers and Ranchers With Physical Handicaps.* This manual contains over 200 ideas currently being used by farmers with physical handicaps which enable them to remain active in their operations. Many of the ideas pertain to farmers with upper extremity amputations. Photographs and detailed descriptions of each modification appears in this manual.

CONCLUSIONS

The various modifications and solutions discussed in this paper are only some of the many attempts that have been tried by farmers and professionals to overcome various problems associated with completing essential farm tasks using a prosthesis. Many of the modifications are homemade to meet one person's need and may not necessarily be effective for another person. They are presented only as ideas to assist farmers, professionals and manufacturers who are still searching for solutions to overcome their specific problems through new designs of prosthetic devices, tools, equipment and machinery that can be used by farmers with upper extremity amputations.

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DESIGN OF INSTRUMENTATION TO QUANTIFY UPPER EXTREMITY MOTOR CONTROL OF CHILDREN WITH CEREBRAL PALSY

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ABSTRACT

The purpose of this project was to develop instrumentation to quantify motor performance of the upper extremity and monitor muscular responses contributing to trunk stability during a reaching task. A 3 position choice response time task was developed as the experimental task and instrumentation was designed to collect temporal and muscular measures using an automated data acquisition system.

INTRODUCTION

Motor control of the upper extremity is a primary rehabilitative objective for children with cerebral palsy. Abnormal reflex activity frequently interferes with the development of efficient motor patterns. Significant conscious effort must be directed toward maintaining trunk stability, thereby decreasing the attention that may be directed toward the performance of upper extremity motor tasks.

In order to improve upper extremity motor control, positioning and postural support have been advocated. To date, research on the effectiveness of these procedures has been limited to clinical observations (1,2) and preliminary studies (4,7). Additional studies have attempted to evaluate muscular action contributing to postural stabilization and upper extremity motor control (3,5). Although authors have suggested the importance of proper positioning and trunk stability on upper extremity motor function, there is little objective evidence to support this premise. One reason for this void of research must be attributed to the lack of instrumentation designed to quantify upper extremity function and monitor the postural adaptations necessary to maintain trunk stability during the performance of reaching tasks.

Speed and time measures have been used extensively in motor control research to quantify performance. Choice response time tasks are commonly used to evaluate efficiency of the motor system because they provide information on both the processing of information and the ability to plan and perform a coordinated movement (6). Response time is defined as the time delay between the presentation of a stimulus and the completion of a specified movement. Response time can be fractionated into reaction time (processing) and movement time (performance) (8). Reaction time, commonly considered as the interpretative component, is calculated from the presentation of the stimulus to the onset of EMG activity recorded from a selected muscle contributing to the action. Movement time is defined as the time required to perform the response and is an indication of the efficiency of the movement once it is initiated. Response time and it's fractionated components as measured for this project are depicted in Figure 1.

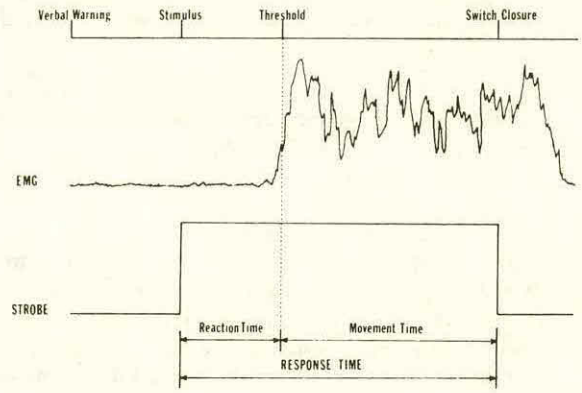

FIGURE 1: Calculation of Fractionated Response Time

PURPOSE

The purpose of this project was to develop instrumentation to quantify motor performance of the upper extremity and monitor muscular responses contributing to trunk stability during a reaching task.

INSTRUMENTATION

Hardware

A choice response time task (3 position) was designed to quantify upper extremity motor function. Instrumentation consisted of a series of pressure sensitive switches (6.5 cm x 6.5 cm) mounted on an adjustable table in a row. This table is placed directly in front of the subject, aligning the center switch with the midline of the body. Distance between switches can be easily modified to accommodate for reach of each subject.

Two computers are interfaced for this instrumentation. This configuration allows one computer to run the experimental protocol and collect temporal data while the other samples EMG. This arrangement simplified software development and maximized the use of computer memory. An Apple IIe computer provides the stimulus by graphically identifying the switch to be closed. A monitor is placed directly behind the experimental task in full view of the subject. Response switches are connected to the Apple IIe through the game port and EMG signals are sampled using an A/D converter. The Apple IIe is interfaced with a Tektronix 4052 computer through the strobe of the game port connected to trigger the Tektronix A/D convertor (Figure 2). Random presentation of stimulus, signals the Tektronix to sample (50 hz) and store three EMG channels and a

marker. Sampled data are collected one second prior to the presentation of the stimulus for a period of seven seconds.

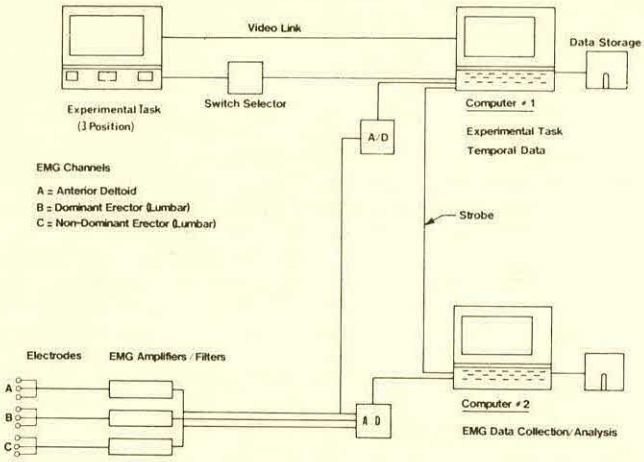

FIGURE 2: Hardware Configuration

EMG activity is monitored from three muscles: anterior fibers of the deltoid, and bilaterally from the lumbar erector spinae. EMG is collected using surface electrodes, amplified, full wave rectified and integrated with a time constant of 50 mS. The signal obtained from the anterior deltoid is used in the calculation of fractionated response time. Prior to data collection the subject is directed to hold the shoulder in a position of 90 degrees of flexion while the Apple IIe samples and averages the amplitude of the integrated EMG signal. The investigator inputs a percentage of this averaged value as the threshold level to be used in the calculation of reaction time.

Threshold level may be adjusted to accommodate for varying experimental conditions. 50 percent was found to be effective when working with children exhibiting cerebral palsy and who are classified as independent sitters. It may be necessary to use a higher threshold percentage when working with children exhibiting increased resting spasticity.

Software

Software developed for the instrumentation was designed to collect and analyze temporal and EMG data. Software written to run the experimental protocol contained subroutines to (1) input demographic data, (2) sample, average and compute the threshold level, (3) randomly present the stimulus, (4) strobe the Tektronix computer and (5) calculate and store temporal data. Time was computed using an assembly language program that ran transparent to the main program. EMG data were collected utilizing software designed to acquire and store multiple channels of integrated EMG.

RESULTS

To date this instrumentation has been piloted with children exhibiting spastic cerebral palsy and who are classified as independent sitters. Temporal and EMG measures for a 6-year-old child exhibiting cerebral palsy are presented as exemplary data obtained using this instrumentation. Mean data for each switch position (center, nondominant, and dominant side) are presented in Table 1.

Figure 3 is a representation of integrated EMG obtained from the anterior fibers of the deltoid, and dominant and nondominant erector spinae (lumbar) during a reaching task (reaction time = 0.84, movement time = 0.54, and response time = 1.38 sec.) These data provide the investigator with information which facilitates a descriptive analysis of the experimental task. It can be observed that the anterior fibers of the deltoid exhibit a distinct activation and that both lower back extensors provide trunk stabilization during the performance of the reaching task.

Position	Reaction		Movement		Response	
	X	S.D.	X	S.D.	X	S.D.
CENTER	0.65*	0.35	0.47	0.32	1.13	0.39
NONDOMINANT	0.70	0.47	0.46	0.37	1.16	0.68
DOMINANT	0.71	0.51	0.32	0.33	1.04	0.50
TOTAL	0.69	0.43	0.42	0.34	1.11	0.52

TABLE 1: Sample Temporal Data: Mean Times for Each Switch Position

*seconds

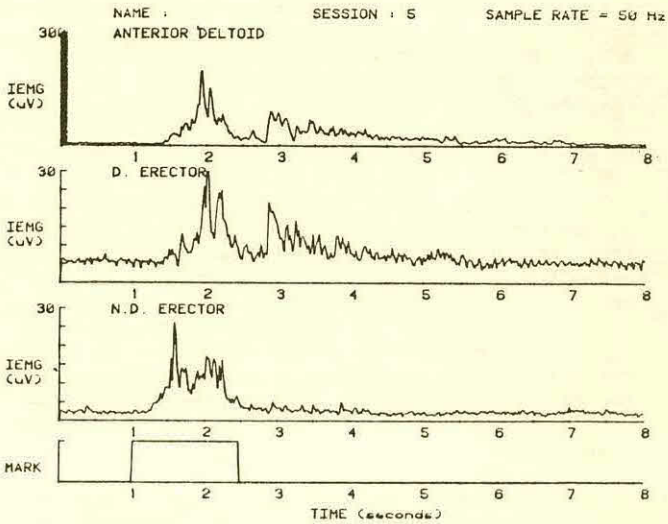

FIGURE 3: Integrated EMG and Marker

Integrated EMG (IEMG) data is analyzed for selected temporal and amplitude components using a peak detection program. Each channel of IEMG data is compared to a baseline obtained prior to the experimental session. This method of analysis allows for the identification of periods when the amplitude of activity is above a selected threshold. Figure 4 is an example of the data

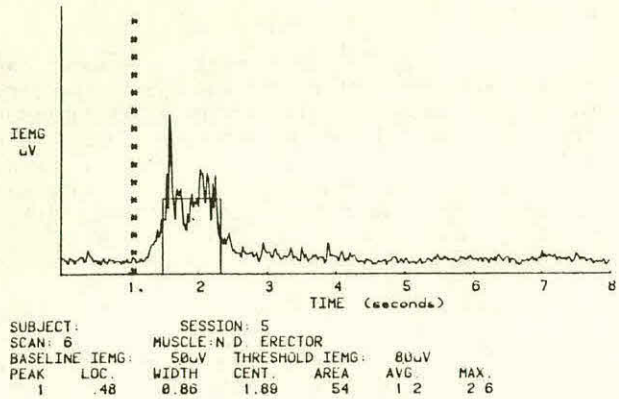

FIGURE 4: Peak Analysis of Nondominant Erector

obtained from the nondominant erector spinae during the performance of the experimental task.

This analysis provides the investigator with valuable information when attempting to quantify the muscular action contributing to trunk stability during the performance of a reaching task. In the example presented the IEMG threshold of the nondominant erector spinae was achieved 0.48 seconds following the presentation of the stimulus and remained at or above this threshold for a period of 0.86 seconds. When compared with the activation of the anterior deltoid (reaction time) these data provide valuable insight into the coordination of the muscular actions contributing to trunk stability during the performance of a reaching task. In this example above threshold IEMG activity was observed in the back extensors prior to above threshold activity of the anterior deltoid indicating stabilization of the trunk prior to activation of the muscles of the shoulder.

FUTURE DIRECTION

This instrumentation is presently being used to evaluate the effects of selected seat inclinations on upper extremity motor control of children with cerebral palsy. Future plans call for incorporating this technique as one component of an extensive study on the effects of adaptive seating on functional ability of children with cerebral palsy. Additional studies investigating the developmental acquisition of trunk stabilization for upper extremity motor control are also planned.

ACKNOWLEDGEMENTS

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QUALITATIVE ASPECTS OF FITTS TASK PERFORMANCE BY MOTOR-DISABLED INDIVIDUALS

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ABSTRACT

Motor assessment of 11 motor-disabled subjects has been performed as part of a project on prescription of non-vocal communication systems. The results reveal a systematic dependence of movement time on distance; a substantial time-independent component; and a strong and sometimes regular variation of movement speed with direction.

INTRODUCTION

During the last three years, the authors and their colleagues have been engaged in an NINCDS-sponsored research project aimed at developing a computer-assisted prescription system. This system will guide the clinician through a sequence of assessment procedures and generate scores which are representative of different aspects of the match between a client and each of the devices under consideration for him/her. One of these scores is a prediction of communication rate, or, more precisely, of the ceiling imposed on communication rate by the user's motor abilities and the operational demands of device. Motor-Determined Maximum (MDM) Rate is calculated by a program, described in detail elsewhere [1], which uses a model of user motor performance to estimate the time required for the particular client to execute each of the movements required to produce a standard body of text using the device in question. The motor performance model is derived (in software) by linear regression applied to data collected during an instrumented motor assessment battery. It takes the form of an equation which relates movement time to the values of the task variables which define the assessment.

This article will be devoted to assessment and modeling of movements relevant to single-body-part use of "keyboard" devices, i.e. those which have as their input interface a planar array of keys or locations on a touch panel. For this type of system, the current version of the assessment requires performance of a sequence of two-point tapping tasks. The controlled variables whose values are changed from task to task presently include movement distance A, movement direction, and target width W.

The purpose of modeling of motor performance in the Tufts-MIT Prescription Procedure is to characterize the individual client in a concise predictive way for the purpose of calculating MDM Rate during clinical device prescription. The authors incorporated this sequence -- abstract assessment, modeling, rate prediction -- in the Procedure based on the following reasoning: It is unarguable that substantial differences exist among presently available NVC devices in terms of their size, key spacing and size, language menu layout, and selection code (if any). It is also indisputable that for a particular client some devices may be ruled out by limits on motor performance, e.g. too small a range of motion for a large electronic lap board or too much tremor for small keys. What the authors have assumed in addition is that among those devices which a client can use, differences in device characteristics can result in significant differences in communication rate. Data presented elsewhere by the authors [2] tends to support this rationale.

In addition to their hypothesis that device-to-device differences can cause important differences in rate, the authors have assumed that user-to-user differences are sufficient to warrant individual assessment. It is important to note the implication of the extreme alternative hypothesis: If all motor-disabled users could be represented by a single model (for the relationship between movement time and task variables), then the outcome of the present research would be the derivation of that model as a universally applicable template for system designers rather than a methodology for extracting it from individual assessment data in the clinic. In fact, the authors' results to date -- presented in the body of this paper -- suggest an intermediate outcome. Models derived from assessment data do differ qualitatively from subject to subject, in some cases in irregular ways which necessitate individual clinical assessment. There do appear, however, to be some patterns which hold for some or all subjects, i.e. regularities to the dependence of movement time on task variables which could assist the device developer in need of a rational basis for optimizing design. The major purpose of this article is to summarize these patterns to assist the designer.

METHODOLOGY

The results presented below have been generated by experiments conducted with 11 disabled and 16 able-bodied subjects. All subjects were cognitively intact adolescents or adults. Most are adventitiously non-vocal as a sequel of ALS, head injury or other conditions, and have unimpaired use of language. Subjects were tested in experimental sessions lasting one to two hours; up to three such sessions were required to complete the desired protocols. Several task sequences were used depending on the subject's abilities and endurance and on the particular goals at the time of the experiment. With the exception of two subjects whose protocols called for repetition of a set of ten tasks three times on each of three successive days, most disabled subjects performed a non-repetitive sequence of 36 tasks. Able-bodied subjects participated in a longer protocol including 112 (distinct) tasks.

Each assessment task consists of reciprocal tapping of a pair of targets with ten movements in each direction. Subjects are instructed to go as fast as possible without making an error. Details of the testing protocols may be found in another publication [1]. The two-target tapping task will be recognized as variant of the task used by Paul Fitts [3]. The authors' task differs in some respects from Fitts' original protocol in that movements are required in directions besides lateral, and in that targets are circular rather than one-dimensional zones which set no limit orthogonal to the movement. It will be seen that, as has been the case in many studies utilized similar but not identical tasks, the original Fitts' Law model did not prove to provide the best fit to the data.

Data from each subject's assessment was processed separately using multiple regression. While a number of mathematical forms have been and continue to be evaluated, at present data from each direction is modeled separately using an equation of the form --

$$T = a + bA + c(1/W), \qquad (1)$$

where $T \equiv$ movement time.

The protocol which includes 36 tasks, for example, yields 720 data points which contribute to the model, so that the number of statistical degrees of freedom is substantial. This analysis and all other statistical manipulations in this work have been performed using a highly capable general purpose software package called SYSTAT.

In order to enhance our understanding of the qualitative features of our subjects' motor performance and suggest appropriate forms for mathematical models, polar plots of their movement times were prepared. Radial distance represents average movement time in seconds for the ten movements performed in one direction in a task, while angle displays the required direction of movement. Typically, three or four contours are drawn, each for a fixed value of movement distance. Target size is held constant over a family of curves. The advantage of polar plots is that the regularity or asymmetry of the dependence of time on distance and direction can easily be appreciated by visual inspection.

RESULTS AND DISCUSSION

1- Time increases significantly with distance. Figure 1, for example, shows a reasonably regular increase in time as distance is increased from 1.5 to 6 inches.

Over the group of seven disabled subjects for whom we have prepared polar plots, roughly half showed one or more "cross-overs", i.e. points for smaller distances with larger values of time. While these irregularities occurred with greater frequency for disabled subjects than for able-bodied (roughly 10% of a representative group of plots), the overall pattern is still a well-behaved dependence of time on distance. This is significant in that it indicates that subjects in the two-target tapping tasks do not

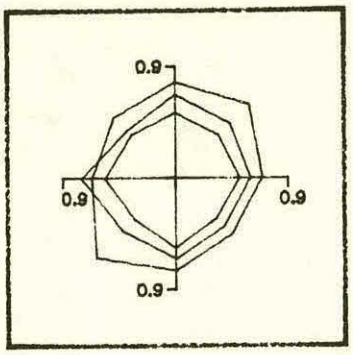

Figure 1. Disabled Subject AK; A = 1.5, 3, and 6 in; W = 0.5 in.

simply find a comfortable rhythm and maintain it independent of distance. Examination of the models derived for movement directions in which subjects' data shows a "crossover" commonly show relatively low statistical significance for "b", the coefficient for A in e.g. (1). This indicates that rather than a systematic phenomenon in which smaller distance take more time, T simply depends less strongly on A than for other data sets, and the anomaly is the result of random fluctuation in movement time.

2- A major portion of movement time is distance-independent. Figure 1 also demonstrates this generalization which applies to all our disabled subjects. It will be seen that as A increases from 1.5 to 6 inches, time increases by far less than a factor of 4. In fact, the data chosen for the figure seems to display a nearly logarithmic dependence of T on A in most directions, i.e. equal increments in T for each factor of two increase in A. This is of course consistent with the classical Fitts Law model for two-point tapping performance --

$$T = a + b\log(2A/W). \qquad (2)$$

As noted above, however, this model was not generally the one which yielded the best measures of goodness-of-fit to the data for our subjects. In our model, defined above, the constant term is typically equal to the time required to move an additional ten inches.

3- Figure 2 displays a particularly clear

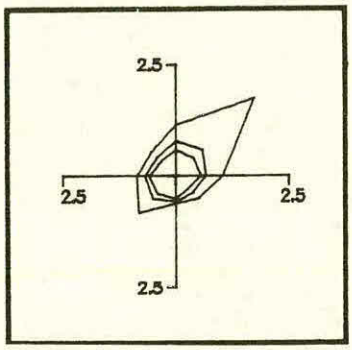

Figure 2. Disabled Subject AC; A = 3, 6, and 12 in; W = 0.5 in.

example of a dependence of movement time on direction which is uniform across values of distance and target size (not shown). Since the long axis of these elliptical contours, i.e. the pair of directions along which the slowest movements occur, is the lower left/upper right diagonal, a simple and appealing interpretation can be made in purely biomechanical terms: This individual performed the assessment with her left hand; she is slowest in the directions which require nearly pure shoulder motion and fastest in the orthogonal direction for which angular changes at the elbow suffice. One can easily demonstrate how much more rapidly the forearm can be moved reciprocally than the whole arm. This easily described and interpreted effect of angle was much more frequently seen in data taken from our able-bodied subjects -- although less dramatically -- than in our disabled subjects' data. See Figure 3, for example. Most of the latter showed asymetries in their contours, but they were not, in general, constant across A and W, and could not readily be interpreted.

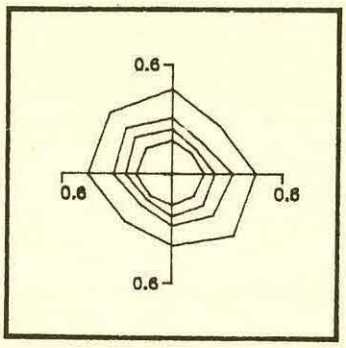

Figure 3. Able-Bodied Subject LEV; A = 3, 6, 12, and 24 in; W = 2.0 in.

The most vivid example in our data of a subject whose data did have an easily explained shape is shown in Figure 4. This subject is a head stick user and was roughly five times faster laterally than in anterior-posterior movements. The test panel was mounted in the plane of this subject's word board (perpendicular to his body) so that the fore-and-aft movements require a complex

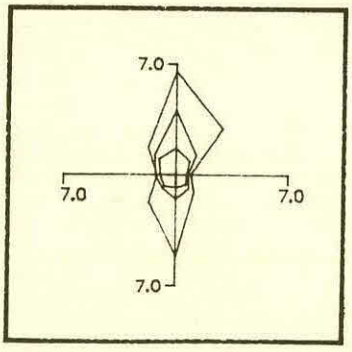

Figure 4. Disabled Subject AE; A = 3, 6, and 12 in; W = 0.5 in.

combination of neck extension/flexion, forward head thrust, and upper torso movements. Relative to the geometrically simple, low inertia head rotation about the long axis of the neck, the lateral movements could be expected to be fast.

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DESIGN OF A NEW SYSTEM FOR THE MEASUREMENT OF HEAD MOTION

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ABSTRACT

A simple method is presented here for the measurement of the range of the head motion for patients with neck injuries. Although the complete motion of the cervical spine is a complex process, a simple model involving three angular measurements was chosen as adequate for comparisons with normal ranges of motion. The designed system involves the use of a hemispherical shell mounted with the help of a frame over the head of the patient. Two light sources are mounted on the patient's head and the readings are taken from circular scales marked on the outer shell surface. The proposed method is simple, relatively inexpensive and can be used in a clinical setting. Moreover, readings for all three primary motions can be obtained for any given position of the head.

INTRODUCTION

Accurate measurement of the range of motion of the cervical spine is of interest in the treatment and rehabilitation of patients with neck injuries. To evaluate such patients, it is important to determine and compare the remaining range of motion with normal ranges of motion. The objective here is to present the development of a new device to measure the head motion. The data obtained on the ranges of motion for patients with neck injuries will be useful in the individualized design of instruments such as headstick operated keyboard or keyboard like devices.

BACKGROUND

The measurement of head motion presents several problems. This is due to the fact that the complete motion of the head and neck system is provided by 23 separate articulations in the cervical spine. None of the cervical vertebrae have simple hinge joints. Each bone in the cervical region of the spine acts in varying degrees as a hinge and sliding joint adding complexity to the mechanism. Measurement of the head motion is further complicated due to the difficulty in the selection of good reference points for the relatively spherical head.

Recognizing these difficulties, the kinematic model considered here consists of three angular motions. These motions as shown in the Figure 1 are flexion or extension in the sagittal plane, lateral bending in the frontal plane and the rotation of the transverse plane.

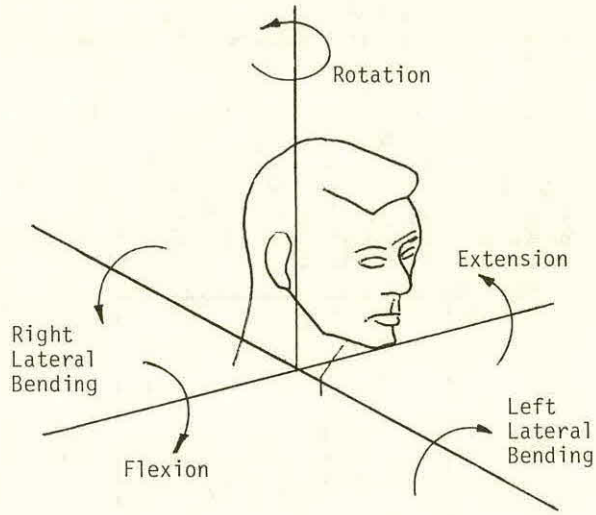

Fig.1 Angular Movement of the Head

Traditionally, several methods have been used to measure the head motion. Defibaugh [1] has done an excellent review of the various methods. These methods are either too complex or do not provide the measurement of all the three degrees-of-freedom at any given position of the head. One existing technique involves the use of X-rays. While this method is very accurate, it is expensive and cannot be used in a clinical setting. Results for any one position of the head would take a very long time to compile. Furthermore, excessive X-ray exposure could be potentially dangerous to the patient. Another method involves the use of a mouth-piece to which is attached a three axis system [2]. The patient sits in front of a gridded wall and digital photographs of the patient in various positions are taken. These photographs are then analyzed using trignometry and the ranges of the head motion are then determined. This method is also relatively accurate, but, like the X-ray technique, somewhat expensive, difficult to perform in a clinical setting and results would take a long time to compile.

There is a more popular method using goniometer as described by Kottke and Mundale [3]. This method is simple to use but the technique introduces inaccuracies particularly for the measurements in the sagittal plane since there are no landmarks on the side of the head. Only two primary motions can be read at a time and the measurement of rotation involves a separate procedure.

In order to meet our objective, we decided to design a system which would have certain features. The primary goal is to produce a method which would measure all three head motions at any given position. Any instrument attached to the head will have to move freely and be extremely lightweight so that there would be no interference to the natural head motion. This criteria is important because our objective is to use this device with injured and handicapped individuals whose necks may be weak. The method should be fairly accurate and easy to implement in a clinical setting. Finally the system should be relatively inexpensive.

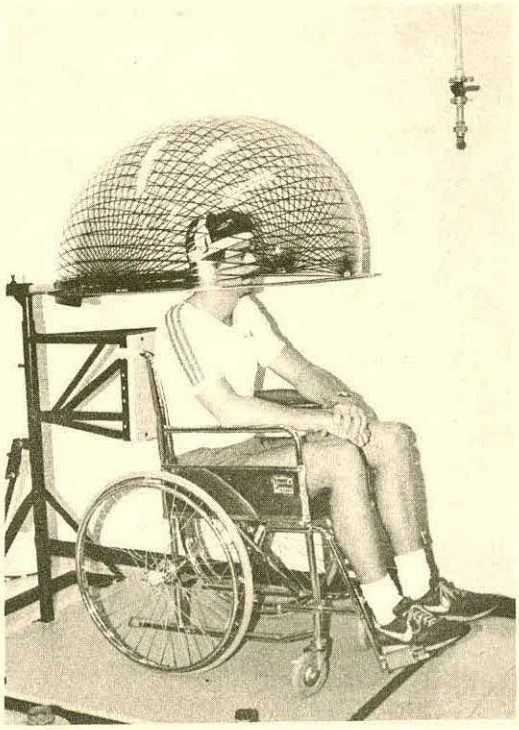

Fig.2 The System With the Hemispherical Shell and the Patient

SYSTEM DESIGN

With the above objectives in mind, the designed system mainly consists of a frame, a hemispherical shell and a light source (Fig. 2). The hemisphere is the primary functional element of the system. It is made of clear Plexiglas and is 36 inches in diameter. Several circular scales are taped on the hemisphere which enables the hemisphere to be used as a goniometer. The inside surface is scuffed or painted with a semi-opaque material so that a point of light from the inside can be easily located on the outside surface. The surface of the shell is marked with two series of arcs or lines called alpha-lines and beta-lines. These are shown in Figure 3. These lines, marked as two protractors, represent degree measures and are continued across the entire shell surface. These arcs are drawn about two perpendicular axes which intersect at the center of rotation of the head motion and lie in the horizontal plane.

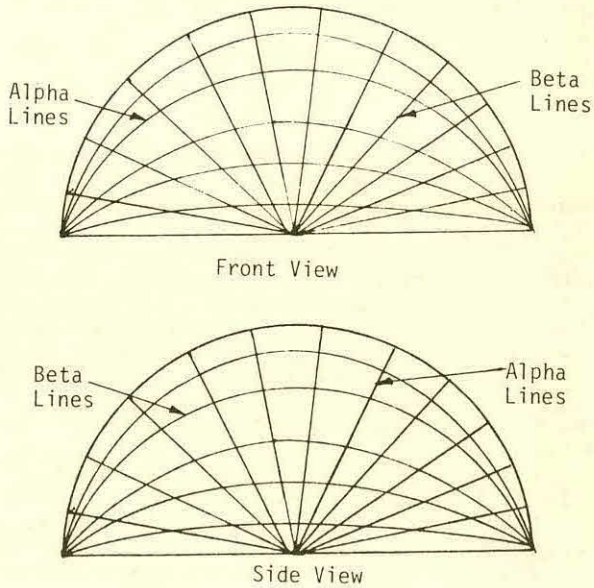

Fig.3 Hemispherical Shell Showing the Marked Scales

The light source consists of two lights and a head strap. The lights are placed on the top of a light weight head strap in such a way that when the head strap is placed on the head of the patient, the lights point radially outward from the center of rotation of the head.

The frame is made of steel and plywood and is used to support the plastic hemisphere above the patient's head. The frame allows the position of the hemisphere to be altered to accommodate different sized individuals. The patient whose head measurements are to be taken sits on a wheelchair which is mounted on the bottom of the frame. The frame is also used to support the back of the patient to restrict any unwanted motion. For easy mobility the frame is designed so that it can be easily folded into a compact size. The hemisphere can be removed to protect it from damage.

METHODOLOGY

The patient sitting on a wheelchair or a regular chair is moved to the frame and the chest of the patient is strapped to the back support of the frame. The head strap with the light source is placed on the patient's head so that the two lights are equally spaced from the centerline of the patient's head. The patient is then asked to hold his head in a position which he feels is a normal vertical position. At this stage, two readings, one on the alpha scale and the other on the beta scale are taken for each light source. This gives a set of four readings for one position of the head.

If the maximum flexion in the sagittal plane is to be determined., then the patient is asked to bend his head in a forward position as far as possible. Again a set of four readings are taken from the two scales on the shell surface. Analytical equations have been developed [4] to convert the data read from these two scales into angular rotation of the head. Space requirements do not permit the inclusion of these equations in this paper. These equations have been programmed into a microcomputer to obtain a quick printout of the results. A similar procedure is performed for each of the five other primary positions of the head. This information is then used to determine the maximum ranges for the three angular motions of the head for any patient.

CONCLUSIONS

Presented in this paper is a simple system for the measurement of the range of head motion for patients with neck injuries. The proposed method enables the simultaneous measurement of the three degrees-of-freedom of the head motion. The method uses two light-weight light sources mounted on the patient's head and the readings are taken from two circular scales marked on a spherical shell which is mounted with the help of a frame over the patient's head. The method is relatively inexpensive, fairly accurate and can be used in a clinical setting. The information obtained on the range of motion of the cervical spine is useful in the treatment and rehabilitation of patients with neck injuries. This data is particularly helpful in the custom design of instruments such as headstick operated keyboard or keyboard like devices.

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NORMS FOR HAND PERFORMANCE DIFFERENCES

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Abstract

Utilizing the Available Motions Inventory (A.M.I.) a normed data base is generated to examine within subject hand performance differences. A three-way ANOVA is used to determine the impact of gender, hand preference, and position upon performance. All three were found to have a significant main effect. Hand preference with gender and hand preference with position also show a two-way interaction effect. Considerations for task design and modification, especially concerning position variables, are discussed.

Introduction

A major goal of rehabilitation engineering is to apply sound technological, quantitative, and otherwise scientific methodologies to the problems of functional limitations encountered by individuals with disabilities. These problems can most generally be described as a lack of congruence between an individual's ability profile and the ability profile inferred from the task design. Research problems in rehabilitation engineering parallel closely classical problems in the broader field of human factors engineering. In the work environment, for example, appropriate design and modification of tasks and equipment that will utilize human physical abilities most efficiently is a primary objective of the engineer. This is true whether designing for employees with or without disabilities.

Assuming that most tasks and devices are designed with the "able-bodied" population in mind, task adaptation must start with the basic concern as to how does a specific individual differ from the original design assumptions. Therefore, this study utilized the current A.M.I. data base of able-bodied individuals in order to examine the specific relationship between three potentially significant independent variables in A.M.I. performance: gender, hand preference, and position (relative position of the task to the individual and the test frame).

Method

Subjects

Subjects for the study consisted of 92 able-bodied adults, exactly half of whom (46) were male and other half female. A total of 54 subjects ranged in age from 20 to 29 years old; 24 were between the ages 30 and 39; and the rest were between 40 and 49 years of age. Within each age group, there was an even distribution between males and females.

Device

The device used in this study was the Available Motions Inventory test package (Malzahn, 1980). The Available Motions Inventory (A.M.I.) has been shown to be an effective and reliable instrument for describing the functional ability of the upper extremities and thereby a useful tool in task design and modification (Malzahn, 1984; Rahimi and Malzahn, 1984). The A.M.I. consists of a total of 142 manual industrial tasks (71 repeated for each hand) that are presented within an adjustable work station. There are six major subtest groupings of the A.M.I.--Switches, Settings, Rates, Strength, Assembly, and Reaction-Reach. Switches, Settings and Assembly tasks are primarily concerned with accuracy of movement while Rate and Reaction-Reach have a strong speed of movement component. And, of course, Strength measures force of movement.

The position variable, mentioned earlier, consists of 5 distinct positions relative to the subject. They are: CLH (Center, Lower, Horizontal); SLH (Side, Lower, Horizontal); CLV (Center, Lower, Vertical); CUV (Center, Upper, Vertical); and SUV (Side, Upper, Vertical). Position CLH is a horizontal work surface at seated elbow height with the center of the task within 75% of reach. SLH is also a horizontal work surface at seated elbow height located 20 inches lateral to the CLH position. Conversely, CLV is a vertical work surface at 90% of reach and 6 inches above seated elbow height. Position CUV is similar to CLV except that it is 18 inches above elbow height. Finally, the SUV work surface is 45 degrees from the frontal plane and 20 inches lateral to the CUV position.

Procedure

Hand preference for each subject was empirically determined by averaging the performance scores for each hand separately. The hand with the higher score was labeled as the superior hand and the lower score the inferior hand, thus eliminating the need to rely on verbal report of hand preference. Raw scores for the superior hand were normalized (mean = 0.0 and standard deviation = 1.0). Raw scores from the inferior hand were trans-

formed using the mean and standard deviation of each task achieved by the superior hand.

Data collected from the A.M.I. procedure were analyzed by the BMDP-P2V mainframe package (Dixon and Brown, 1979). A three-way analysis of variance with repeated measures design was specified. As mentioned earlier, three factors were examined. Gender (male, female) was treated as a between subject factor while hand preference (superior, inferior) was treated as a within subject factor. The third factor, also within subject, was position.

Results and Discussion

Results from the three-way ANOVA are summarized in Table 1. Gender showed a significant main effect upon performance where $F(1,90) = 7.54$, $p = 0.0073$. This revealed that, overall, males performed higher on A.M.I. tasks ($z = 0.048$) than did females ($z = -0.2573$). Previous A.M.I. research has shown that the difference in gender is most significant in those tasks which involve hand velocity and that both males and females performed equally in those tasks requiring accuracy (Rahimi, Malzahn, and Musa, 1985).

Table 1

Analysis of Variance Summary Table

Source	SS	df	MS	F	Prob.
Mean	10.07	1	10.07	3.54	0.063
Gender	21.46	1	21.46	7.54	0.007
Error	256.05	90	2.85		
Hand	10.07	1	10.07	153.47	0.000
HxG	0.27	1	0.27	4.04	0.047
Error	5.90	90	0.06		
Position	1.39	4	0.35	4.47	0.001
PxG	0.19	4	0.04	0.62	0.649
Error	27.97	360	0.07		
HxP	1.38	4	0.34	7.57	0.000
HxPxG	0.05	4	0.01	0.31	0.871
Error	16.51	360	0.04		

Hand Preference also revealed a significant main effect on performance, $F(1,90) = 153.47$, $p < 0.0001$. Comparison of the two levels showed that subject's superior hand out-performed their inferior hand. Recalling that the normalized scores are based on superior hand performance, mean z score for the superior hand was equal to 0.00. For the inferior hand, z was equal to -0.2093. This finding is consistent with previous literature on hand preference involving a variety of performance criteria (e.g., Annett et al., 1979; Flowers, 1975).

Position was also shown to have a significant main effect upon A.M.I. performance, $F(4,360) = 4.47$, $p = 0.0015$. A post-hoc comparison of cell means, utilizing Tukey's method, for the five positions was performed. This analysis indicated that CLH and SLH differed significantly ($p < 0.05$) with subjects performing highest in the SLH position ($z = -0.0451$) and lowest in CLH ($z = -0.1661$). All other pairwise comparisons were non-significant ($p = 0.05$). Position SLH is between 40 and 45 degrees from the mid-saggital plane (as opposed to CLH which is directly in front of the subject) and is actually more distal to the subject than is CLH, yet position SLH scores are the highest of any position and CLH are lowest.

One possible explanation for this phenomenon could be that as the degrees of freedom in motion are restricted in the SLH position (shoulder and elbow tend to be more fixed) the efficiency of motion in the remaining body members would rise. Certainly, this finding warrants further research into task criteria and relative position to the body.

Hand Preference and Gender were analyzed together as source of variance and revealed a significant interaction effect, $F(1,90) = 4.04$, $p = 0.0475$. This interaction was analyzed to determine the simple effects of Hand Preference and Gender. First, in the case of Hand Preference, at neither level (superior or inferior), was there a significant difference between males and females ($p < 0.05$).

Table 2

Hand Preference x Gender

Mean Scores

	Superior	Inferior	Difference
Male	0.1358	-0.0396	0.1754
Female	-0.1358	-0.3790	0.2432

However, there were differing degrees of significance between the superior and inferior hand with regards to Gender. Males showed a significant difference between hands, $F(1,90) = 5.38$, $p < 0.001$. Females showed a greater difference where $F(1,90) = 10.36$, $p < 0.001$. Table 2 shows the relationship between the factors Gender and Hand Preference.

A significant two-way interaction for Hand Preference and Position was also indi-

cated, $F(4,360) = 7.57$, $p < 0.0001$. Subsequent simple effects analysis revealed that there were significant differences in Position for the inferior hand, $F(4,360) = 15.37$, $p < 0.001$. Pairwise comparison of cell means showed that, like the main effect for Position, SLH and CLH represent the respective upper and lower boundaries of the range of means. Furthermore, both SLH and CLH differ significantly from all other positions as well, $p < 0.05$ (see Table 3). Again, no other pairwise comparisons were significant, $p = 0.05$. The above finding confirms similar results of Position as a main effect and also indicates that this trend becomes more dramatic when the performance of the inferior hand is considered.

Table 3

Hand Preference x Position

Mean Scores

Position	Hand Preference Superior	Hand Preference Inferior
SLH	0.0	-0.0902
SUV	0.0	-0.1911
CLV	0.0	-0.2014
CUV	0.0	-0.2316
CLH	0.0	-0.3323

Further simple effects analysis of hand preference at each position revealed that there were significant differences between the superior and inferior hands ($p < 0.01$). However, there was no significant difference between hands at position SLH $F(4,360) = 2.07$, $p < 0.01$. Therefore, in position SLH, subjects generally perform better than in any other position while doing so efficiently with either hand.

As can be seen in Table 1, there was no three-way interaction between the independent factors.

Summary

Norms were developed to aid in the evaluation of manual ability. The relative impacts of gender, position, and hand preference were documented. Males performed 0.3 standard deviations better than females. The greatest difference between superior and inferior hand performance occurs in the CLH position and the least difference in the SLH position. The difference between CLH and SLH is 0.24 standard deviations. The average difference between the superior and inferior hands was 0.21 standard deviations. This data allows the quantitative comparison of the relative level of hand superiority demonstrated by an individual and the basis for further investigation of the impact of task positioning on performance.

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12.1

DEVELOPMENT OF AN ADVANCED ROBOTIC AID: FROM FEASIBILITY TO UTILITY

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ABSTRACT

Many severe physical disabilities, such as quadriplegia, are characterized by a greatly reduced capacity for manipulating the environment, a condition often exacerbated by reduced mobility. The Veterans Administration is sponsoring a continuing effort to develop a robotic manipulation aid that can be controlled by severely disabled individuals to provide functional restoration of lost motor skills. Hardware and software systems have been implemented for a mobile manipulator and its stationary control station.

INTRODUCTION

The psychological and economic importance of access to, and control of, one's surroundings has long been recognized by the disabled and by rehabilitation professionals[1,2,3]. This need can only be partially addressed by human attendants, trained animals, and the use of specific technology-based solutions such as environmental controllers[4]. While robots can also offer only a partial solution, their true value lies in their ability to serve as a general-purpose manipulation tool, capable of being applied in changing or unpredicted situations, capable of mobility, and under the complete command of the user.

THE INCENTIVE

At this time, effective care for physically disabled people is labor intensive. The incentive to apply robotics technology is thus economically similar to the situation in industry. The cost of human labor averages $15/hour and is increasing. Robot costs average $5/hour and are decreasing[8]. Furthermore, robots are most effective when applied to multiple shifts per day as would be the case for robots in human service applications.

Kalsbeek, et al.[6] estimate that the average cost per year (44% direct costs; 56% indirect costs) for all *new* head and spinal cord injury cases in the U.S. is $2.4 billion. In 1983 there were over 16,000 cases of spinal cord injury treated in the VA alone; with a direct cost of $1.5 million over the estimated lifespan of each of these individuals, the total cost to the VA will be $4.5 billion[5]. Is it worth investing in research and development? Bowe[7] has estimated that every dollar spent for rehabilitation research returns $11 in cost benefits to society.

THE PROBLEM

The feasibility of our approach to robotics has been demonstrated through the development of two first-generation prototype Robotic Aids[9], one for clinical evaluation and one for technical evaluation. Each is composed of a Unimation PUMA-250 arm that can be controlled by spoken commands, a joystick, stored programs and rudimentary hand sensors. Clinical evaluation and analysis of the capabilities and limitations of the system[10] have helped us define the areas of research and development we are now pursuing to move robotics from *feasibility* to *utility*. Specific problems are outlined below as elements of the current research and development program:

1. A communication manager is needed to facilitate human-machine interactions by sustaining an effective "conversation" between the user and the computer.
2. Sensor-driven reflex control loops are needed to reduce the operator's work load and assure rapid response to grasp and obstacle-avoidance problems.
3. Intelligent motion planning is needed to reduce the system's dependence on a highly structured environment, and to reduce the operator's work load.
4. A vehicle is needed to extend the Robotic Aid's working volume beyond the fixed tabletop environment.
5. Instrumentation and a formal strategy are needed to measure utility objectively.

PROJECT OBJECTIVES

To define specific research projects, we first compiled a set of **tasks** from spontaneous, unsolicited comments made by disabled users during training sessions with the first Robotic Aid. We then identified an associated set of **capabilities** that would be required to assist in the performance of these tasks [Table 1]. **Capabilities** were generated by breaking down each task into the fundamental functions requisite to its performance. Finally, we determined the **devices** and **algorithms** needed to implement the desired capabilities. The capabilities are explained below:

1. COMMANDS are discrete transactions between the user and the robot. They are, for instance, words spoken to a speech input system or items selected from a menu.
2. CONTROLS are continuous variables, of controllable magnitude and duration, used by the operator to refine commands and do real-time piloting. The operator could use a joystick, mouse, or head-position detector to move a screen cursor or the mobile base.
3. COMMUNICATION MANAGEMENT is a set of procedures and rules designed to assure that effective two-way communication is maintained between operator and machine.
4. AUTONOMOUS PLANNING is performed by the machine when sensed data are operated on by applications programs, with the result that the machine makes navigating (or equivalent) decisions. These decisions are subject to human supervision and veto. Trajectory planning for the mobile base and switch toggling on an appliance by the robot hand are two examples.
5. GRASP is the property of a robot that allows objects to be selected, positioned and oriented. It is typically associated with "hands" and includes a variety of functional attributes, such as detection of slippage and evaluation of object geometry for stable holding.
6. MANIPULATION is the capability to move objects from one place to another while maintaining a correct orientation of the hand and avoiding collisions with stationary objects.
7. PROGRAMMABLE REFLEXES are designed to protect the operator and/or the robot from adversity. They act quickly and do not require to human supervision.
8. MOBILITY extends the Robotic Aid's working volume beyond the desk-top workstation environment. In this project it also includes vertical "mobility" to facilitate access to floors and shelves.

The laboratory version of the first Robotic Aid was the starting point for these goals. Two general problems that were solved at the start of this phase are the design constraints of the mobile system itself[11], and the requirements of the computing environment[12]. Both of these topics have been addressed satisfactorily to proceed with the problems enumerated above.

CURRENT RESEARCH STATUS

The project is one year into its current phase. Companion papers in these Proceedings discuss individual areas of endeavor within our rehabilitation research project. This section outlines recent results and ongoing efforts.

Human/Machine Interaction

We continue to emphasize speech input as the primary communication medium. A recently acquired Kurzweil 1.1 recognition system, combined with improved software support, should significantly enhance the speed, accuracy and facility with which the Robotic Aid accepts and processes words. Specifically, we are implementing a new computer-based Dialog Management System

TABLE 1

To live independently, one must be able to perform a variety of TASKS. These can be correlated with generalized CAPABILITIES needed to get the job done in minimally structured environments.

- = don't need ○ = should have • = must have

TASK	COMMAND	CONTROL	DIALOG	PLANNING	GRASP	MANIPULATION	REFLEX	MOBILITY
MAINTENANCE								
Food Preparation	•	•	○	○	•	•	○	•
Food Service	•	•	○	○	•	•	○	○
Personal Hygiene	○	•	○	○	•	○	•	○
Personal Grooming	•	•	○	-	•	•	•	○
Clothing Management	○	○	○	•	•	•	○	•
Appliance Usage	•	•	○	•	•	•	○	•
VOCATION								
Storage & Retrieval	•	○	○	○	•	•	○	•
Equipment Operation	•	○	○	○	•	•	•	•
Assembly	•	•	•	•	•	•	○	○
Word Processing	•	○	-	-	○	○	-	-
Computing	•	-	○	-	-	-	-	-
Materials Processing	•	○	○	○	•	•	•	•
RECREATION								
Reading	•	○	-	○	[illegible]	[illegible]	[illegible]	-
Film/Video	•	-	-	-	•	•	[illegible]	•
Performing Arts	○	•	○	[illegible]	•	•	[illegible]	•
Graphic Arts	○	•	○	[illegible]	•	•	[illegible]	-
Sports	○	•	○	•	•	•	•	•
Social Interaction	○	○	•	-	[illegible]	[illegible]	[illegible]	•

which recognizes and analyzes commands given in the form of simple English sentences. "Pick up the cup," for instance, would be translated to appropriate low-level robot-specific commands.

The user is informed, via synthesized speech through a DEC Dectalk unit and a visual display, of the commands passed on to the robot controller. Since the user exercises a supervisory role, the system provides a real-time color image of the internal state of the control system, and the state of the computer's current world model, currently on an IBM PC/AT equipped with a Professional Graphics Display.

Mobility

We have chosen to extend the range of the Robotic Aid by mounting the arm on a unique three-wheeled base that allows the unit complete three-degree-of-freedom mobility: left/right, fore/aft, and rotation are all possible simultaneously. This design is discussed more completely in a companion paper [Van der Loos et al.], and the computer architecture and control software is reviewed in another [Michalowski et al.].

Machine Autonomy

We assert that a degree of autonomy is crucial if the Advanced Robotic Aid is to lessen the control burden on the operator, defined by the amount of time and the degree of focused attention needed to complete a given task. The emphasis is on a "symbiotic" structure, in which necessary control functions are shared between the operator and the robot's supervisory computer. We have identified those aspects of the manipulation process most advantageously assigned to the robotic system. They fall into two categories, **reflexive** and **strategic**:

Reflexive Functions. In the first version of the Robotic Aid, the movements of the arm were determined directly by the user, who had to rely on a single mode of perception —vision— to guide the arm through such elementary operations as gripping and obstacle avoidance. But it is clear that visual (i.e., position) information alone is inadequate for the control of certain important tasks. User fatigue and frustration can be traced, in part, to the fact that direct control of the robot's movement requires constant, conscious attention. Our Advanced Robotic Aid is endowed with the sensory capabilities appropriate to alleviate this burden.

We label as "reflexive" those functions that result directly from sensory stimuli. The user's role is limited to initiating the function and monitoring its successful execution.

Consider the bumper system of the base. While providing physical protection to the base, it also provides contact-point localization and the capability to "bounce-off" any obstacles, given that the user has invoked that procedure, and not the one used to shove boxes across the room. This provides an example of two reflexes appropriate to one sensor system. Two additional systems, an ultrasonic ranging system and a laser-scanner for absolute position detection, are currently under development.

The hand on the robot is equipped with proximity sensors to deal with objects at distances of 1-2 centimeters, for object detection during grasping and collision avoidance during piloting. Additionally, the arm is equipped with a force sensor at the wrist to facilitate tasks such as placing a grasped object on a table and operating a push button. These two sensor systems are discussed in a companion article [Park et al.].

Strategic Functions. The term Strategic Functions refers to manipulative tasks that require some overall knowledge of the environment, and are preceded by a planning phase. Examples are: meal preparation, light assembly, navigation in a confined space. The autonomous execution of such tasks in an industrial setting is the focus of considerable activity within the fields of robotics and artificial intelligence. A rehabilitative application is complicated further since the robot is meant to function in an unstructured, everyday environment. We believe that the participation of the disabled user provides a unique opportunity for overcoming and circumventing some of the more intractable problems associated with environmental modeling and task planning.

In the Advanced Robotic Aid, the user and robot work together to complete a task. The role of the operator is that of a supervisor. Commands are no longer *motion-specific* (for example, "forward",) but *task-specific* ("pour from the bottle into the cup"). To assist the user in strategic planning applications, the Robotic Aid must be endowed with the following specifics, which we are currently implementing:

1. it must be able to *acquire data* about the environment;
2. it must interpret the data in terms of an *internal representation* (or *model*) of the environment;
3. it should be able to *plan* and *execute* simple manipulative motions;
4. it must have a means of *communicating with the user* to present the results of data-taking, analysis, and planned motions;
5. it must be *safe*, in that the user must be able to stop any planned or ongoing motion at any time.

We are applying these general principles to the specific case of manipulating medium-sized objects on a table and maneuvering in a typical room environment.

Evaluation

Since our goal is the increased utility of the Robotic Aid, a formal evaluation process is being pursued to extract qualitative and quantitative measures of success. We maintain a Laboratory System (located at the Center for Design Research at Stanford University) to develop new hardware and software features, and a Clinical System (located in the Spinal Cord Injury Service of the Palo Alto VA Medical Center). Specific evaluation objectives are discussed in a companion paper [Glass et al.], and are summarized below:

1. To train disabled users of all ages in the use of the system. A companion paper [Holloway et al.] describes the training manual produced for the first Robotic Aid. With an advanced system, the features themselves could lead one to predict a higher degree of learning difficulty; however, the concomitantly enhanced user-interface should instead serve to *simplify* the use of the system.
2. To develop applications for the robot. The robot is being studied in specially-prepared environments, using commercially available components such as domestic appliances.
3. To assess device performance under realistic conditions. Every new feature of the Advanced Robotic Aid is being evaluated through a series of time-to-completion studies of selected tasks, including comparison with baseline measurements performed on the first Clinical System.

A major goal at this stage of research is for the laboratory development process to receive feedback from the evaluation effort, tightening the design loop and creating a more useful system for our user population.

CONCLUSION

We believe that robotics technology, carefully and sensitively applied, can help bridge the gap between physically limited individuals and their environment in a cost-effective manner. The current phase of research is advancing on well-defined fronts to shape a useful manipulation tool, one that utilizes the ability of people to organize, plan, see, and supervise, and complements that with the strength of machine sensing, motion, and control.

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COMPUTER CONFIGURATION FOR A ROBOTIC MANIPULATION AID

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ABSTRACT

Researchers at Stanford University and at the VA Medical Center in Palo Alto are developing a robotic device to assist the severely disabled. The computational elements of the Robotic Aid are being expanded and upgraded to accomodate an increasingly sophisticated set of robot capabilities and human/machine interactions. Hardware and software systems have been implemented to control a mobile manipulator and a stationary control station. The real-time control software is modular, to allow for future expansion.

INTRODUCTION

The Robotic Aid Project is an ongoing effort to apply robotics technology to serve the needs of the severely disabled. The goal of the project is to develop a compact, general-purpose manipulation aid that will assist quadriplegics (and other individuals with limited upper limb function) in the performance of activities of everyday living, as well as vocational and recreational tasks[1]. The work is funded by the United States Veterans Administration. Technical development and clinical evaluation are pursued at the VA Medical Center in Palo Alto and in the Design Division of the Department of Mechanical Engineering at Stanford University.

A prototype Robotic Aid has been undergoing evaluation trials at the VA Medical Center in Palo Alto[2]. Its major components are: a PUMA 260 robotic arm, a two-fingered gripper, a Z-80 based supervisory computer; a speech recognition unit with a vocabulary of 70 words; a speech output device; and a small flat panel display. A disabled user can "pilot" the arm in real time by uttering a series of explicit position and velocity commands. Motions can consist of rotations and translations in one of three coordinate systems. Any configuration of the arm can be assigned a name and the arm may be returned to that position at a later time with a single command.

Based on clinical experience, a major effort is under way to expand the capabilities of the Robotic Aid and to make it easier to use. The important elements of this upgrade are: increased participation of the user in the operation of the device, incorporation of sensors into the motions of the robot, enlarged physical range of manipulation, and enhanced autonomy of the robot for selected generic tasks. It has become clear that to achieve the desired goals, a significant increase in computing power is essential. Accordingly, during the past two years, the computer system has been completely replaced and reconfigured[3][4]. This paper describes the most recent configuration.

DESIGN SPECIFICATIONS

- The computer system must support a variety of heterogeneous functions. These may be distributed into two broad categories: real-time hardware control algorithms that need to be performed at a rate of 10 to 100 Hz., and decision-making and user-interaction tasks that occur at 1 Hz. or less.
- The manipulator itself must be mobile. This imposes important physical constraints on the real-time portion of the computer.
- The instantaneous motion of the manipulator results from the combined influence of several motion sources: operator commands, sensor signals and autonomous computations. The computer architecture must therefore allow for concurrent running of separate software processes. The execution of these processes must be coordinated in an efficient way: access to common resources (memory, CPU time, data) must be coordinated.
- The real-time portion of the system must conform to the specifications of the PUMA robotic arm: for smooth motion, joint positions must be provided to the six individual servo controllers at 29 millisecond intervals.
- A variety of peripheral devices must be supported: the PUMA arm, a robotic vehicle, sensors, communications and user-interface equipment.
- The computer must be able to perform the kinematic calculations that are necessary to control a six degree-of-freedom manipulator and a three degree-of freedom vehicle.
- Approximately one dozen collaborators are taking part in the research and development process. Readable and modifiable code must be produced in a standardized development environment.

HARDWARE IMPLEMENTATION

The final system configuration conforms to the above design specifications and a number of realistic constraints: the availibility of funds, space and personnel, plus the need to utilize existing facilities whenever possible.

The complete Robotic Aid consits of two separate systems: a mobile manipulator and a stationary console/worktable. Computational resources are distributed between the components as needed.

The Mobile Robot

The mobile computer system is based on the 22-bit QBUS, which is compatible with a widely-used set of products manufactured by Digital Equipment Corporation: the LSI11 family of 16-bit microprocessors (including the KXT11 single board processor) and the new MicroVAX 32-bit machines. The QBUS was chosen because of the great variety of available peripheral devices and, most importantly, to take advantage of a high-level real-time programming language — MicroPower Pascal.

The configuration of the mobile computer is shown schemetically in the figure. The main processor in an LSI11/73. Five kinds of peripherals also reside on the bus: 512 kB of solid state memory, a sixteen-channel analog-to-digital convertor, a four-channel serial interface (DLV11J equivalent), three bidirectional 16-bit parallel interface modules (DRV11 equivalent), and a DEC KXT11-C single board processor. The timing of events on the bus is controlled by an internal 10 MHz. oscillator. An external 140 Hz. clock interrupt synchronizes the execution of applications programs with the robot's servo controllers.

The serial lines are assigned as follows: the console channel is dedicated to downloading and dynamic debugging of the

real-time software within the context of the MicroPower Pascal system. A second serial line is used to provide bidirectional communication over the 4800 baud FM radio link, allowing for interaction between the mobile and stationary systems. The third serial line provides a connection to an on-board terminal that features a 40-character, 16-line LCD display and keyboard. The terminal displays status and error information and also can be used by the development staff to enter commands to the computer.

Two of the parallel I/O modules are used to interface to the six commercial joint controllers of the arm and three identical wheel controllers of the mobile base. During normal operation, the nine controllers interrupt the main CPU every 29 milliseconds. The LSI11/73 responds with a set of new joint/wheel positions which are then implemented by the servo controllers.

The third parallel I/O module is used as a general-purpose interface to a variety of control and diagnostic functions such as enabling of power to the robot and sensing the state of the 14-segment bumper system that surrounds the vehicle.

The KXT11 microprocessor controls the hand and wrist of the robot. It records readings from the twelve photoelectric proximity sensors that are mounted on the fingers of the gripper and the eight pairs of strain gauges that record forces and torques in the wrist joint. The processed data is transferred to the main system memory via an on-board DMA controller.

The User Console

The stationary component of the Robotic Aid's computer system consists of an IBM PC/AT. It is equipped with an IBM Professional Graphics Controller (with a resolution of 640x480 pixels) and an IBM Color Graphics Adapter (resolution of 320x200 pixels). The PC/AT has serial connections to a DECTALK speech synthesis unit, a Kurzweil KVS1.1 speech recognition system, and a Cattron 4800 baud radio link module. In addition, it is interfaced to an ultrasonic head position detector.

REAL-TIME CONTROL SOFTWARE

MicroPower Pascal is a dialect of Pascal that also includes operating system features. The code for running a real-time application on a target LSI11 processor (including the KXT11) is first prepared on a remote development system. The complete memory image is downloaded over a serial line for execution on the target. The memory image consists of a kernel and a set of user-written applications. MicroPower Pascal provides for the breakdown of the application into concurrently executing processes. Through a system of semaphores, priority tables and inter-process communication facilities, the processes gain temporary control of the CPU in an organized way. During execution, the program may be halted from the development system. Variables can be examined and modified. Watchpoints and breakpoints can also be set. The Robotic Aid Project uses a VAX11/750 computer as a development system. The VMS operating system allows several programmers and engineers to edit, compile, link, download and debug their applications simultaneously.

A complete MicroPower Pascal applications package now exists for the mobile component of the Stanford VA Robotic Aid. It consists of nine concurrently executing modules, linked together through a variety of synchronization flags and shared data structures. The modules are listed below:

COMMANDER. This module receives commands from the stationary IBM computer. It either responds to them directly or routes them to one of the other modules. Typically, a command is a query for data (the position of the robotic vehicle, for instance) or a change in one of the data structures (for example, the velocity of the manipulator arm). In addition, this module performs startup operations when the robot is first turned on. The COMMANDER is activated by the presence of a command in its input queue. Any other module may add commands to this queue.

CLOCK. This module is interrupt-activated by the 140 Hz. external clock that is applied to the QBUS BEVENT line. On each beat of the clock, it consults a (command-modifiable) table that contains a reference to a process that should be activated (for example, gathering data from the robotic hand).

RADIO. This module decodes and routes commands to/from the 4800 baud radio link via an input queue. Typically, these commands originate from (or are intended for) the COMMANDER module.

KEYBOARD. This module is activated by an input queue. It accepts commands from the on-board terminal and puts them in the COMMANDER input queue. It also displays information on the LCD display.

TROUBLE. This module responds to error conditions, as flagged by other modules. Typically, it queues a COMMANDER command (to shut down a particular subsystem, for instance) and relays an error message to the user base station.

MOVER. This module performs motion computations for the arm, hand and mobile base. It is activated at 29 millisecond intervals by the CLOCK process. It in turn activates a varying number of subprocesses, each corresponding to a special-purpose function of the robot. At this time, seven of these subprocesses are implemented: piloting and navigation functions of the arm and base, and simple control of the hand, the wrist and the bumpers. Results of these computations are reported in a standard format as velocities (in a labeled coordinate system) for the vehicle and arm. MOVER then performs the coordinate transformations and kinematic computations that result in a new set of joint/wheel angles. These are passed on to the driver processes. MOVER is highly modular, so that future motion subprocesses will be easily "plugged in" to the existing structure. By design, there is no built-in mechanism for interpreting the combined effect of the operation of the subprocesses although, in principle, the resulting motion may be chaotic. The role of coordinating the behavior of the robot is assigned to the operator who may, for example, disable any given function based on an evaluation of the situation and an understanding of the high-level goals that are being pursued. This results in a very flexible control system, although it places an extra burden on the designers to produce an efficient human/machine interface.

ADC. This module is a driver process that accepts interrupts from the sixteen-channel ADC module. The digitized values are packed into a standard data packet that is available for inspection by other modules.

JOINTS. This module processes interrupts from the nine joint servo controllers of the arm and the mobile base. It is activated at a 35 Hz. rate by the CLOCK process.

Two types of inter-process data structures are used: permanent data packets that have a fixed format and are dedicated

to data interchange between two modules (for example. the array of user-commanded velocities that is passed from COMMANDER to MOVER), and dynamically assigned packets that are created as the need arises (for example, the array of system voltages that may be requested by COMMANDER from ADC). All of the above modules are assigned command-input queues. COMMANDER, RADIO and TROUBLE are activated by the presence of an item in the queue. The clock-activated modules (MOVER, ADC and JOINTS) always check the queue and are able to process one queue item on each execution cycle.

ACKNOWLEDGEMENTS

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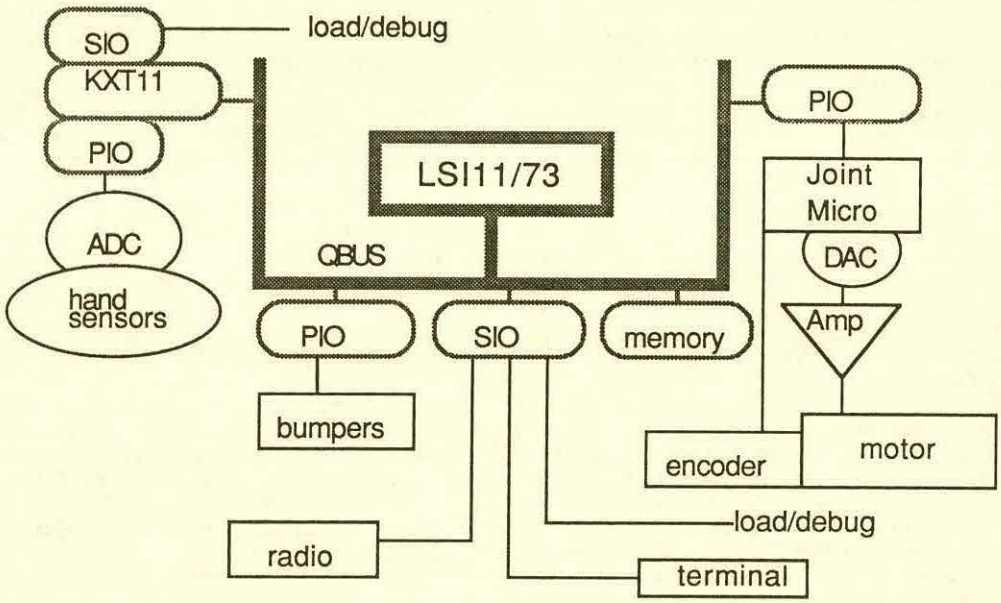

Figure 1. The components of the mobile robot's computer system are shown grouped around the 22-bit QBUS and the LSI11/73 processor.

CALVIN: A Robot Control Language For Rehabilitation Robotics

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ABSTRACT - A new control language has been developed for use with small, microcomputer-controlled robotic manipulators. The language was specifically designed for use in rehabilitation settings. Particular attention has been paid to the user interface, programming environment, portability of programs, and extensibility. The language has been introduced and well accepted for use at two clinical sites investigating occupational applications of small robotic manipulators. Application program development time has been drastically reduced and the language has permitted the robots to be used with clients for whom a suitable interface could not previously be found.

INTRODUCTION:

The application of remote robot manipulators has tremendous potential in the field of rehabilitation. A robotic arm may allow a severely disabled person to have increased control over the physical environment. The concept of allowing a person to interact with an electromechanical arm is not new. A number of existing research efforts have demonstrated that the use of modern robotic arms controlled by minicomputers can be programmed to perform important and necessary tasks for disabled persons [Corker et al., 1979; Leifer, 1981; Leblanc and Leifer, 1982]. While there has been the promise of large gains in independence, a number of issues have prevented the potential of robotic arms from being realized.

One of those issues is the lack of a robot control language suitable for rehabilitation robotics [Buckley, 1983]. This problem manifests itself in numerous ways in the various contexts of contemporary rehabilitation robotics. Examples are the difficulty of developing interactive application programs, obstacles to trying out new interface devices or control strategies, and having to start over when a new robot is purchased. The specifications of CALVIN address these particular needs and provide a flexible environment for rehabilitation robotics development.

ROBOT PROGRAMMING:

While a large number of robot control languages exist, they can all be grouped into three categories [Lozano-Perez, 1983]. Guiding languages wherin the user leads the robot through the motion to be performed, robot-level programming in which the users write a computer program specifying motion and sensing, and task-level programming in which the user specifies operations by their desired effect on objects. A brief examination of some characteristics of the existing languages will provide a useful context for consideration of the features that are desirable in a robot control language.

Robot Guiding - Robot guiding, also known as point-to-point guiding, remains the most common type of robot programming available on the market today [Snyder, 1985]. The robot is programmed by saving a series of points obtained by guiding the robot through task positions, usually with a teach pendant. Robot guiding is best suited to applications such as spot welding, painting, and simple material handlings. While the limited capabilities of guiding can in no way serve the complex requirements of rehabilitation robotics, the playback mode of programming may be useful.

Robot Level Programming - This type of robot programming language allows the user to program a task in terms of robot motions and sensor states without actually guiding the robot through the desired motions. For instance, a robot can be instructed to move to a defined position, close its gripper, and if it detects an object in its hand do one thing, otherwise do another. Languages of this type can be further divided into two catergories: unstructured languages and structured languages.

Unstructured languages provide simple conditional branching as their primary form of flow control. A well known example of an unstructured robot language is VAL™. The sensor interfacining capabilities and robot motion control sophistication of these languages vary widely [Bonner, 1982].

Structured programming languages such as KAREL™, RAIL, and AL incorporate structured flow control into the robot language. This will generally result in more readable, maintainable code. The structure and syntax of most of these languages are based on existing structured languages such as PASCAL or PL/1 [Lozano-Perez, 1983].

Task Level Programming - These languages allow programming in terms of desired effects on objects, effectively concealing low level decision making from the user. An example of an executable statement in such a language is "place block A on block B." Languages of this type being developed include IBM's AUTOPASS®, and LAMA at MIT [Lozano-Perez and Wilson, 1977]. Obstacles to successful implementations of task oriented languages are collision avoidance and emergency decision making [Lozano-Perez, 1983].

Most low-cost, microcomputer-controlled robots come with a limited robot-level programming language [Eshed, 1984; Sandhu, 1982]. These languages are geared toward the educational function of these manipulators; some are small subsets of industrial control languages. A more advanced language, ROBOTALK®, was recently developed as a $500 option for the Rhino Robot [Sandhu and Schildt, 1985]. Each of these

languages are only compatible with the robot they came with, effectively prohibiting program portability.

The SCORBOT ER-III™ used while writing CALVIN is supplied with an interpreted language called SCORBASE® (for SCORBOT BASIC) that is written for the IBM PC™. It is a menu-driven environment for program development limited to 100 previously defined positions for motion specification and 400 lines of program code. This language, representative of teaching languages, was a major obstacle to the development of useful rehabilitation applications using the SCORBOT.

IMPLEMENTATION AND DESCRIPTION:

CALVIN was designed and implemented at a time when progress on a major project was being impeded by its absence. This created an atmosphere of constant conflict between expediency, elegence, and robustness. However, numerous goals existed for the final product from the outset:

- That the language be usable by and useful for clinicians, application programmers, researchers, and the disabled users themselves.
- That support for additional interface periphereals and sensors be extensive and easy to implement.
- That the language be compatible with a number of candidate robots (Rhino, Scorbot, Microbot, UMI, etc.).
- That the language support and simplify the interactive development of interactive tasks.
- That the language be structured so as to create readable, maintainable code.

These will be addressed as they come up in the development process outlined below.

CALVIN is written on the IBM PC entirely in **C** and uses the YACC (**Y**et **A**nother **C**ompiler **C**ompiler) utility to construct the parser. YACC takes a description of the grammar of a language and produces **C** source code for the parser [Kernighan and Ritchie, 1978]. This has the effect of simplifying the process of extending the language. Adding some of the simple commands that have been requested by the sites has taken as little as 15 minutes. Creating a version of CALVIN for a particular robot for which primitives have been written is accomplished simply by linking the particular object file to the core of the language.

CALVIN incorporates features from Logo and **C**. Logo encourages an inquisitive, modular programming style and has been successfully used as an instrument for teaching computer programming [Papert, 1980]. The **C** programming language has powerful flow control and a convenient, compact syntax.

CALVIN was implemented as a programming environment. A user is provided with all the facilities needed to do program development, run programs, make listings, etc. The user has a full screen editor (a commercial editor being run as an executive process), an interface to the disk operating system, and a powerful interpreter for commands and application programs.

CALVIN is first a fully functional language with complete mathematical and triginometric operators, program flow control (if-else, for and while statements) and input-output routines. Statements are parsed into code that is then "run" on a simple software stack machine. While this software machine is much slower than compiled code, speed has not been a limiting factor due to the comparatively slow speed of the manipulator.

CALVIN commands may also be combined into procedures that can be executed at a later time. As a procedure is being parsed, the generated code is placed onto a program stack. Procedures on the program stack can be called in the same manner as the built-in routines or primatives.

The core of the language is augmented by a set of robot-specific driver routines for communication with the robot controller. This allows the language to move the robot joints, query the robot about move status, and any additional features supported by the particular controller (e.g. external switch inputs). The controllers of each robot for which CALVIN is being implemented communicate with the microcomputer via an RS-232 link. The major portion of the work of implementing CALVIN on a new robot involves the writing of these routines.

These primitive robot driver routines are combined in the language to provide the user with more convenient methods of describing the results that are desired from a program. Positions may be referenced in a cartesian coordinate system, named, stored and recalled. Motions can be expressed in absolute terms or relative to the current position. The size of an object in the gripper can be found. These routines contribute to a powerful system for describing desired robot motions.

A number of routines are present to support user interface hardware such as joysticks, a speech recognition and synthesis board, a digitizing tablet, etc. Robot tactile sensors, workspace instrumentation, and possibly a rudimentary vision system interface could also be implemented in this manner.

As a very simple example, the following procedure would make the gripper repeatedly move in squares each having sides 10 mm less than the previous one until the side length goes negative:

```
define square(side) [
        if (side > 0) [
                moveup(side)
                moveleft(side)
                movedown(side)
                moveright(side)
                square(side - 10)
        ]
]
```

The ability of a routine to call itself is called recursion, this feature can be very convenient in certain programming situations. This procedure is then available just like a primitive and would be run by typing a line like: **square(100)**. The incremental method of program development supported by this transparency of the primative/procedure distinction permits granular construction of applications covering a wide range of complexity [Papert, 1980; Harvey, 1985].

Programs may be written with what are being referred to as robot independent routines. These routines are a subset of the CALVIN primatives that refer only to world coordinates, and use only controller queries that are universally implemented. These programs will possess a maximum amount of portability. It is not possible to say how useful these routines and practices will be until additional experience is acquired.

An important aspect of CALVIN that is currently being investigated is what might collectively be called environment sensitive routines. These routines will monitor the position of the robot and objects in the world as the language knows it. These routines will provide much of the safeguarding that can done for interactive robotic users as well as protecting the robot itself from commands given by the user that may cause damage to the robot or workspace.

DISCUSSION:

The CALVIN language already meets many of the presented design goals. The clinical testing sites are a constant souce of suggestions and critiques. As expected, many of these changes and additions are simple to effect, others are considerably more difficult.

One major change involves the implementation of interrupt service routines to improve the real-time behavior of the language. It is imperetive, for safety reasons, that there be reliable ways to stop the robot in the event of an imminent or detected collision, or upon receiving various types of input from the user. These routines may be complicated by the issue of communicating with the controller via an RS-232 link.

Another change that may prove challenging is the incorporation of inverse differential kinematics (straight line routines) routines into the language. Not only do these routines involve a great deal of calculation, they also require velocity control of the motors, a feature seldom present on inexpensive robots. Nonetheless, approximations of these routines can be implemented using many through points and/or the limited velocity control provided [Paul, 1981; Brady, 1983].

CONCLUSION:

CALVIN is a fully functional robot control language that offers its users numerous benefits over the languages traditionally supplied with the purchase of a microcomputer-controlled manipulator. Clinical trials are demonstrating that the language is robust, easy-to-learn, and extremely useful for developing rehabilitation robotics applications.

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INTEGRATED SENSORY PERCEPTION FOR REHABILITATIVE ROBOTIC AID

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ABSTRACT

A practical robotic manipulation aid must posess an adequate sensory awareness of its surroundings in order to display the level of autonomy required in the rehabilitative environment. Sensor integration for robotic aid offers an effective solution to this problem. Current research focuses upon integration of force and proximity sensing in the context of the Veterans Administration/Stanford University Robotics Aid Project.

INTRODUCTION

The notion that manipulators may be used to aid severely handicapped people is based upon the premise that robots will be capable of managing the issues of safety and environmental variations with a sufficient level of autonomy. Because a robotic aid must operate in close proximity to human beings, in a constantly changing environment, its ability to adapt to its surroundings and to interact with its user(s) is fundamental to its effectiveness.

Any manipulation that is either interactive or adaptive requires an efficient and robust sensory system. Sensory perception is, in fact, prerequisite to intelligence in robots. The Veterans Administration/Stanford University Robotic Aid Project (VA/SU RAP) is in the process of implementing a multiple-sensory system.

Given a set of sensors, however, a straight summation of the sensory signals is not enough. An intelligent coordination mechanism must control the sensors and process their signals to maximize selective information acquisition and minimize the computational burden--hence the name, "integrated sensory perception" (ISP).

As an exploratory case for the general multiple-sensor system, VA/SU RAP is currently combining a 6 degree of freedom force information with optical proximity sensor information. The force/proximity ISP will provide valuable insights for a subsequent, more sophisticated ISP system. In addition, it will augment the existing robotic aid with new features such as:

1. Reflex Motion.
2. Object/obstacle groping capability.
3. Surface/Force contour following.
4. Sensory information reinforcement and verification.

ISP DESIGN CRITERIA

Background

Range of information. For a multi-sensor system, the model of the world that is derived from a set of sensory data can vary widely depending on which sensors are used and how. The controlling computer system could conceivably process a huge block of sensory data that is totally irrelevant to the task at hand. Thus, the selection of the sensors is important, particularly in regards to the type and resolution of the sensory data. Neither large gaps nor unnecessary overlaps in sensory windows are desirable. Once appropriate sensors are selected, the robot gains an added dimension in perception as the sensory window to its environment can now move from one sensory input to another.

Parallel processing. Parallel processing becomes imperetive when several sensors are linked to a robotic system. Each additional sensor adds a corresponding computational overhead. The central processor generally cannot afford to handle the extra computational burden of controlling each individual sensor. Consequently, a multi-sensory system is almost always implememted on a multi-processor system.

Selectivity. Due to limited computational capacity, a multiple-sensor system must process the sensory data selectively. How a robot determines upon which set of sensory data it should base its next joint actuation at a given moment and situation represents a crucial task for the real time control of a robot. An integrated sensory system must automatically extract and process only the pertinent information in the least amount of time.

Intelligent Controller

The above design requirements indicate that an intelligent control structure is an essential component of the integrated sensory system. Several different configurations of this structure are currently being explored in various research centers. Among these are the hierarchical[3], distributed[2], and layered[1] control structures. Although each has its strong and weak points, this choice is strongly dependent upon the hardware and software availability. Although other configurations are being considered for subsequent ISP systems, the force/proximity ISP system is based upon a hierarchical configuration. Figure 1 shows the three primary control hierarchy: the Planner, Coordinator, and Controller.

Planner. The fundamental reasoning and decision-making routines for the robot reside in the Planner. This component incorporates the given task command and the current sensory status to generate a set of control parameters for both the Coordinator and the sensor processors.

Coordinator. The Coordinator is basically a number crunching module that generates the actuator command torques from the given control parameters.

Controller. The Controller directly drives the actuator with the given command torque via a feedback loop.

Human interaction

Clearly, the most interesting part of the above control structure is the Planner. For RAP, this is also where the concepts diverge from an industrial robotic system: the difference is the human interaction. Because a robotic aid is almost always under direct command of the disabled user, the user can take over much of the difficult non-time-critical decision-making tasks which normally would be assigned to an AI routine module. The Planner's role simply reduces to one of interpretation and confirmation.

The extent to which the human will take over the Planner's role is not yet definite. It is likely that RAP will develop both autonomous and human-assisted planner modes. In any case, this human interaction is an essential part of the rehabilitative robotic application (and conceivably in industrial applications as well.)

Watchdog processes
One major drawback of a rigid hierarchical control structure is the sluggishness that results from the accumulated time delays as the command and status information are passed from one level of the hierarchy to the next. This does not pose a problem as long as the commands issued from the top level are not time-critical. However, situations do arise where the system must react quickly to detected environmental or internal conditions. For this reason, a set of watchdog processes reside outside of the central control loop. Effectively spanning the entire hierarchy, these provide some essential safety and adaptive features, which include system status checking, sensory data verification, learning, and reflex motion functions. Unlike other processes in the hierarchical structure, these operate asynchronously, and almost independently. Certain sequences of events and/or a special combination of sensory inputs can trigger these processes to run simultaneously with the master processor, modify the command parameters of the Coordinator, or even take over the Controller directly. These watchdog processes are not completely independent because the planner can override or bypass them.

IMMEDIATE GOALS

With the force/proximity ISP system, RAP seeks to add the following features to the Interactive Robotic Aid system:

Reflex Motion
One very valuable feature of this project is reflex motion. Parallel processing allows the system to monitor the robot motion continuously for extraneous or undesirable conditions. Occurrence of such conditions--e.g. detection of a surface where it is not expected, or sudden rise in force--will immediately trigger an appropriate reflex routine. These routines, by necessity, are simple, and yet, they could prevent costly damage to both the environment and the robot itself; they could also prevent injury to the user. Through the Planner the disabled user may select a particular reflex motion at the beginning of a task execution--e.g. go-limp, freeze, retract, etc. He also has the option of overriding the reflex trigger mechanism if so desired.

Groping
For both grasping objects and avoiding obstacles, the manipulator's ability to detect a surface using both the proximity sensing and tactile force sensing will greatly enhance the success and safety of most operations.

Balancing
The force sensor allows the manipulator to determine the best way to grasp an object by simple weighing and balancing procedures. Objects can be transported with greater ease and safety.

Surface and Force Contour Following
The combination of the two sensors allow surface-following capability by proximity, by force, or by both. Numerous applications are feasible, such as groping for an object along a surface, turning a knob of unknown axis-orientation, uncapping a bottle, etc.
Sensory Data Reinforcement
Sensors can be cross-calibrated by detecting and comparing against a common reference point. For example, a surface can be detected by either the force sensor or the proximity sensors. Given the surface as the common reference, the validity and the calibration of either the force or the proximity sensor may be checked. This feature allows the sensory data to be much more dependable and robust than if only one sensor were available.

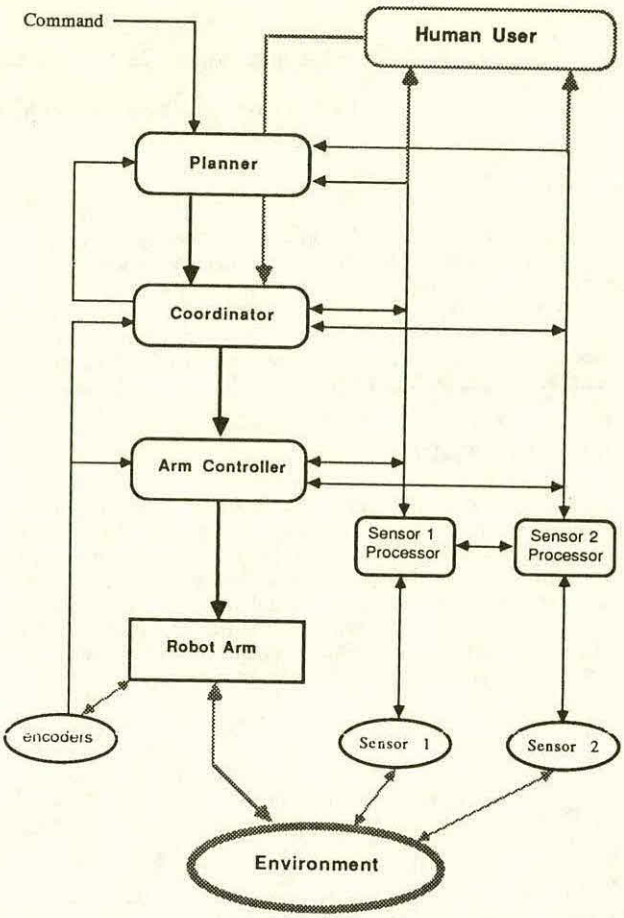

Figure 1.
The architectural layout of the Force/Proximity Sensor Integration consists of three major components: Planner, Coordinator, and Controller. Direct human interaction is incorporated into the system design.

RESEARCH STATUS

The current overall status of RAP is described in companion papers presented at this conference. The force/proximity ISP system now has working sensors and interfaces to the system processors. The force sensor extracts 6 components of the applied force on the end-effector with 8 sets of strain-guages mounted on the stressed members of the force wrist. The proximity sensor is a gripper equipped with 16 infra-red emitter-receiver pairs [4]. These IR sensors are divided into 12 groups, that measures the proximity in 5 directions of the end-effector plus the orientation of the hand with respect to a surface.

Each set of sensors interfaces with a KXT11-CA dedicated peripheral processor. These peripheral processors communicate directly to the arbiter processor (DEC LSI-11/73) and I/O port drivers via the common Q-Bus. Although the KXT has proved to be too limited for any high-speed computation, its ability to link directly to the LSI's Q-Bus has provided a simple and effective means of parallel processing. The interfacing and communication between the KXT and LSI processors are now operating with satisfactory results.

A symbolic processor (DEC MicroVAX II) and human/machine interface (IBM Graphics Monitor, DECTALK speech synthesis and Kurzweil speech recognition systems) are currently under development as separate components of RAP.

The software development and algorithm testing for the force/proximity ISP is under way. The stability of the system response to various sensor inputs merits special attention within the context of force/proximity ISP. The problem will be analyzed to insure proper system behavior.

Once the force/proximity ISP reaches a satisfactory state, RAP will proceed to build an extended ISP system by including the sensors of the mobile base.

CONCLUSION

Sensory perception is essential for the adaptive and interactive manipulation of a rehabilitative robot. A random combination of a large group of sensors will not achieve an effective sensory system, however. Instead, an intelligent coordination and processing of a set of carefully selected sensors will provide an integrated sensory perception for the rehabilitative robotic system. Such a system, when realized, will add a new level of autonomy and safety to the present state-of-art Robotic Aid, and thereby increase greatly its potential benefit to severely handicapped persons.

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12.5

An Independent Vocational Workstation for a Quadriplegic

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With today's technology, people who have impaired mobility can make use of their mental talents and join the mainstream of working force in our highly computerized world.

This paper briefly describes the Boeing voice-activated microprocessor-based workstation which allows a physically limited individual to manipulate data electronically and conduct business activities naturally. With the aids of the voice driven robotic arms mounted on a specially designed furniture, the quadriplegic can perform the physical tasks required in an office and become independent of frequent external assistance.

The paper also discusses the difficulties encountered in the available technology, the safety issues, the human factors and how we have dealt with them to ensure independence for an individual in an office environment. It also briefly touches on the future enhancements planned for this workstation in intelligent robotics and machine understanding.

Background and Motivation

Of the physically impaired individuals, the majority are quadriplegic, caused by disease or injury. It is believed that they constitute a pool of intelligence and highly self-motivated resources.

Boeing has several reasons for sponsoring this project. Most importantly, it is felt that the experience gained in our developing speech understanding applications can be applied to industrial situations and the company's aerospace work. Of course, Boeing is, as ever, interested in recruiting qualified, dedicated computer professionals locally in a what has become a very competitive market. Boeing sees this as an opportunity to tap this pool of largely conscientious and talented workers, while performing a community service.

Phase 1 of the project, which began in early 1984, initially focused on providing a speech controlled workstation that could be used by programmers and analysts and on voice control over data later in October. We believe that if a workstation can be built for a programmer, others might be able to adapt it to satisfy the specific needs for other professions such as financial analysis, engineering and manufacturing.

Phase II of this project, started to consider that a quadriplegic individual when using the workstation might impose a burden to co-workers in the environment in which the unit was to be placed. The physically limited person could not handle such routine functions as referring manuals, retrieving printed output, loading diskette into a disk drive without assistance, etc. The voice controlled robotic aids were then added to the workstation.

Workstation and Speech Components

The workstation that is in use now has a microcomputer system that supports and is driven by two voice recognition products and a voice communication product.

The microcomputer is an IBM PC XT. This unit is capable of emulating different terminals which give access to multiple vendors' mainframe products. Besides, there is also a greater freedom of selection amongst voice products for the IBM PC.

The voice communications product comes from Dialogic, and the voice recognition products, from Keytronic and Microphonics. The following explains why the two voice products are being used.

The Keytronic 5052V keyboard unit was initially programmed so that voice could be used to simulate a full keyboard. When text editing and spread-sheet software were added, the functions of the Keytronic product were expanded to substitute spoken commands for keyed command sequences. While the keyboard entry method could be used to perform the required tasks, to achieve the entry rate of an experienced keyboard operator, single commands were needed which produced the equivalent of multiple keystrokes.

Keytronic, at Boeing's request, modified the product so that up to 160 command words or phrases were supported at one time. But 160 words were not enough when the user was called upon to employ graphics, support multiple programming languages, and to obtain access to different networks. The vocabulary reached nearly one thousand words. While the 5052V supported multiple syntax structures, the available vocabulary limitation did not meet the requirements of this environment.

Microphonics approached Boeing with the request to assist them in testing their OTO-1 product (now called "Pronounce"). This product permitted the saving and downloading of different files each containing up to 128 words or phrases. It was decided to marry the two products together by using the Keytronic unit to emulate the keyboard and initiate the downloading of different application vocabularies to the Microphonics board. The products were also set up to logically turn each other off and on.

Since the user is dependent on phone communication, a voice communication product is also required. The Keytronic product provides the commands for dialing; and the Dialogic DIALOG/2 product, the phone interface -- dialing, answering, recording conversations, note-taking and messaging. These functioning additions necessitate, of course, additional equipment (e.g. an expansion chassis and 25 Mbytes of hard-disk storage).

Robotics Requirement

The functions of robotic aids include such diverse tasks as handling manuals, floppy disks and individual sheets of paper. Use of a robot in an office environment requires cleanliness and minimal noise. Robotic control by means of the IBM PC is desired to provide local control. Other requirements include relatively smaller size, at least 4 ft of reach, a 10 pound payload, programmability by the operator, force sensing (for safety) and (again) low cost.

As a result of the search for a low cost robot which met these requirements, Universal Machine Intelligence, Ltd. (UMI) of London, England, brought their robot to our attention. Two RTX arms were ordered by Boeing and delivered in March, 1985.

Tasks laid out for the robotic aids to perform are as follows:

Display reading material on the reader-board.
Manuals and books (weighing up to 10 pounds) are placed in the specially designed book shelves. The robotic aids will retrieve them from the shelf, display them on the operator's reader-board, turn pages and return the materials to the requested location.

Handle filing in a file cabinet.
The robot arm will open/close the file cabinet, insert, retrieve documents from file folders, and assist the operator in searching for document folders.

Handle floppy diskettes.
Most of the commercially available business packages used by the microprocessor require their key diskettes to be loaded in the floppy disk drive while processing. The robotic aid is to pick up a requested diskette, load it into the floppy disk drive and return it to the requested diskette storage slot -- when instructed.

Adjust printer according to the operator's commands.
The robotic aids are to load printer paper, retrieve printed pages and display them on the reader-board. The printing function, however, creates several difficulties. There is a need to produce diverse types of output -- graphic, standard and condensed print, draft and letter quality. Paper handling can be especially difficult for the robot. Loading paper, removing single or fanfold sheets and alignment can not be accomplished without a great deal of engineering, programming or vision. To overcome this difficulty a Quadram laser printer is used. This printer collates the output so that multi-page reports can be stacked in the proper order.

Perform miscellaneous support activities.
The robotic aids can place materials in the waste basket, on the user's lap, retrieve or place materials in the user's back pack or an in/out tray.

Furniture Consideration

In this application the location of robotic aids becomes more than just a functional consideration. Safety of the operators becomes critical. Maneuverability of the operator is limited. The operator's attention is frequently on the computer monitor or the reader-board. The robot is expected to carry out assigned tasks without attendance and without posing a hazard to the operator.

The robot's placement determines the design of the workstation. The robot must be able to access the disk drive(s), printer, reader-board, file cabinet, book shelves, in/out tray, waste basket, diskette storage, user's lap and back pack. The implementation of the robot demonstrated that a single arm could not be positioned to perform all these functions unless mounted on a track. The track may be added after other requirements
(which will be discussed later in the paper) are satisfied, but two arms are now needed to perform all the tasks.

File cabinets present another problem. Manufactured vertical and horizontal cabinets have been rejected. Horizontal cabinets demands more space than is available. Vertical ones do not furnish the kind of access required by the robot. A vertical file that has a 10" X 11" footprint and is divided into 4" shelves has been designed so as to make it an easy access to the robot.

Manuals and books are stored in specially designed vertically partitioned book shelves. Each partition contains one book or manual.

Peripherals Support Consideration

Diskette loading and unloading require a modification to the drive face. Floppy diskettes sag when supported from the edge. Care must therefore be taken not to crash them against the drive face or crush them in the gripper. A tray is used to align the diskette both vertically and laterally to facilitate the loading and guide removal.

The operating system is a multi-tasking environment to permit independent activities (e.g. robotics, telephone management and applications). Microsoft WINDOWS is employed to manage the multi-tasking function.

Robot Programming

The robotic tasks are divided into four categories -- task, ad hoc, teach and system.

The task operations involve directing the robot to perform a sequence of retrieval, storage and manipulative instructions. The computer or the robot controller must remember prior placement location of several elements of the system. The workstation is also called upon to remember what to do with materials after their retrieval.

These knowledge operations are part of either the "source" instruction set or the "destination" set. The source memory activities include knowing the starting position of the arm, where the material is stored, etc. The destination activities involve

remembering if the space where the material is to be placed is already occupied (like the disk drive) so that accidents will not occur. Other destination movements embrace transport of the material from a fixed position (manual partition) and return the arm to "home".

Several manipulative tasks concerning a fixed sequence of operations are performed -- i.e. reset the printer, turn a page, etc. The emergency stop sequence is always available to permit the operator to halt operations and restart with several options.

The ad hoc category of operations deals with the movements of the arm in world coordinates -- left, right, up, down, in, out, pitch, yaw, roll,open and close. These are used by the operator for non-routine activities.

"Teach" operations are simple commands to the system to store ad hoc movements in a reference set so that the robot can be instructed to perform new routines (tasks) that are reproducible. "Remember" is the only word added to the ad hoc command syntax.

All the above are under voice control.

There are also some computer system controlled tasks. Included in this category are such actions as return to home before shut-down or initialize the robot when the system is started up. System emergency stops take place if an obstacle is encountered.

Looking into the Future

The above takes this project through Phase 2. All the tasks were completed in April 1986. There is an intent to integrate the speech controlled robotics with further developments in Real-Time Systems Research at Boeing's Artificial Intelligence Center. The following possibilities are being considered.

One of the Real-Time programs is sensor development. Gripper feedback, for example, has a variety of capabilities. Amongst them are: shape determination, thermal sensitivity, reflective surface awareness, resistance detection and awareness of the center of gravity. Force sensing is already included in the arm, but not yet in the gripper.

Interchangeable grippers have been proposed by UMI which has not yet, however, forecast the modification of the arm to support this capability.

Voice response is now under consideration to provide feedback from the robot and emergency situations when the robot is performing unattended activities. Upgrade of voice functions (speech recognition, response and communication) are also among the items to be considered, as the results of the research and evaluation in Speech Understanding prove to be applicable.

Summary

By April, 1986 the workstation and its operator are fully functional in a business programming environment. The operator is completely independent of supportive aid from co-workers. We, at Boeing hope that our work will provide an example and encourage others to continue on many fronts to help the handicapped to overcome their physical limitations.

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"COMPUTER COMFORT": THE NEIL SQUIRE FOUNDATION
Computer Training for Severely Physically Disabled Adults

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ABSTRACT: The Neil Squire Foundation was created in 1984 and is dedicated to research and development of technical aids for severely disabled persons. "Computer Comfort" is our nationwide computer familiarization program whose structure and organization is the focus of this paper.

INTRODUCTION: THE PROBLEM

"Basic rehabilitation is a drag. It teaches you what you learn in the first four years of life - mobility, personal hygiene, avoidance of hazards, muscle development, and other pretty unoriginal stuff. Remember that the genuine aim of rehabilitation is to achieve your goals. And what's your goal? To escape, of course."[1]

Severe physical disability can make the goal of escape an impossibility. However, technological advances in recent years have made it feasible for severely disabled individuals to control and interact with their environment more autonomously. Many of the immediate technological hurdles have been overcome. The problem that remains is the transfer of this technology from the engineering laboratory to the disabled user.

Currently, disabled people are waiting years for equipment which could increase their independence today. Society questions the $3,000 price tag on a bedside computer-accessed environmental control unit that may eliminate the need to provide 24 hour care in an intensive care unit (ICU). The increase in personal dignity and independence and the resulting reduction in unnecessary placement in ICU for the severely disabled individual is harder to measure but glaringly apparent to all who live or work in that environment. Even in those institutions fortunate enough to have acquired a few pieces of this new technology (computers, adaptive electronic devices, etc.), one often sees a tragic breakdown in the technology transfer process. Switches lie unused in drawers, computers gather dust because clinical personnel lack the necessary training and time to integrate this technology into their patients' lives.

Computer Comfort

The Neil Squire Foundation's mandate is not only research and development of technical aids for severely disabled persons, but also the facilitation of technology transfer - making these aids available, accessible and affordable.

"Computer Comfort" is one of the Foundation's development programs. Funded by the National Research Council of Canada - Industrial Branch, it was started as a pilot project with some of the severely physically disabled adults on the extended care unit at Gorge Road Hospital in Victoria, British Columbia. The Neil Squire Foundation describes "severely physically disabled individuals" as those who utilize an electric wheelchair, who may be non-vocal or may require a ventilator to breathe, and who are otherwise physically dependant on others for their activities of daily living.

The objective of the program is to help remove the fear that many (disabled) people have in approaching a computer by allowing those who indicate an interest in the computer to learn at their own pace. The teaching is done by computer science and engineering university students on four month work terms. The "Computer Comfort" program is now Canada-wide and there are instructors in more than 50 extended care units, acute hospitals, rehabilitation centers, group homes and private homes. A review of instructor reports from the first year of operation indicates that more than 600 disabled Canadians have been helped by "Computer Comfort". These instructors provide one-to-one teaching sessions to individuals who would otherwise not have the opportunity to learn computer skills. The skills acquired range from using the computer as a communication aid (in the case of a non-vocal person), to designing a home using a voice controlled computer-aided design package (in the case of a high lesion quadriplegic individual). Many are writing letters independently and confidentially for the first time in their lives, some are designing greeting cards using morse code input (via sip 'n puff switches), some are embarking on small buisness pursuits, while still others are communicating on a larger scale via computer networks.

Training and Supervision

The instructors are supervised by professionals in each center in addition to the direction they receive from the Foundation. They are considered computer resource persons for the treatment team. Across the country these supervisors represent the full spectrum of therapeutic disciplines. In each instance, the clinical supervisor is a staff member who has a particular interest in using computers as everyday tools with their patients.

Our "Computer Comfort" instructors attend a two-day Neil Squire Foundation workshop which introduces them to the special needs of the disabled person and to the range of adaptive equipment available. The instructors then receive orientation to their individual program sites by their respective clinical supervisors. It is not uncommon for one "Computer Comfort" instructor to be shared by two centers. They also carry out independent projects. These projects may take the form of specialized software (eg;"BIG PRINT" word processing package for the visually impaired) or small scale hardware adaptations.

Because each instructor is part of a nationwide program, an immediate network is in place to share

projects and information. Therapists are discovering an easier link to resources outside their institution, city or province.

Future Directions

Although "Computer Comfort" is helping some disabled people overcome some of the obstacles that stand in the way of easy access to computer technology, many hurdles remain to be surmounted. Computers are not readily available as technical aids. Often they are seen as a luxury - not as a necessary extension of the rehabilitative process. The procedure for acquiring this equipment is usually long and frustrating.

The Neil Squire Foundation is actively seeking the cooperation of various government programs, corporations, insurance agencies and public service groups to address these problems. Through participaion in conferences and in international events such as Expo '86, the Foundation is continually seeking to increase public and professional awareness so as to facilitate the transfer of high technology to the severely physically disabled adult.

Summary

'Escape' into a new world of options is now technologically possible for physically disabled individuals. As rehabilitation professionals, we must also accept the responsibility and challenge to ensure that this possibility becomes a reality.

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COMPUTER-BASED TOYS FOR DISABLED CHILDREN: A FEASIBILITY STUDY

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ABSTRACT

Computer programs and special hardward were developed and tested with 10 disabled and 7 non-disabled children to learn how they would respond. After some modification to the programs, 6 children with varying degrees of disabilities were tested with both the computer programs and conventional adapted toys. The results show that the computer programs were at least as effective in maintaining attention as the adapted toys. Cost estimates show that the selling price would be realistic for rehabilitation facilities.

INTRODUCTION

This project was begun on the basis of a perceived need for an educational toy model that is appropriate and functional for use by numerous children at various stages of cognitive development and with limited and varied physical capabilities. It was hypothesized that a computer could be employed as a toy (having graphic display, sound output, etc.) and as a control center for activating a variety of toy variations adapted for use by children of a wide age range and with limitations caused by a variety of disabilities. The goal of this project was to study the feasibility of utilizing computers as an educational aid (a "toy") for disabled children.

PROJECT DESCRIPTION

The project began with the formation of a working committee of professionals including representatives from Rehab. Engineering, Psychology, Physical, Occupational, Speech and Educational Therapy departments. The committee met regularly to monitor progress and facilitate development and assessment of computer based learning aids (toys).

The general approach was to:

1. Develop the "toy" itself, including any special hardware.
2. Test the concept on several children to learn as much about the applicability of the technique as possible.
3. Modify the "toys" based on this experience.
4. Conduct a test to compare disabled children's response to computer "toys" and conventional adapted toys.
5. Study the economic feasibility of developing a complete computer-based educational system for disabled children and marketing it to schools and rehab. facilities.

Each of these program elements is addressed below.

Development Of The "Toy" Elements

Hardware. A total of 10 different switch units were fabricated for this project, plus 4 duplicates. These were:

1. large plate switch (SPST, NO, momentary)
2. large plate switch with texture (SPST, NO, momentary)
3. rocker switch (SPDT)
4. finger pinch switch (SPST, NO, momentary)
5. squeeze bulb switch (SPST, NO, momentary)
6. plate switch (SPDT)
7. reed switch, magnet activated (SPST, NO)
8. mercury level switch (SPST)
9. sip and puff switch (DPDT, NO, center off)
10. photocell switch (cover to activate) (SPST)

In addition to the above, 2 joysticks were purchased (one of these was modified to require less force to activate) and one light pen. An interconnector box was also made to connect up to 10 switches up to the 2 joystick ports of a Commordore 64 computer. This provided the means for a child to access the computer using only switches. The Commodore 64 computer was used in this project along with the associated 1702 Color Monitor, 1541 Disk Drive, and 1526 Printer.

Software. A total of 14 computer programs (games) have been developed with variations on speed, graphic display, and auditory components. These are "cause and effect" learning games that are activated by closing switches, and combinations of these into "activity box" games. Some commercial software was purchased to better determine what was available, and one of these, "Match-Up" by Hayden Software was modified to work with our external switch interconnection device.

Preliminary Testing And Conclusions

Ten children with varying disabilities plus seven non-handicapped children were exposed to and observed using the developed software and hardware with the following conclusions:

1. Disabled children can and do attend to the video screen, however, it appears that the child must be functioning at or near the 18 mo. level in cognitive skill to maintain their interest.
2. All children tested showed some attentiveness to sound effects (youngest was 10 mo. old).
3. Each child tested had toy preferences, ie. fast or slow-moving displays, colors, and sound effects.
4. Computer-based toys offer options which cannot be obtained with conventional toys. These options include selection of colors, sound effects (including synthesized voice output), automated record keeping for each trial and for each child, and adjustable speed of displays.

The variable speed at which toys can be played appears to be extremely important. Some children respond very positively to quickly moving, fast-paced toys while others may be overwhelmed at that pace. For the latter group, toy speed can be reduced to their comfort and learning level. In both cases, toy speed can be increased or decreased to meet specific learning objectives.

The automated record-keeping also appears to be very important to rehabilitation professionals. They have a need to document progress and the computer offers a relatively easy way to record data, store information, generate reports, and other related activities.

5. A large number of programs would be required to effectively serve disabled children in the age range desired because of the different effects of various disabilities and individual preferences.
6. Preliminary investigation of commercially available microcomputer "matching games" and "puzzles" appear to be too advanced for the population addressed by this project. The children at this age are just beginning to develop concepts of size, shape, color, and number.
7. During the course of this project, it was found that several commercial vendors are now directing their efforts to developing programs for preschool children, often recommended for ages as low as three years. These programs are usually well developed with good sound and graphics, however, they are not intended for the under three population and/or for those with physical disabilities. The potential exists for the development of new programs for the younger population as well as modifying commercially available programs for use by disabled children.

Final Testing

Subject Selection. Six children were selected from a group of 16 disabled children according to the following criteria:

1. Moderate to severe fine motor involvement.
2. Cognitive/Language development at a minimum level of 18 months.

Skill levels were assessed according to the Early Intervention Developmental Profile (EIDP) and clinical observation prior to the selection of the test sample. Actual ages ranged from 26 to 40 months.

Training Session. One training session was performed for each child with three purposes:

1. Familiarize children with the equipment.
2. Position children appropriately at the equipment.
3. Select appropriate switches for the equipment.

During the training session, one adaptive toy and one computer program were selected as an introduction to the project, these differing from the toy and program used in final testing. This training session preceeded the testing by 3 to 7 days.

All training and testing sessions were performed in a consistent environment, the children being brought to a specific room reserved for this purpose.

Testing Sessions. Testing consisted of two 20 minute (maximum) sessions, the sessions separated by 3 to 7 days. During the testing sessions, the tester presented the adapted toy and the computer program. In the first session half of the testing sample played with the adapted toy first and then the computer program, while the order was reversed for the other half. In the second session the order of presentation was reversed. Only one adapted toy (the Musical TV) was used and only one toy program (the Clown Face). There was no minimum time of presentation, but a maximum of 10 minutes was set for both the adapted toy and the program. A presentation was ended prior to the 10 minutes when the presenter judged that attention was decreasing significantly. One other staff member was present to observe the child's reactions and complete a anecdotal record.

Test Results. For all of the tests overall time was recorded as well as the time that the toy was "on". This was accomplished by using the clock built into the Commodore 64 computer (for total time of presentation) and timed delay loops in the program (for time"on"). During Test 1 there was a problem with the adapted toy timing technique during the test of two children; for this reason Test 1 will be ignored. The results were as follows:

TEST 2	AVERAGE	RANGE
Battery Toy		
Session time (sec)	236	140-357
% Time "on"	46	14-57
Computer Toy		
Session time (sec)	454	313-620
% Time "on"	70	51-92

A paired sample T-test was performed on this data. The differences in % Time "on" were not significant (ie., the data have a .654 probability of being from the same population, 95% confidence level). The differences in Session Time are more significant, having a .006 probability of being from the same population, 95% confidence level.

ECONOMIC ANALYSIS

The purpose of an economic analysis is to develop an understanding of what the expected costs and sales would be if this "toy" concept was taken to the marketplace. This is difficult to do with precision since the "product" is not completely defined. A more complete package

(ie., many subprograms, special options, extensive record-keeping, etc.) should sell more units, but the cost of developing it would also be higher and a higher cost would tend to reduce the sales. The intent here is only to make a first approximation of expected cost and sales.

Cost of the Desired System. The primary cost of producing the type of system under consideration is that of developing the "toy" computer programs. The assumption is that 50 separate programs would probably be required to offer the variety of features expected with this type of product. If it takes 20 hours for a computer programmer to develop each of these 50 programs, that's 1000 hours total. A typical rate here for programmers is $40/hr.; this would be a development cost of $40,000. For this first approximation, all marketing costs are assumed to also be $40,000 and all other costs (copying to disks, packaging, testing, investment costs, etc.) also at $40,000. Thus the total cost of delivering the product to the consumer for the initial market thrust (say 1 to 2 years) is estimated to be 3 x $40,000 or $120,000.

Sales. This type of product would be of interest primarily to rehab. facilities and schools that provide services to disabled children. A look at the Milwaukee, WI area reveals 8 facilities that would likely be seriously interested in this type of product. In a 1 to 2 year market thrust perhaps 40% of these would be persuaded to buy; that would be 3. The Milwaukee area has a population of about 1.5 million. Assuming this facility to population ratio holds for the rest of the country with a population of roughly 225 million, then country-wide sales would be 450.

Cost Per Unit. The total cost of $120,000 divided by expected sales of 450 yields an expected unit price of $270, or round up to an even $300. The total system with the computer, monitor, disk drive, printer, and an allowance for switches and interconnectors should be under $1500. This seems to be a realistic price compared to other rehabilitation related programs, and consistent with the features that would be offered.

CONCLUSION

1. Children respond at least equally well to the two toy types. The data indicates that the computer toys might hold their interest a little longer.
2. The cost of a computer-based educational learning device or "toy" system is realistic for a rehabilitation facility.
3. Many variations of the "toy" programs would be required to serve the disabled population considering the wide range of age and disability that it would be desired to serve.

ACKNOWLEDGEMENTS

A 1984 Grant from the Service Club of Milwaukee to the Curative Rehabilitation Center provided the basis for this investigation into the use of computers for disabled preschool children. Additional financial support was provided by the Curative Foundation, Inc. This project also benefited in part from a computer equipment donation by Commodore Business Machines, Inc.

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COMPUTER-AIDED DESIGN OF REHABILITATION DEVICES

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ABSTRACT

A computer-aided design procedure is described and assessed, documenting the use of CADAM, an interactive computer-aided design (CAD) system, in the development of four projects: a rehabilitation research device, a mobility device for the handicapped, an apparatus for acute spinal injury support and a mobility system for a robotic aid device.

INTRODUCTION

The Rehabilitation Research and Development Center (RR&D) of the Palo Alto Veterans Administration Medical Center has recently undertaken design projects which differed from earlier efforts in one significant aspect: the availability of state-of-the-art computer-graphics techniques for aiding the designer in his search for an optimal solution to the problem at hand. Design personnel at RR&D are fortunate in their ability to access the CAD resources of Stanford University's Center for Design Research, a constituent part of the Stanford Institute for Manufacturing and Automation. The core of these resources consists of an IBM 4341 computer and ten IBM 5080 graphic workstations, running a variety of design/engineering software packages.

All design work described here was performed on CADAM (Computer-Graphics Augmented Design and Manufacturing), developed by Cadam, Inc. and licensed by IBM. The computer-aided design portion of CADAM consists of a high-function design and drafting package, as well as a number of aids to design analysis, while the computer-aided manufacturing segment provides for computerized numerical control (CNC) part programming. System configuration during the design development phase of the projects described here permitted descriptive geometry-based construction of two-dimensional orthographic and isometric projective representations of three-dimensional objects. These are essentially the same capabilities offered by traditional drafting, although with greatly enhanced capacity for manipulating the geometries in Cartesian and polar coordinate systems by selective scaling, translation, rotation, merging, duplication and other editing operations. Hardcopy output was produced on pen plotters, resulting in finished drawings complete with standard or user-defined symbols, text and notes at any size, angle, slant or spacing, and with full dimensioning to ANSI or ISO standards.

PROJECTS

A brief description of each of the four projects is outlined below.

Sunburst II

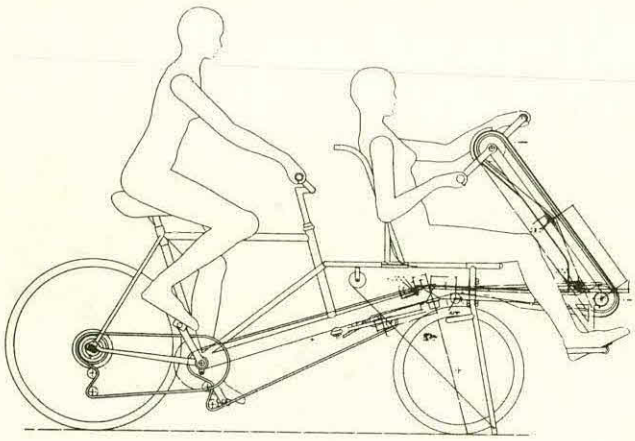

Figure 1. The Sunburst II tandem bicycle.

This tandem bicycle combines arm and foot-powered recumbent cycling in the forward position, with a standard operating position in the rear, permitting both able-bodied and disabled users to enjoy the freedom and challenge of cycling [1,2]. The Sunburst II uses many standard bicycle components, in conjunction with custom frame-building technology, allowing flexibility in configuration and relatively short turnaround in prototype iteration (fig. 1).

Turning Frame

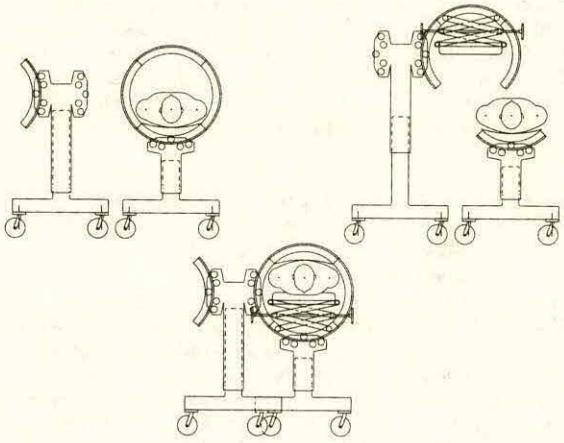

Figure 2. The turning frame, illustrating rotation of the patient from the supine to prone positions.

Part of a continuing effort to develop devices to reduce the risk of exacerbating acute cervical spinal cord injury during patient transportation and treatment, this project builds upon the experience gained from clinical trial of a carbon-fiber composite backboard and constant-force traction device[3]. This integrated system (fig.2) combines the characteristics of a turning frame, in which the patient is periodically rotated from a prone to a supine position; and a kinetic bed, which rocks continuously from side to side[4].

Arm-Powered Ergometer

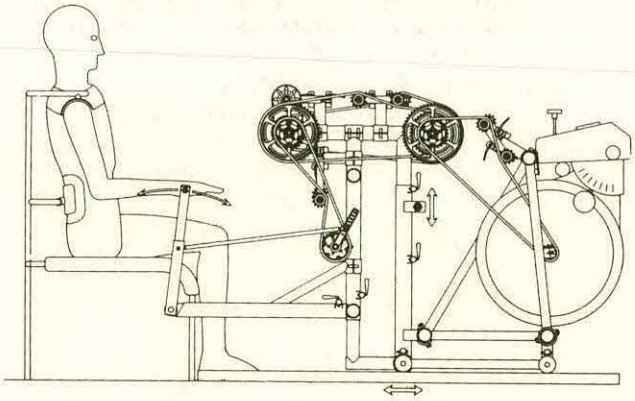

Figure 3. The arm-powered ergometer, configured for oscillating lever drive and Monark 868.

The arm-powered ergometer is a laboratory apparatus for the collection of electromyographic and kinematic data on subjects performing a variety of arm-powered tasks involving rotary, reciprocating and eccentric upper-limb trajectories (fig. 3). This physiological and biomedical data will provide the basis for a task-specific mathematical model for the upper limb, to be used in the design and development of optimal arm-powered drive systems for application in prototype exercise and recreational mobility devices for the lower-limb disabled. Components and construction techniques for the apparatus follow standard bicycle technology whenever possible[5].

Mobile Robotic Aid

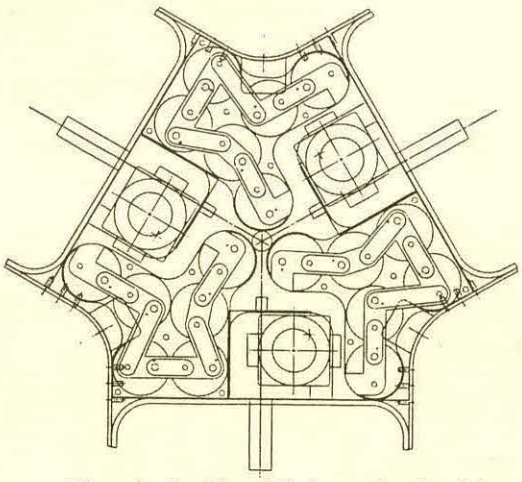

Figure 4. The robotic aid mobile base, showing drive train components and battery packs.

The Robotic Aid Project, an ongoing sponsored endeavor to develop a manipulation aid for the severely disabled[6], has as its current goal the design of a novel three-wheeled omnidirectional mobile robot, outfitted with a small industrial robot arm and an in-house developed gripper (fig. 4). The prototype nature of this system combines the reliability of such off-the-shelf parts as motors, drivetrain components, batteries, and computer card cages, with the innovation of the specially-designed chassis and various subsystems[7]. The work of this phase is expected to lead to evolution of the design through subsequent evaluative steps of developing a market-ready product. The use of the CADAM environment is expected to greatly facilitate the process.

DESIGN PROCEDURE

The procedural variations enountered in the various design phases of these four projects illustrate CADAM's ability to support the designer in differing developmental roles.

Sunburst II

Sunburst II is conceptually an extensive redesign of a previous version which was also developed in the CADAM environment. In a sense, then, what might be termed preliminary or conceptual design was accomplished at an earlier date, the previous project having provided a considerable store of information in the form of details, specific components and generalized solutions, all easily accessible for incorporation in the new version, whether in modified form or as originally conceived.

Illustrating CADAM's capabilities in this regard, the larger and stronger front wheel on the Sunburst II is a design change which causes a ripple effect throughout the vehicle. The front seat has been raised to accomodate the larger component, and the main frame tube was rerouted, permitting easier access when transferring on and off the bicycle. These changes were easily visualized and executed, beginning with drawings of the previous design and selectively redrawing specific components or features as required, while maintaining all relevant original geometry. Subsequent versions could then be compared and analyzed by simply superimposing their graphic images on the CRT screen.

With basic design completed, positioning of the front rider relative to various components was simulated with a digitized outline drawing of a 50th percentile female figure in side view. Adaptation for a specific user was accomplished by measurement of significant features of the individual and by rescaling portions of the outline drawing to correspond to those measurements.

Finally, working drawings were generated for the use of the frame builder, providing all necessary dimensions and specifications for construction of the bicycle. Other presentation hardcopy was also generated, including posters, slides and overhead projection transparencies, all based on the same geometry sets produced during the design process.

Acute Spinal Injury Support

With the turning frame, a somewhat different situation was encountered. Although the system was not based on an earlier version, its function centers about a previously designed complementary device, the composite back board mentioned above. Design of this element was by traditional methods, and the constraining parameters it provided for turning-frame design were input to the CADAM database in parallel with other project-specific conceptual constraints.

While a CAD system cannot substitute for the mental imagery needed to conceive of and convert an idea to a functional device, its ability to locate geometric envelopes in a simulated working environment is of great value in determining the viability of a design hypothesis. Thus the major components of the modular turning frame (cylindrical patient enclosure, enclosure support frame and enclosure handling gantry) were first drawn as simple boxes, scaled to reflect the dimensional constraints and ergonomic aspects of in-hospital patient movement and maintenance. Load-bearing structures were then inserted in their respective boxes, preparatory to installing the various required mechanical components.

The ability to generate and place a mechanism (and incrementally move, turn, invert or remove it) is valuable in evaluating alternatives and determining whether subsystems will be dimensionally compatible. The present mechanism for rotating the turning frame about the patient's body axis was selected for compactness and minimum parts count. As with the Sunburst II, a dimensionally-correct human figure was placed in the model to identify potential man-machine incompatibility. In one situation the backboard presents such an incompatibility, due to the conflicting requirements of the available space inside a helicopter ambulance and the ideal length of backboard needed for a 95th percentile male patient. The conflict is resolved by elevating the patient's knees as required.

Sequential drawings illustrating the operation of a system and the relative movement of detachable components are facilitated by CADAM's easy replication of subsystem drawings, translated or rotated with respect to various coordinate axes. A series of functional drawings of the turning frame was generated showing the separated gantry, holding a frame section above the patient, and connected to the patient-supporting cylindrical cage, which has been revolved to place the patient in a face-down position.

Hardcopy output illustrating the various operating modes and interfaces of the turning frame system and its constituent components has been used for proposals and presentations and can be readily adapted for instruction manuals for clinical use.

Arm-Powered Ergometer

By comparison with the preceding projects, design of the arm-powered ergometer was essentially unconstrained, with parameters set only by ergonomic limitations and by preliminary concepts of potentially useful (in terms of data collection) drive systems linked to a readily available load device. On the other hand, there was no existing store of drawings, computer-based or otherwise, that could be incorporated in the design effort.

Ergometer design began by establishing basic two-dimensional geometric parameters, based on such constraints as anthropometric data for a seated operator/subject and on the dimensions of the selected load device, a modified Monark 868 flywheel ergometer. Ranges of adjustment were also established at this time, and a tentative design for an adjustable hand

crank, together with constraints set by foreseen gearing arrangements, provided a basis for a schematic frame layout and general component placement. Preliminary analysis was performed for the tentative structural layout, providing rough approximations of frame component specifications.

The second design phase was concerned with the drawing of individual components ("built or bought"), and their inclusion in the preliminary general arrangement sketch. Since it was intended to reduce to a minimum the number of purpose-built parts, a significant portion of design time was dedicated to preparing detail drawings of commercial components such as bearings, chain rings, instrumentation elements and others. These drawings were then replicated as needed for inclusion in other drawings throughout the project, and will be available for use in later iterations, as well as in future projects. The design was refined by working alternately in detail models of specific components and in an inclusive model composed of the various detail or component models. Parts were modified, relocated and checked for fit and interferences, using standard and section views in three orthogonal planes, until a satisfactory solution was attained for the device as a whole.

The two-dimensional geometries thus established were then used to generate standard working drawings, with the addition of dimensions, notes and other information necessary for fabrication. Additionally, the same sets of geometries were used to generate illustrative presentation materials. Drawings were produced in schematic form, as were slides and transparencies for overhead projection.

Mobile Robotic Aid

The mobile base consists of a chassis made of bolted-together aluminum-nylon sandwich panels. The roughly triangular nature of the chassis, coupled with the generally rectangular or circular nature of most of the internal components, leads to a difficult design problem. Often the solution is to explore various parts-fitting options, and then choose the least-aversive configuration. This process is greatly facilitated by the use of an interactive design tool such as CADAM, and was exploited to the fullest on the version-2 prototype.

The chassis of the first prototype was originally drawn by conventional drafting techniques, and subsequently transferred to the CADAM environment. The original configuring of the internal components occurred before the transfer, and the ad-hoc nature of the disposition of the internal componentry is evident.

A further point to emphasize is the advantage in being able to share resources and drawings. In this academic design environment (Stanford University's Dept. of Mechanical Engineering), often one person, typically a graduate student developing a prototype in the context of his thesis work, is responsible for a relatively small part of the whole, yet that work must mesh with the rest. Until now, three to five designers have been involved in various aspects of the work, and the integration of their efforts has been largely automatic because the layout drawings and the individual parts drawings were available to all.

SUMMARY

In general, RR&D engineers have found computer-aided design to be a useful addition to their repertoire of design techniques. Although the system available to these engineers is not ideally suited to supporting the initial stages of conceptualization, it is extremely useful once a basic working set of parameters has been established. The ability to manipulate large sets of geometry in a non-destructive manner, performing various transformations and comparisons with other geometries, provides the freedom to test alternate hypotheses quickly, allowing for either reducing the project turnaround time or, alternatively, for generating a wider range of solutions in the available time frame. Also, the facility with which an anthropometric model may be generated, scaled for a specific application and superimposed on the design model, is of great importance to the effective design of the human-machine interface, critical to the results of any rehabilitative engineering project.

ACKNOWLEDGEMENTS

Sunburst I and II and an earlier prototype, the Handbike Tandem, were developed at the Palo Alto VA Rehabilitation Research and Development Center. The Handbike Tandem project was carried out in conjunction with the University of British Columbia, Department of Athletics, under a grant from the British Columbia program for the 1981 International Year of the Disables Person. Custom frame building and design consultation were provided by Gary Hale of Gary Hale Bicycles Eugene, Oregon (Sunburst I and Handbike Tandem), and by Keith Bontrager of Santa Cruz, California (Sunburst II).

Turning frame design has been carried out under a grant provided by the Paralyzed Veterans of America.

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The Design of a Fixation Detection Algorithm for Ocular Communication and Control

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ABSTRACT - An ocular control strategy particularly well suited for rehabilitative aids has been developed. Signal processing algorithms are presented that employ a solid state imaging array to accurately and quickly provide data representative of visual gaze for a eye position measurement method based on corneal reflection. The algorithms consist of a SCAN mode for image acquisition and a TRACK mode for target identification. The TRACK mode incorporates a strategy that defines a window around a possible image to substantially increase overall system speed. The system implementation is based on a commercially available and low cost computer peripheral called the MicronEye™. Measurements of resolution and linearity for both a mechanically simulated eye and human subjects indicate that an ocular communication system could successfully be implemented with the MicronEye and the developed algorithms.

INTRODUCTION:

The concept of visual gaze as a control strategy has been suggested for a variety of applications [Calhoun, et al., 1984; Luster and Vanderheiden, 1975; Eye-Controlled Switch, 1979]. The principle common to and underlying all ocular control systems is the basic method of operation. The user is presented with an array of targets that represent selections for the control environment. The system measures the user's gaze with respect to the target array and a selection is made by fixating upon one of the targets for a given period of time (dwell time). This modality presupposes that the user possesses the ability to maintain gaze on distinct points in the visual field, and requires a system capable of determining gaze with sufficient accuracy to resolve between neighboring selection targets.

The approach suggested by Mackworth [1958] has been chosen by several researchers in the field of rehabilitation [Rinard, 1978; Rosen, 1978; Fincke, 1980; Demasco, 1983]. In this approach, a video camera is mounted on the head and the position of the eye is measured with the corneal reflection method [Carmichael and Dearborn, 1947]. Common to these systems is some method of compensation for head movement.

Rinard [1983] utilized a 1k byte dynamic RAM as a 32 x 32 matrix image sensor. This device functioned as a low cost corneal reflection monitor with a target resolution of 9 items. Subsequent technological advances have resulted in memory devices of increasing capacity. Imaging sensors based on a larger chip (e.g., 64k byte) potentially extend the total system resolution. This enhancement of resolution is obtained, but significantly increases the image processing requirements.

The ocular control strategy of the present work is based on the corneal reflection method, and includes algorithms that allow the use of larger imaging arrays to accurately and quickly provide fixation determination. The implementation of the algorithms utilize an optical imaging array called the MicronEye that is based on a 64k byte dynamic RAM. This device is an inexpensive microcomputer peripheral consisting of two arrays of 256 by 128 photosensitive pixels (only one is being used to image the corneal reflection). Although the MicronEye is used in this work, the algorithms developed are applicable to any system where corneal reflection is represented as an image in an array of pixels.

FIXATION DETECTION:

In an ocular control system, the major premise is that the position of the individual's fixation as representing the selection of a target from the visual display. Eye movements represent transition between selections and consequently are of little importance to the measurement process. Elimination of eye movement measurement makes a software implemented Fixation Detection Strategy possible. A general methodology exists for the determination of fixation. Each component of the process is conceptually independent, which allows for the convenient implementation and analysis of a variety of algorithms:

1. Expose Array - Pixel arrays, like photographic film, are essentially light integrating devices. The array must be exposed for some time period for an image to form.

2. Acquire Image - The exposed image must be transmitted to the host computer and stored in memory.

3. Process Image - With the image stored in memory, signal processing algorithms can be utilized to calculate the position of the corneal reflection. There are four basic components to this process:

    1. Identify bright pixels
    2. Identify corneal reflection
    3. Determine position
    4. Determine if fixation has occurred

The major feature of the Fixation Detection Strategy is a method called windowing. **Windowing** is defined herein as a mode of operation where a process, normally applied to a set of signal data, is applied to some subset. That subset is defined by the previous history of that signal. An example of windowing is in the use of imaging arrays in satellite surveillance systems [Neel, 1981]. In this environment the sensor "stares" at a

very large area in an effort to detect movement of objects. The operating algorithms examine a window around potentially moving objects to validate their existence and track their path. This approach represents an important filtering stage in the processing of the large quantities of data typical for these systems.

The Fixation Detection Strategy embodies signal history to increase the efficiency of measurement through the use of windowing. A window is placed around the last known image position. The area of the window is examined for the presence of an image. If an image is found, the window is repositioned with the image at its center. This process is repeated as long as the image is in the window. If the image is travelling at a sufficiently high speed, it will not appear in the window, and must be found with a subsequent scanning operation.

ALGORITHM DEVELOPMENT:

The development of a Fixation Detection Algorithm was accomplished through a series of design and analysis phases. The first design phase implemented a reference algorithm. This module examined the entire array for bright pixels and calculated the centroid of the largest cluster. Although this program was extremely slow, it provided the most accurate static measurement of corneal reflection position.

The Fixation Detection Strategy was designed around two component algorithms. The **SCAN** mode searches the imaging array for the presence of the corneal reflection. Given an initial image position, the **TRACK** mode then defines a window around the image, and repeatedly examines the neighborhood defined by the window to determine if a fixation has occurred. If the TRACK mode loses image position, as it may during a rapid saccade, SCAN is once again initiated. In the case of ocular control, the window defined by TRACK will contain the image for the duration of the fixation.

The algorithms were implemented with several filtering models that accomplished various degrees of noise rejection. Greater filtering capabilities were obtained at the expense of execution time. These algorithms were evaluated for execution speed and accuracy.

The evaluation was performed with a mechanical jig that simulated eye rotation in one axis. This mechanical eye represents an idealized method for determining corneal reflection position. The jig consists of a stainless steel sphere that rotates off center about a pivot that represents the center of rotation of the eye. Connected to this pivot is an aluminum arm 80 cm in length. This arm traces an arc during rotation to provide measurement of $\pm.06$ degree precision over a range of 96 degrees. Because the Fixation Detection Strategy depends on a relatively static eye position, it was not necessary to implement dynamic simulation characteristics (such a system was proposed by Abramov and Harris, [1984]).

An intermediate filtering model was chosen for algorithm implementation. This filter incorporated centroid determination of the first bright cluster of pixels. Relative to the reference algorithm, the SCAN mode was 1.2 to 5 times faster depending on image position. The TRACK mode was found to be 7 times faster than the reference. In the simulated environment, accuracy was not compromised by the increased speed of these algorithms.

RESULTS:

The final implementation was evaluated for resolution and linearity in both measurement axes. These experiments were performed with the mechanical simulated eye. Figure 1. shows a plot of eye rotation versus measured position in the horizontal axis. The worst case resolution can be seen in the neighborhood of -2.0 degrees, where 4 consecutive measurements representing $\pm.25$ degrees are indistinguishable. The statistical analysis obtained from this data indicates an average resolution of $\pm.08$ degrees. Similar measurements were performed in the vertical axis and are shown in Figure 2. The average resolution obtained was $\pm.06$ degrees with a worst case of $\pm.19$ degrees. The higher vertical resolution (i.e., lower number) is the result of closer pixel spacing in that axis. Linearity was measured for a $\pm$ 20 degrees in the horizontal axis and $\pm$ 16 degrees in the vertical axis. In both cases linearity was approximately 98%.

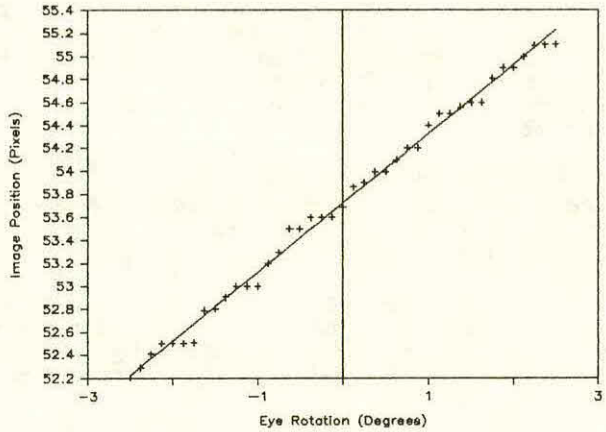

Figure 1. Horizontal Resolution Measurements Simulated Mechanical Eye

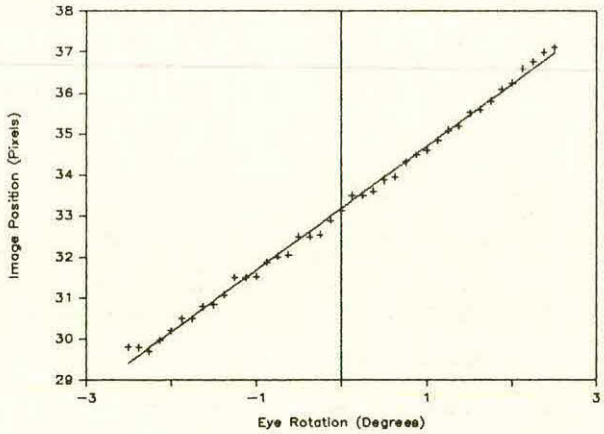

Figure 2. Vertical Resolution Measurements Simulated Mechanical Eye

The experiments were repeated with human subjects using a bite plate to stabilize head position. Resulting resolution and linearity measurements were decreased by approximately 50%. This was due to corneal surface irregularites and increased complexities in the optics associated with the testing stand. These experiments indicate that this system is capable of measuring ocular rotation with an accuracy of $\pm.5$ degrees. This accuracy is within the limits of the corneal reflection technique and is suitable for a direct selection system with at least 64 targets.

CONCLUSION:

The motivation for this work was to utilize an inexpensive microcomputer peripheral as an eye position measurement system. This device will serve as a subsystem in the Line Of Gaze communication system that also incorporates a head position monitor. It is also planned to incorporate this device into a head referenced system that will provide lower resolution ocular control at a reduced cost to the end user.

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14.1

INTRA-ABDOMINAL PRESSURIZATION FOR LOW BACK REHABILITATION: FAILURE TO REDUCE SPINAL LOADS

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ABSTRACT

Abdominal muscle strengthening or support is frequently used in low back pain rehabilitation, based on the assumption that intra-abdominal pressure reduces intervertebral disc pressure and the forces on back muscles. However, no direct testing of this assumption has previously been performed. Ten subjects performed isometric trunk extension pulls at 33%, 66%, and 90% of maximum voluntary effort for 10 seconds in both an upright and 45° forward flexed posture. Force, erector spinae EMG, and intra-abdominal pressure (IAP) were simultaneously recorded. During these pulls, the subjects consciously varied their intra-abdominal pressure using an uncoached mode, a relaxed (pressure drop) mode, or a Valsalva (pressure increase) mode. No significant change in EMG levels for constant trunk extension effort were caused by changes in intra-abdominal pressure, indicating that IAP reduce the loads on the spinal muscles during lifting.

INTRODUCTION

Intra-abdominal pressure has long been hypothesized to play an important role in providing support to the spine during lifting tasks (1). This is based on an observed correlation between IAP and trunk extension efforts (2). Unfortunately, this assumption has never been directly tested, thus no cause-and-effect relationship has been established.

MATERIALS AND METHODS

Ten healthy male subjects (ages 20-35) were recruited. The protocol used in this experiment had three components:

1) Each subject was instructed to perform a force-EMG calibration by performing isometric trunk extension pulls using maximum voluntary effort while wearing a shoulder harness. A maximum pull was done in both an upright and 45° forward-flexed posture for 10 seconds. During the actual experiment, the subjects were told to pull at 33%, 66% or 90% of these maximums for 10 seconds. Each volunteer maintained a constant effort during the experiment by viewing a digital readout of the force on a load cell. A thirty second rest was allowed between pulls to reduce subject fatigue.

2) EMG measurements were made simultaneously using Ag-$AgCl_2$ sensing electrode pairs placed over the left and right sides of the erector spinae at the level of L3. A ground electrode was placed at the T1 level.

3) For each test at a specified effort level and posture, the volunteers were instructed to vary their intra-abdominal pressure as follows: uncoached pressure for the first five seconds of the 10 second pull, followed by either a Valsalva pressure or relaxed pressure for the second five seconds. The Valsalva maneuver generates high levels of intra-abdominal pressure and is performed by tensing the abdominal muscles and closing the glottis. In the relaxed pressure mode, the subjects attempted to reduce their IAP by viewing a digital readout of the pressure during the experiment. The intra-abdominal pressures were measured by using a water-filled 6 French nasogastric tube connected to a Howmedica Interstitial Pressure Monitor. Between tests, the subjects performed sniffing maneuvers to allow determination that the nasogastric tube remained in the abdominal cavity and did not become clogged.

Data for all four channels were recorded on FM tape using a TEAC R71 Data Cassette Recorder. The two EMG channels were processed using an analog RMS/DC converter prior to data acquisition. The information was then fed into an IBM PC/XT computer using the Labtech Notebook data acquisition software produced by Laboratory Technologies Corp. and a TECMAR Labmaster PGL 12 bit A/D converter set on ±1 volt. All data were acquired at 50 Hz. Next, a 3-second constant-force segment was selected from each 5-second mode of each 10 second test. Each 3-second segment was then integrated to obtain mean force, IAP, and EMG levels for each test. The EMG data from the left and right sides were then averaged. In order to compare EMG activity between the subjects, EMG in volts was converted to percent maximum voluntary effort (%MVE) using calibration curves obtained from each subject.

RESULTS AND DISCUSSION

During uncoached extension efforts, there is an approximately linear relationship between %MVE and EMG, as well as between %MVE and IAP. This is consistent with previous reports.

Figure 1 presents the the relationship between the measured change in EMG (y-axis) and the controlled change in IAP (x-axis). The values of all subjects at all effort levels and postures are shown. Points on the left side of the Y-axis represent tests in which the volunteers attempted to relax their abdominal muscles, thereby reducing their IAP below uncoached levels. Points on the right side of the graphs represent tests in which the volunteers attempted to increase their IAP above uncoached levels. Because not all the the subjects were able to produce relaxed pressures that were less than their uncoached pressures, somewhat more than half of all the data points are to the right of the Y-axis. The least squares linear regression line of the data has been plotted on the graph, as well as a 90% confidence band for the expected position of that linear regression line.

Table 1 presents the mean slopes of the linear regression lines for each effort level and posture

separately, and for the regression line of the combined data that is shown in Figure 1. A 95% confidence limit (C.L.) for each parameter of the regressions is also included. Note that the slopes of the lines at all effort levels and postures are not significantly different from zero.

Thus it can be seen that changes in intra-abdominal pressure have almost no effect in reducing the EMG of the erector spinae muscles during lifting tasks over a wide range of effort levels and abdominal pressure. Abdominal muscle strength-EMG, widely used for rehabilitation of low back pain patients, does not appear to operate according to the mechanism originally proposed by Bartelink in 1957. Further research is in progress in our laboratory to determine whether IAP also fails to provide any load relieving effect in low back pain patients, and whether lumbosacral corsets or braces allow such an effect to occur. These findings will affect the manner in which low back pain rehabilitation is attempted.

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Figure 1.

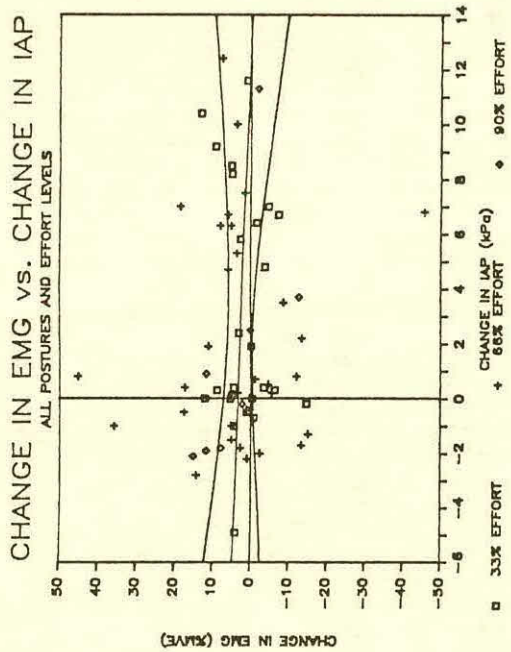

Table 1.

95% CONFIDENCE LIMITS OF SLOPE AND INTERCEPT OF LINEAR REGRESSION LINES FOR CHANGE IN IAP vs. CHANGE IN EMG FOR EACH POSTURE AND EFFORT LEVEL

Posture	Effort	Slope	95% C.L.	Intercept	95% C.L.
UPRIGHT	33%	0.24	±0.61	1.41	±3.05
UPRIGHT	66%	0.08	±1.71	0.59	±2.44
UPRIGHT	90%	-2.10	±14.85	8.24	±2.20
FLEXED	33%	0.03	±1.84	0.37	±3.55
FLEXED	66%	-1.23	±3.69	9.39	±13.78
FLEXED	90%	-0.85	±4.55	2.53	±27.39
ALL POINTS		-0.25	±0.74	3.20	±3.26

THE RESPONSE OF THE SEATED HUMAN TO VIBRATION AND IMPACT

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ABSTRACT

Low back pain has been shown to occur more frequently among vehicle drivers than control groups. Previous studies of vibration transmission through the body has employed sinusoidal excitation, but the body's response to impact has been not been studied in depth. This study investigated the response of the spine to both impact and sinusoidal excitation. Five males and 5 females were evaluated for seated mechanical response using both methods. Both the magnitude and phase spectra of the acceleration transfer functions were used as outcome measures. Comparisons were made between test methods and between postures (erect and relaxed seated). Spectra were compared at 2-4, 4-8 and 8-16 Hz intervals using a sign test. In 11 of the 24 cases, either test method or posture had a significant effect on mechanical response. This suggests that the impact method may be a useful and rapid testing tool.

INTRODUCTION

Low back pain and degenerative diseases of the spine have been shown by many studies to occur more frequently among vehicle drivers than in representative control groups. Thus, there has been great interest in measuring the transmission of vibrations through the human body. The majority of these studies have utilized accelerometers at the head and at the seat surface so as to establish the mechanical response of the system. Usually accelerations at either a helmet or a bitebar are measured and related to input at the seat to give the transmissibility (transfer function). These studies give valuable information about the overall response of the subject, but only indirect information about the response of various segments of interest, for example, between the seat and the pelvis and between the pelvis and different parts of the lumbar spine. For these reasons, there has been a limited number of attempts to measure vibration directly at the regions of interest. Many reports in the literature regarding response to vibration are contradictory. Some different results can be attributed to different experimental techniques, but not all. Some workers report considerable differences in biodynamic response between human subjects under the same experimental conditions. In addition, even similar experimental conditions have produced different results. An attempt has been made to resolve some of these contradictions by investigating the relative response as measured by transducers mounted to the body surface compared to those rigidly attached to the skeleton. The present study is a comprehensive investigation of the response of the spine to both impact and sinusoidal excitation, looking at both the overall and regional responses of the system.

MATERIALS AND METHODS

Sinusoidal Excitation Apparatus. The sinusoidal apparatus used is a resonating system consisting of two parallel wooden beams, simply support- ed, that can vibrate. By changing the length between the supports, the resonance frequency of the system can be changed from 2 to 15 Hz. The seat is fixed in the middle of two beams. The input energy is produced by an electric motor with an attached eccentric device. An accelerometer is placed onto the seat to measure vertical acceleration input to the subject.

Spinal Impact Apparatus. A platform was suspended by eight rubber springs (designed for use as spring lines when docking boats) and guided by two linear bearings. The rubber springs had a viscous component so that they would tend to dampen system oscillations after the device was activated. Adjustments were made to the spring platform after the subject was seated so that the platform was always struck at the same point in the swing of the impact pendulum. The pendulum was released by means of disc brakes and a manual trigger started the data collection. The starting point of the pendulum was always at 10° above the horizontal. This translated to a transfer of 1.9 joules of energy to the platform-spring-human system. A known impulse versus time impact was applied to the platform. This system tended to produce a relatively square impulse force in the time domain. The natural frequency of the system was 1.8 Hz, well below the frequencies of interest. The frame natural frequency was 90 Hz and the system was unaffected by crosstalk in other directions. The system was calibrated by attaching known weights and masses, computing the natural frequency and comparing this to the measured response.

Ten subjects (5 male, 5 female) were evaluated for their seated mechanical response using the two different methods described above. The subjects were aged between 15 and 45, weighed between 65 and 90 kg and were free of low back pain and other health problems. Two different postures were maintained by the seated subjects. The first was the erect posture where the back of the head, the peak of the thoracic spine and the midpoint between the posterior superior iliac spines were colinear. The subject's posture was then adjusted so that this line was oriented normal to the seat. The second posture studied was relaxed, with the only constraint being that the subjects had their eyes looking forward to a local horizon.

The mechanical response was determined by recording the measured acceleration versus time at both the seat and at a bite bar held in the subject's teeth. Transfer functions could then be determined for each subject.

Data were recorded on two channels of a four-channel, open reel FM tape recorder (Racal

Storage 4) at 38.1 cm per second. Accelerometer data for both channels were amplified and low pass filtered at 50 Hz (fourth order filter) to an A/D converter onto a Digital PDP 11/23 computer. These data were graphed in both time and frequency domains and transfer function (magnitude and phase) relationships were determined by spectral analysis. This method was employed for both the impact and spinal exciter devices.

The sinusoidally excited device was utilized to expose subjects to both postures and vibrations of 1 M/sec^2 at discrete frequencies, from 2 to 14 Hz. Fifteen seconds of data were recorded at each of the vibration exposures. The vibrator was adjusted to produce the discrete vibrations by adjusting the input displacement and frequency at an eccentric displacement drive and by adjusting the end support locations of wooden beams on which the seat was placed. Ten repeats were recorded in each of the two postures.

Two series of tests were run using the spinal impactor. In each series, the two postures (erect and released) were studied. In the second series, the subjects listened to light music via lightweight headphones in order that they have no aural cue as to the release of the impact mechanism. Thus, they were not subconsciously able to prepare their muscles for the impending impact. However, it was necessary for the researchers to remind the subjects not to move or sway with the music. Ten repeat impacts of each posture were recorded for each subject.

Data were obtained for each subject to determine the effects of both test method type (vibration or impact) and posture (erect or relaxed). The data were presented in the form of both magnitude and phase spectra of the acceleration transfer functions. Comparisons were then made between like spectra to find the effect of (a) test type with the same posture and (b) posture type with the same test method. The spectra were evaluated in a simple manner. The averages were computed in three intervals: 2-4 Hz, 4-8 Hz and 8-16 Hz. Spectra were superimposed over each other and were studied to find the larger average value. Occurrences of the larger values were then recorded for all ten subjects for all the comparisons. A sign test was performed to determine significance of the differences of the samples.

RESULTS

Two types of test were performed to establish (a) the effect of test method on the system and (b) the effect of posture. The test method is an important consideration because an impact method can be made portable and simple to use. Hence, results which show no significant difference between test methods would suggest the impact method could be used as well as the vibration method. Significant differences due to the methods may require more investigation to find the exact nature of the differences. The effect of posture, however, will reflect the behavior of the seated human and can be compared to previous similar work.

In each comparison, the response is always similar. For example, in both the relaxed and erect postures, the magnitude of the vibration response (in the 2-4 Hz region) exceeds that of the impact response. Also, in each of the vibration and impact magnitude comparisons (in the 2-4 Hz region), the relaxed posture exhibits a greater magnitude than the erect posture.

In the 2-4 Hz region, the vibration magnitude and phase response is greater than the impact response. This also occurs in the magnitude spectrum in the 4-8 Hz region. The impact response exceeds the vibration response in the phase spectrum of the 4-8 Hz region and both the magnitude and phase spectra in the 8-16 Hz region. Posture has a more consistent effect. Excepting the magnitude spectrum in the 2-4 Hz region, the erect posture response was always greater than the relaxed posture response for both the magnitude and phase spectra in all of the 2-4 Hz, 4-8 Hz and 8-16 Hz regions.

Thirteen of the 24 comparisons made were not significantly different. The only other remarkable trends were that there was no significant effect due to posture in the magnitude spectra in the 2-4 Hz and 4-8 Hz region and there was no significant difference in the magnitude spectra due to the test method in the 8-16 Hz region. There were seven other comparisons which showed no significant differences between either magnitude or phase spectra.

Highly significant differences did occur (a) due to test method in the erect posture in the magnitude spectra in the 2-4 Hz region and (b) due to test method in the erect posture in the phase spectra in the 4-8 Hz region.

DISCUSSION AND CONCLUSIONS

There are considerable advantages in using the impact method. The method provides rapid testing over a large frequency range. Further, it can be made portable, is easy to use and readily allows for testing different levels of such independent variables as posture, muscular contraction and seat characteristics. In this paper, we have validated the impact method by comparing the results with those produced by sinusoidal excitation.

For those conditions in which there was no significant difference between results obtained with the vibration and the impact method, it is reasonable to assume that one method can be substituted for the other. Further analysis of the data may explain why there is a difference due to test methods in some cases. Some differences may be due to the non-linearity of biological tissue and may explain the different responses in different frequency ranges. One major difference in the systems which must be kept in mind is the support of the legs. The vibration test method provided a platform on which to rest the feet, whereas the impact method did not.

CONCLUSIONS

An impact apparatus has been tested and proved to be useful in the study of mechanical responses of the seated operator. The impact response has been compared to that of sinusoidal excitation. Certain differences in response are attributed to the role of the musculature.

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REHABILITATION IMPLICATIONS OF IN VIVO HIP PRESSURE MEASUREMENTS

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ABSTRACT

Pressure in the human hip joint is measured in vivo with an instrumented endoprosthesis employing ten focal transducers integral with the pseudofemoral head. Concurrent kinematic data permit coordinate transformation from transducer positions to areas of cartilage on acetabulum. This paper focuses on early post-surgical and rehabilitation data to evaluate consequences of different patient management procedures and rehabilitation walking aids following major hip surgery. Pressure, kinematic, and dynamic (force plate) gait data at six-months post-implantation are also described for comparison with the early recovery and rehabilitation data.

INTRODUCTION

The functional assessment of the consequences at the hip joint of alternative patient management and rehabilitation procedures could in principle guide optimum choices for patients undergoing major hip surgery. No such quantitative information has existed. The very limited force data extant from instrumented hip nails, endoprostheses, and total hip replacement prostheses have included no immediate post-surgical data nor have they compared alternate rehabilitation procedures (1,2,3).

MATERIALS AND METHODS

Figure 1 illustrates an endoprosthesis whose physiological function and material biocompatibility are comparable to the standard femoral head reconstruction prosthesis, some sixty-thousand of which are implanted annually in the United States (4). Recesses (3.96-mm. diameter) are electron-discharge machined from inside the cobalt-chromium-carbon alloy, load-bearing hemisphere at fourteen locations, leaving 0.445-mm. thick, slightly convex diaphragms whose external surfaces are integral with the polished sphere of the pseudofemoral head. Diaphragm deflection due to pressure from the opposing acetabular articular cartilage is center-point detected and measured with a four-arm strain-gauge bridge diffused onto a single-silicon-crystal cantilever beam. The pressure transducers are serially excited by an oscillator and a multiplexer switch, with each isolated transducer output modulating an FM telemetry transmitter, all contained within the hermetically sealed endoprosthesis ball. The resulting pulse-amplitude-modulated signal, together with high and low calibration pulses, is transmitted at a frame repetition rate of 253 Hz. from an antenna at the distal end of the prosthesis intermedulary stem. The otherwise passive electronics are powered inductively from an externally mounted 100 KHz transmitting loop mounted about the patient's thigh during data acquisition(5).

Interpretation of the pressure data requires concurrent acquisition of data on the kinematics and dynamics of the subject's movement. The locations of the femoral head transducers on the acetabular cartilage surface are calculated using kinematic data on the position and orientation of the subject's thigh segment relative to the pelvis. Similar kinematic data, but also including that from the shank and foot, permit generation of computer graphics displays of the subject's pelvis/thigh/ shank/foot positions and orientations simultaneous with display of the transducer locations and the corresponding pressures. The kinematic data are acquired via a unique hardware/software system (6), employing four opto-electronic Selspot II cameras and the M.I.T. TRACK^{c} computer program. Photostereogrammetric reconstruction of the 3-D position of serially illuminated but multiple infrared LED's on rigid arrays mounted on the body segments, followed by the aggregation of individual LED position data, permits precise calculation of the 6 degrees of freedom of individual body segments with very high precision. Foot-floor force vector data from two Kistler forcemeasuring platforms complete the data stream. Kinematic and force-platform data are acquired at 153 or 200 Hz, which assure high fidelity recording of the frequency components of gait (7). For the data reported herein, high frequency cutoff filters were applied to the raw data, the cutoff frequencies being 10 Hz. for the pressure data and 15 Hz. for the kinematic data.

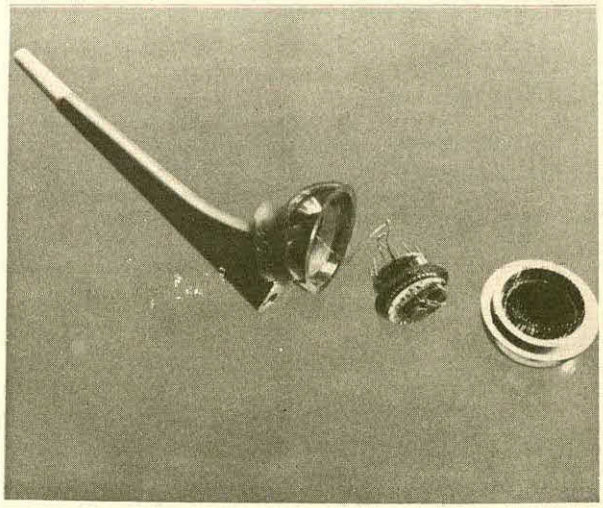

Figure 1

In June, 1984, an active seventy-three-old female (68 Kg. and 1.68 m.) with a displaced fracture of the right femoral neck consented to

the implantation of the instrumented prosthesis in substitution for the standard orthopaedic device. The exposed acetabular cartilage was carefully inspected visually to insure that it was normal. The importance of proper fit of any endoprosthesis to achieve optimum distribution of interarticular pressure has been demonstrated by our prior in vitro studies (8, 9). Optimum fit of the natural acetabulum to the 47.5-mm. diameter spherical prosthesis head was achieved using 1) go no-go gauges on the excised natural femoral head, 2) clear acryllic hemispheres in 1-mm-diameter increments observed as they contacted the acetabular surface, and 3) the instrumented prosthesis to check the uniformity of the individual transducer readings as the prosthesis was loaded into the acetabulum intra-operatively.

RESULTS AND DISCUSSION

We have reported elsewhere (10,11) on cartilage mechanical, tribological, and muscle co-activation aspects of this de novo data. Here, we focus on selected aspects of patient management and rehabilitation procedures.

To put this early post-operative data in context, however, we first present data illustrative of more normal gait. Figure 2 displays data as the subject passes through the single-leg stance of normal walking at six months post-operative. TRACK position and orientation kinematic information are illustrated via a novel, clinically useful, three-dimensional, computer graphics dynamic display (12). The force plate vector (below the foot) represents 103 percent of body weight. The pelvic diagram on the right shows the concurrent projections of the femoral head transducers on a two-dimensional map of the acetabular surface and lists the pressures at this instant. Note the wide ranging and high individual pressure readings. These in vivo data correspond favorably to our extensive in vitro series (8) with pressure values just about what we measured in vitro, and several times higher than the putative assumption of 2 to 3 MPa. based on earlier research (13, 14).

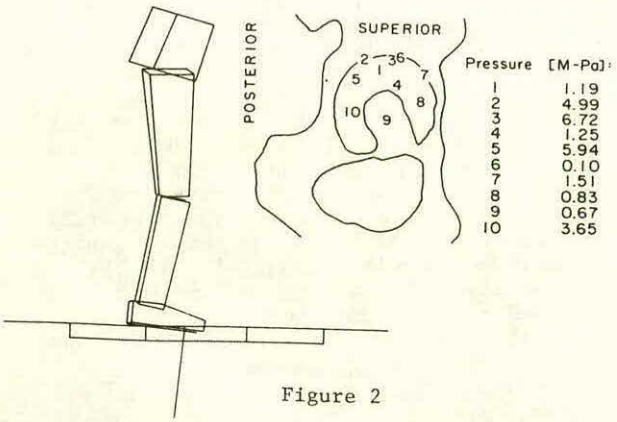

Figure 2

With this preview of the magnitude of pressures experienced in normal activities, we now return to early post-operative data. Table I lists the maximum pressures registered during a selection of early post-operative patient management procedures. Note, for example, the relief the balanced suspension offers over the unsupported leg during immobilization and the consequences of moving the patient onto a bedpan.

In rehabilitation following all major hip surgery, a progression of walking aids are employed--see Table II. Very early, parallel bars partially support body weight; a maximum pressure of 3.4 MPa. was registered. Using a four-legged walker at two weeks post-operative, the maximum pressures ranged between 3.8 and 4.0 MPa. and were in the same range (3.8 maximum) at six months post-operative. Figure 3 shows the time pressure profile at the transducer which registered the highest pressure; Figure 4 displays the same data as a 3-D time progression of vectors, each showing position on the acetabular surface and pressure magnitude. The view is from the center of the body looking through the pelvic bone toward the back surface of the acetabulum in a posterior and downward direction. Table II also catalogues typical maximum pressures while using the walker, crutches, and a cane. Note the significant pressures in the hip joint when no external load is applied on the leg and the pressure differences between knee extension and

TABLE I

Typical Maximum Pressures Recorded During Patient Management Procedures Conducted during the First Two Weeks Post-Operative

Immobilization Procedure	Pressure (MPa)
Balanced Supension	0.67
Buck's Traction (5 pounds)	1.50
Abductor Splint	2.58
CPM Machine	1.71
Unsupported	2.11
Mobilization Procedures	
Onto Bedpan	3.21
Tilt Table	2.00
CPM Machine	1.71
Bicycle	1.59

TABLE II

Typical Maximum Pressures Recorded While Walking with Assistive Devices

	(MPa)
Parallel Bars	3.4
Walker	4.0
Crutches	
Foot Off Ground	
Knee flexed	2.4
Knee straight	4.1
Light Toe-Touch	2.3
Partial Weight Bearing	3.9
Cane	
Light Force (15 pounds)	5.1
Medium Force (30 pounds)	4.8
Maximum Force (50 pounds)	4.8

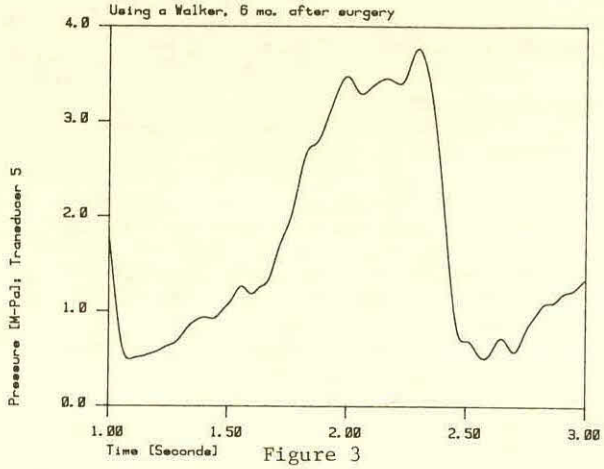

Figure 3

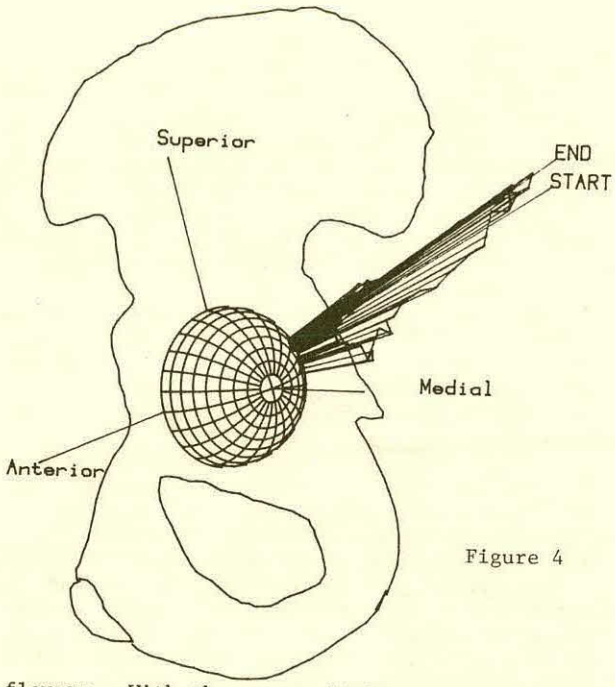

Figure 4

flexure. With the cane, slight to non-existent maximum pressure changes occur as the force exerted on the cane increases. Almost doubling the cane load does not affect the maximum pressure.

The high pressures measured on acetabular cartilage in the absence of external load on the leg, and the apparent insensitivity of joint pressure to leg loading, appear perplexing until one considers the contributions to hip joint load of agonist/antagonist muscle co-contraction across the joint, with only the net difference producing motion. Thus the pressures measured in the joint when the body is supported by crutches result from muscle co-contraction to support the leg mass against gravity.

The most dramatic example of the influence of co-contraction to date occurs during rising from a chair. This movement is slow; thus the inertial components are small. Yet at one year post-operative, a maximum pressure of 18 MPa. has been registered in the posterior region of the subject's acetabulum. For those unfamiliar with metric system measurements, this corresponds to 2,610 pounds per square inch!

CONCLUSIONS

The ability to measure the pressure on acetabular cartilage in vivo is producing de novo and surprising information on the consequences at the hip joint of a wide range of supposedly benign and graded post-surgical procedures and exercises. One optimally fitted, (100 on the Harris Hip Rating Score at one year post-operative) wonderfully cooperative subject is producing a wealth of useful data. Subsequent implantations will illuminate the extent of subject variability. Superb implant instrumentation (still fully effective at one-and-one-half-year post-operative) in combination with unique movement analysis capability, is producing extraordinarily rich scientific and clinically relevant information.

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14.4 BIOMECHANICAL PROBLEMS WITH CEMENTED TOTAL KNEE REPLACEMENT

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INTRODUCTION

Metal-plastic Total Knee Replacement (TKR) was introduced in the USA in about 1970. Among the first designs were the Polycentric, the Freeman-Swanson, the Geomedic and the DuoCondylar (1). During the 1970's, many other designs were introduced, some being adaptions to counter problems with the early designs, and others being new concepts. During that decade, frequent clinical problems were due to inadequate instrumentation, inaccurate component alignment, poor cement technique, excessive device constraint, absence of patello-femoral replacement, and lack of ligament balancing. However, institutional reviews published in the early 1980's, suggested that with some designs at least, problems occurred in only a few percent of cases at up to 10 years follow-up (2,3,4,5). We have recently studied two series of knee replacements, in order to determine what are likely to be the long-term problems with contemporary TKR. The first series was 90 removed TKR's to study wear and damage to components. The second series was 1069 clinical cases of Kinematic-TKR at up to 7 years follow-up.

ANALYSIS OF 90 RETRIEVED METAL-PLASTIC TKR'S

Materials

The series consisted of TKR's removed from patients at the BWH Hospital. The plastic was ultra-high molecular weight polyethylene. Clinical records were available for 82. Implant time ranged from 3 mos. - 10 years, mean 5.6 years. Twenty TKR's were in place for more than 8 years, only 5 for less than 2 years. The designs displayed a variety of femoral-tibial geometry.

Unicondylar	17	Geomedic	6
Duocondylar	12	Kinematic PC	5
Duopatella	23	Robert Brigham	3
Marmor	7	Others	17

The reasons for removal were:

Loosening, component shifting	35
Pain in unresurfaced area of joint	15
Sepsis	9
Pain for other reasons	6
Instability/deformity	8
Inadequate data	17

Methods of Study

The plastic components were examined using low power microscopy. Different wear morphologies were characterised and assessed on a scale of 0-6 for severity. Thin sections of plastic from each component were cut with a scalpel and examined under transmitted light microscopy. Five mm^3 blocks were cut from 20 components, and studied with scanning electron microscopy. Similar blocks of plastic from 21 components were used for molecular weight analysis, using high temperature gel permeation chromatography. (Jordi Associates, Millis, MA) Data was obtained of high, medium, and low molecular weight fractions.

Results

The observed wear and surface damage were as follows:

Embedded acrylic cement particles (43% of components). These were easily identified.

Cement particle craters (74%). These were identifiable by often jagged sides, dimples at the base, and often high abrasion surrounding.

Deformation (cold flow)(90%). This occurred in the main bearing area, due to excess contact stresses, near the edges due to subluxation, and on raised parts such as the tibial eminence due to digging in of the metal femoral component in internal-external rotation.

Burnishing (85%) This was glossing of the surface, which did not appear to result in much material removal.

Abrasion (95%) Scratches, grooves, and stretching of multiple fibers of plastic (~ 10-100um) on the surface, fell into this category. The causes was mainly three 3-body abrasion from worn plastic and acrylic fragments.

Discoloration (8%) Possibly caused by fluid embibement.

Delamination (38%) This was seen as sub-surface cracking, a raised flake or flakes of material, multiple cracks and fragmentation, and surface cavities. The subsurface cracks were often parallel to the surface and were at least 100um in width.

Sections of unused plastic often showed many small cavities or discontinuities. In some cases, fine lines could be seen in a network. This pattern became accentuated in used implants, and in some of the longer-term cases, almost complete breakdown of the material could be seen. Since the materal is compression molded by fusion of small granules, it is believed that the fine lines represented granule boundaries and that high stresses and inadequate fusion led to the breakdown.

It has been found in lab tests that wear increases with decreasing molecular weight (6). In our group, there was a wide variety.

	Low M.W.	High M.W.
Minimum	46,000	723,000
Maximum	654,000	4,670,000
Mean	148,000	2,300,000
S.D.	142,000	1,594,000

There was a close correlation between low and high m.w. at $p<0.001$

The overall wear/year correlated with best ambulatory status at $p<0.01$. For 39 components from one manufacturer (J & J) there was a linear relation between 'overall wear' and years of implant. (C.C.0.8) There was an inverse correlation of wear/year with molecular weight ($p<.02$). Particularly striking were compressive molded flat Marmor components (which would show high contact stresses) of high m.w. which showed minimal wear at 8-10 years follow-up.

The most striking finding from this wear study was the severity in many cases. Where there was delamination and intergranular cracking, it appeared that severe material breakdown would have occurred in ensuing years.

CLINICAL FOLLOW-UP OF 1069 KINEMATIC TKR

The first 1069 consecutive cases of primary Kinematic Condylar TKR carried out by nine staff orthopedists at the Brigham and Women's Hospital were reviewed. The clinical material was as follows:

1069 cases in 798 patients
667 complete clinical history, complications determined for all cases 12-90 years age range, 67 average
77% female. 81% cases over 60 years of age.
47.7% OA 46.8% RA 2.4% JRA
822 cases radiographic study

Follow-up times (years)
0-1, 386 cases 1-2, 191 cases. 2-3, 113 cases. 3-4, 67 cases, 4-5, 43 cases, 5-6, 20 cases, 6-7, 2 cases.

All clinical data was recorded in the computerized Joint Registry. Patients were seen at 6 months, 1 year and then every year. If the patient could not be seen, information was obtained by telephone. The most recent radiographs were measured for component and leg alignments. The cement-bone interface was studied for radiolucent lines (RLL) in 7 tibial zones (a-p) and 5 femoral zones (m-l).

Total RLL score was calculated as the sum of the RLL thickness (to nearest mm) x each zone. Thus a uniform 1mm RLL all round the tibia would have a score of 7.

Clinical Results

Postoperatively 93% of patients had none or minimal pain on all activities. 82% could negotiate stairs normally, and the others had to lead with the other leg. Range of motion averaged 2.5-107 degrees.

The Aseptic Revisions were:

Loose femoral component	0
Loose tibial component	1
Skin necrosis (patellectomy to close)	1
Patella button loose	6
Patella subluxing	5
Patella transverse fracture	2
Patella painful	1
Patella dislocation	1
	17 (1.6%)

Other complications were:	
Deep sepsis (9 retained, 1 exch.)	11 (1%)
Transient problems (skin, hematoma etc.) (no additional surgery)	29 (2.7%)

Radiographic Results

An RLL can be taken as an indicator of interface breakdown, but localised RLL may not be significant. The RLL's on the tibial component occurred primarily at the extreme medial or lateral zones. The inner zones had only half the incidence. RLL around the peg occurred in only a few cases. In the femoral component, the posterior and anterior flanges were most frequently affected. The scores on the components were mainly 1, indicating 1mm RLL in only 1 zone. The RLL's occurred in regions where it is difficult to achieve good cement pressurization and where high stresses and micromotion occur. The incidence of a score of 1 or more was 12.9% femoral, 26.5% tibial, and 6.3% both. The incidence of tibial cases with RLL increased from 21.3% at 0-1 years, to 45% at 5-6 years, but the numbers of RLL's with score > 1 did not similarly increase.

The data on cases with RLL scores ≥ 4 was:

Total number of cases	822
Cases with RLL score ≥ 4	21 (2.6%)

There were only 21 tibial cases (2.6%) with RLL ≥ 4. The data on these was:

Cement filling wedge-shaped defect, or thick cement layer	12
Both medial and lateral RLL, cause not evident	7
Due to varus or valgus deformity	2

DISCUSSION AND CONCLUSIONS

Aseptic loosening was not a significant problem in the Kinematic series. No femoral components and only one tibial component needed revision, the latter occurring due to deficient bone stock and component misalignment at surgery. The frequency of cases with RLL's increased over time, but most of the increases were of small RLL's at the sides of the component, consistent with a crack propogation mechanism initiated at the sides (7). The situation did not suggest a serious failure rate in the future. This result is consistent with other reports (2,3,4,5). Most of the thick RLL's occurred beneath thick cement layers, such as in a wedge defect. We recommend bone graft, or a metal wedge spacer, in combination with a long stem, for such cases.

Most mechanical problems were with the patella. The causes are not fully understood but include both design and surgical factors (8). Similar problems have been reported in other series (e.g.8, 9)

Short of a rigorous quality control of the plastic for internal porosity, intergranular bonding, and molecular weight, wear at 10 years and more is likely to be a serious problem in many cases. The effects are unknown but may include capsule fibrosis, inflamation, and component loosening.

All of the above applies to uncemented TKR as well, except that the interface is different with unknown long term results at this time.

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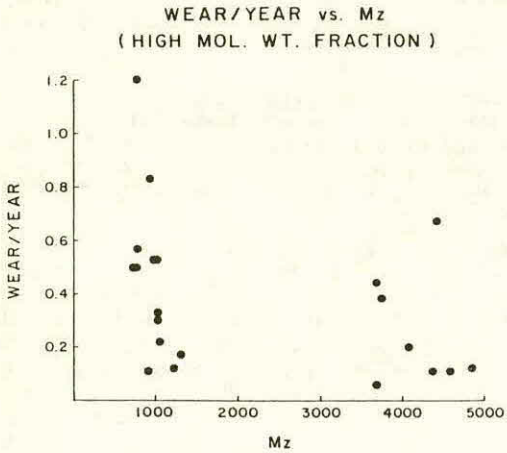

Figure 1

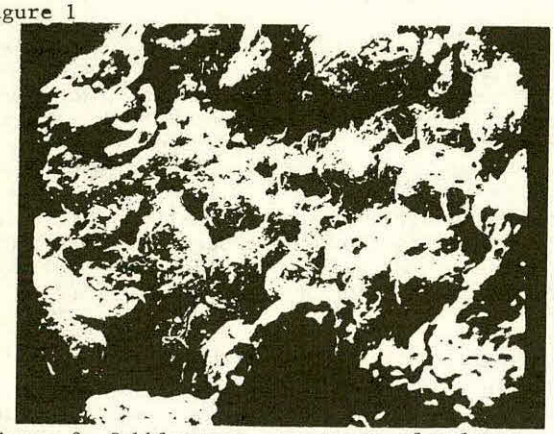

Figure 2 Cobblestone appearance of polymer granules breaking up.

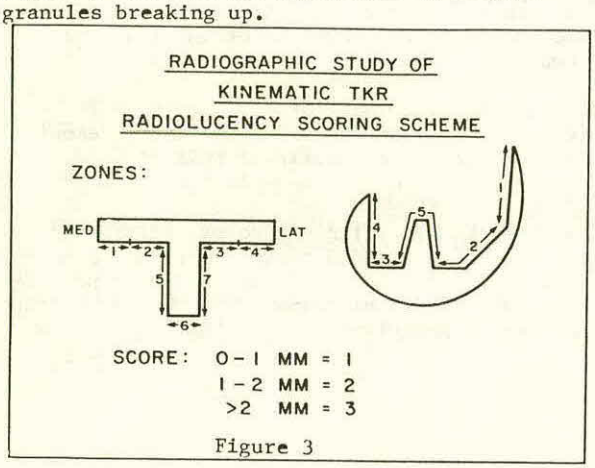

Figure 3

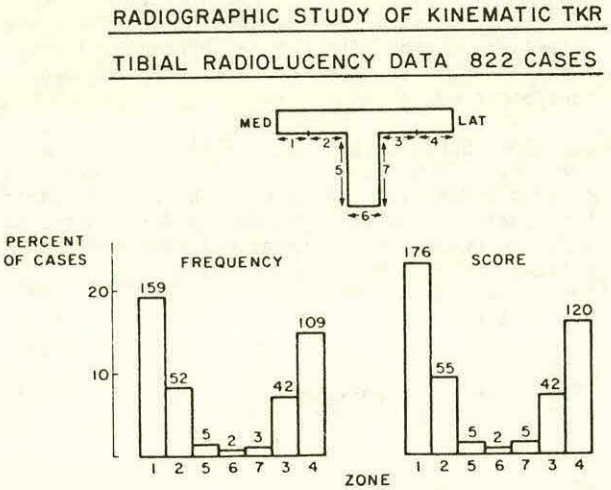

Figure 4

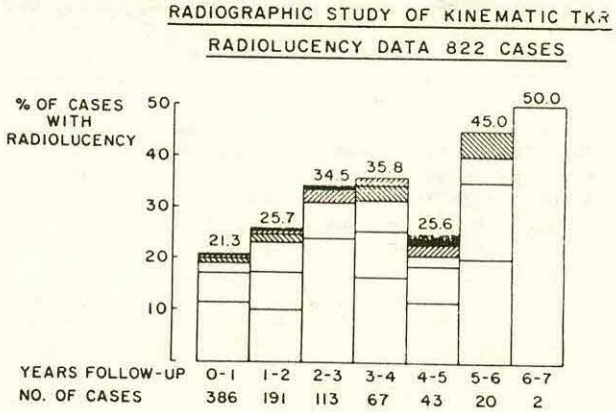

Figure 5

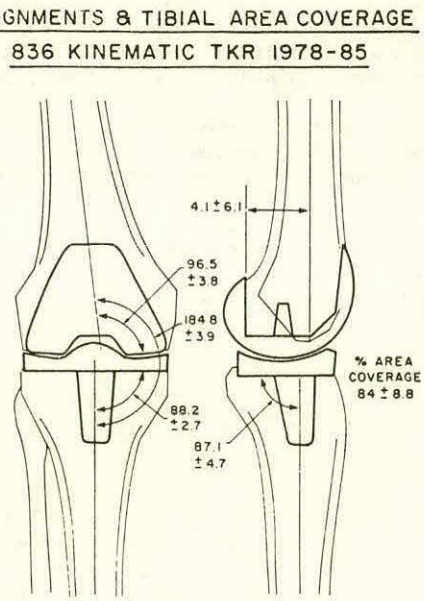

Figure 6

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COMPARATIVE ANALYSIS OF INTERLOCKING INTRAMEDULLARY FRACTURE FIXATION NAILS IN BENDING AND TORSION

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ABSTRACT

This study was undertaken to compare different designs of interlocking and unaugmented intramedullary nails, with respect to their stability in bending and torsion, and their resistance to rotational motion. Embalmed human femora with transverse midshaft fractures were instrumented with the following configurations of intramedullary nails: (1) Grosse-Kempf (non-interlocked), (2) Grosse-Kempf (interlocked), and (3) Brooker-Wills. Bending and torsional stiffnesses were determined by loading the femora in four point bending and torsion, respectively, on an electrohydraulic materials testing machine. Bending stiffnesses were similar for the three nail configurations tested. The interlocked Grosse-Kempf nail was able to maintain the largest torque at fixed rotation, and exhibited larges springback, followed by the Brooker-Wills and then the non-interlocked Grosse-Kempf nails.

INTRODUCTION

Intramedullary nails have become a popular tool for fixation of long bone fractures. Optimal indications for this treatment are midshaft, noncomminuted, stable fractures. The newly developed interlocking nails have extended indications to include fractures which are located proximally or distally on the long bone shaft, or those which are comminuted.

The interlocking nail extends indications by providing positive fixation to the proximal and distal segments, allowing fractures distant to the narrow midshaft isthmus region to be stabilized. In addition, shortening can be controlled by the proximal and distal interlock, in cases of severe comminution. Interlocking intramedullary nails often allow earlier weight bearing and/or patient mobilization.

Two designs of interlocking nails are considered in this study, the Grosse-Kempf nail, (interlocked and non-interlocked) (1), and the Brooker-Wills nail.

The Grosse-Kempf nail is a slotted cloverleaf intramedullary nail with proximal and distal transfixion screws for proximal and distal cortical fixation. The non-interlocked Grosse-Kempf nail is similar to a standard Kuentscher type intramedullary nail. The Brooker-Wills nail is a slotted cloverleaf intramedullary nail with a proximal transfixion screw, and distal delta tri-fins, for proximal cortical fixation and distal cancellous fixation.

The objective of this study was to compare the stability of these nails in bending and torsion, with regards to the type of proximal and distal fixation (all cortical, cortical/cancellous, none).

MATERIALS AND METHODS

Eleven unpaired formalin fixed human femora were used for testing of the Grosse-Kempf interlocked and unaugmented nail, and twelve formalin fixed femora were used for testing of the Brooker-Wills nail.

The femora were measured for appropriate size of Grosse-Kempf and Brooker-Wills interlocking intramedullary nails. The femora were reamed as necessary. Transverse midshaft fractures were created and the nails inserted according to recommended practices. The following configurations were tested:

1) Grosse-Kempf
   - no transfixion screws
2) Grosse-Kempf
   - 1 proximal and 2 distal screws
3) Brooker-Wills
   - 1 proximal screw, and
   - deployed distal fins.

Experimental testing

The femora were tested on an MTS electrohydraulic materials testing machine in 4-point bending and torsion.

The bending moment was applied on the posterior-medial aspect of the femur with a constant moment arm of 6.2 cm. The femur was clamped at outer supports to prevent rotation during bending (Figure 1). A midshaft displacement of 0.32 cm was applied.

BENDING TEST

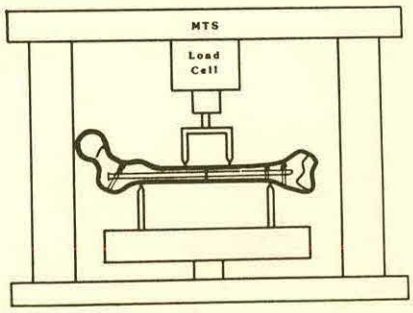

Figure 1

In torsional testing, the distal fragment was externally rotated 10 degrees, and maximum torque measured (Figure 2). This torque was released and the slippage between bone and nail was measured as the springback angle. Figure 3 graphically illustrates the springback angle, or angle between proximal and distal fragments with the torque reduced to the initial zero level.

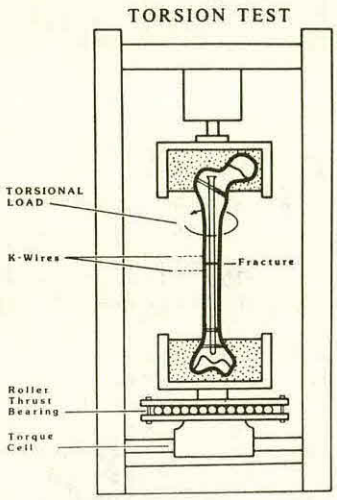

Figure 2

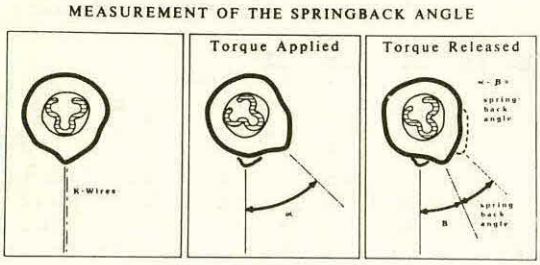

Figure 3

RESULTS

Table 1 lists mean values for 4-point bending stiffness (N/mm), maximum torque at a forced ten degree rotation (N-cm), and springback angle, or slippage between proximal and distal segments (degrees).

TABLE 1 - Mean Values

	4-pt.bending stiffness	Maximum Torque	Springback Angle
Grosse-Kempf (interlocked) (1)	259. N/mm	253. N-cm	8.4 deg
Brooker-Wills	207. N/mm	122. N-cm	6.1 deg
Grosse-Kempf (non-interlocked) (1)	232. N/mm	58.1 N-cm	.91 deg

Table 2 summarizes results of statistical analysis comparing the three nail configurations (Students t-test).

TABLE 2 - Statistical significance

	4-pt. bending stiffness	Maximum Torque	Springback Angle
B-W vs. G-K (interlocked)	NOT sig. T=1.4/p=.16	Sig. diff. T=5.2/p<.001	Sig. diff. T=2.7/p=.01
B-W vs. G-K (non-interlocked)	NOT sig. T=.89/p=.38	Sig. diff. T=2.9/p=.01	Sig. diff. T=6.0/p<.001
G-K (interlocked) vs. G-K (non-interlocked)	Sig. diff T=2.7/p=.02	Sig. diff. T=14./p<E-6	Sig. diff. T=12./p<E-6

DISCUSSION

4-pt. bending: The Brooker-Wills nail showed no significant difference in bending stiffness as compared to both the interlocked and non-interlocked Grosse-Kempf nails. The interlocked Grosse-Kempf nail was shown to have a slightly increased bending stiffness over the non-interlocked configuration (statistically significant).

Torsion: The interlocked Grosse-Kempf nail showed the highest maximum torque at 10 degrees rotation, followed by the Brooker-Wills nail and then the non-interlocked Grosse-Kempf nail (both statistically significant).

Springback angle: The interlocked Grosse-Kempf nail showed the largest springback angle, followed by the Brooker-Wills nail and then the non-interlocked Grosse-Kempf nail (both statistically significant).

The differences between the Brooker-Wills and Grosse-Kempf interlocking nails may be attributable to the fixation quality distally; with the Brooker-Wills having cancellous fixation, and Grosse-Kempf cortical fixation. Slightly higher values of torsional resistance may be anticipated in vivo with the Brooker-Wills nail since cancellous bone is weakened by formalin fixation.

As expected, both types of interlocking nails were more stable in 4-point bending and torsion, and had larger springback angles than the non-interlocked Grosse-Kempf nail.

CONCLUSION

Interlocking and non-interlocking intramedullary nails were tested to determine their stability in bending and torsion.

Bending stability was similar for the interlocked and non-interlocked Grosse-Kempf intramedullary nails, and the Brooker-Wills nail.

The interlocked Grosse-Kempf nail showed highest maximum torque, and largest springback angle, followed by the Brooker-Wills nail, and the non-interlocked Grosse-Kempf nail.

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ACKNOWLEDGEMENT

Howmedica and Biomet are thanked for generously supplying the intramedullary nails tested.

14.6

HISTOLOGIC EVALUATION OF TISSUE GROWTH INTO RETRIEVED NONCEMENTED KNEE PROSTHESES

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Numerous designs of orthopaedic prosthetic devices today incorporate various types of porous metallic coatings in an effort to enhance the fixation of these implants. Currently these porous coated devices are being implanted both with and without the use of polymethylmethacrylate bone cement. The use of porous devices without bone cement is an attractive alternative, since fixation is provided, in theory, by ingrowing bone spicules, rather than acrylic cement. It is hoped that such a method will provide long-term stable fixation. The evaluation of devices retrieved from patients can provide information on the efficacy of particular design considerations, and can guide in the development of more effective prostheses. This paper describes the histologic and microradiographic evaluation of tissue growth into 18 retrieved noncemented porous coated knee components removed from 10 patients.

INTRODUCTION AND BACKGROUND

There is considerable evidence from animal studies that bone growth and maintenance in porous coated implant surfaces is possible and can be expected if a number of factors are met. These include acceptable pore size, pore volume, percent porosity, as well as the lack of any initial implant micromovement [1-3]. This research has led to the development and sale of porous coated joint replacements for humans which can be implanted without the use of PMMA bone cement.

Clinical results after 5 years have been essentially equivalent for porous coated hip prostheses when compared to the simultaneous experience with cemented prostheses [4]. Similar results have also been reported for porous coated knee prostheses [5,6]. Little information is available on the histologic results from porous coated devices implanted in humans, partly because of the difficulty encountered in obtaining useful specimens.

Collier, et al. [7] evaluated 5 retrieved porous coated femoral stems and reported bone ingrowth in 2 cases although fibrous tissue was also present. In 2 other cases only fibrous tissue was found, and the final case was retrieved after only 10 days in situ and showed only fibrous tissue adherence but no ingrowth. Brooker and Collier [8] published a case report in 1984 of a femoral stem recovered from a 29 year old male during a revision surgery. Histologic sections showed both bone and fibrous tissue ingrowth after an implantation period of 4 months. At the Society for Biomaterials Meeting in 1984, Collier, et al. updated their retrieval study of the porous coated Moore femoral stem to include a total of 11 devices with both a fine (50-100 micron) and a larger (150-350 micron) pore size [9]. They reported bony ingrowth into all of the devices with the larger pore size coating, with accompanying fibrous tissue and the small pore size devices showed only fibrous tissue ingrowth. The most bony ingrowth was observed to occur on the lateral side of the stem, approximately 10 cm above the distal tip of the device.

Bobyn and Engh [10] reported histologic evaluations performed on 2 femoral stems, one from an 87 year old female after 40 days in situ, and one from an 82 year old male after 6½ months. These specimens exhibited some bony ingrowth, even though the patients were of an advanced age. As a follow up, Bobyn and Engh [11] updated their findings at the Society for Biomaterials in 1985 to include 2 additional femoral stem specimens. One was from a 75 year old male at 2 years after implantation, and the other from a 77 year old male at 3 years post-implantation. These longer term specimens showed considerable bony ingrowth in the medial cortex area, and the anterior and posterior introchanteric regions, and also medially and laterally distal stem areas.

Also at the Biomaterials Meeting in 1985, Bloebaum et al. [12] reported on one retrieved PCA knee and one porous coated femoral stem. The femoral stem showed mostly fibrous tissue with some spotty bony ingrowth. They reported bony ingrowth into some areas of the PCA knee components, particularly the pegs, but found fibrous tissue separating the tibial tray and femoral condyle surfaces from the underlying bone. To date this represents the only published report of a histologic evaluation of a retrieved porous coated knee prosthesis.

MATERIALS

The retrieved devices were components of the PCA Primary Knee System manufactured by Howmedica, Inc., and the Tricon-M Knee System manufactured by Richards Medical Company. All devices were inserted without the use of bone cement, and in no case was the retrieved component removed due to loosening.

Howmedica PCA Knee Components

The 11 PCA knee components included: a patellar component removed after 11 months due to adhesive capsulitis and inadequate knee flexion from a 64 year old female; a tibial component removed after 10 months due to subluxation from a 74 year old female; femoral, patellar and tibial components removed after 12 months due to pain from a 58 year old female; femoral and tibial components removed after 19 months due to a coagulase negative staph infection from a 73 year old male; a femoral component removed due to infection after 4 months from an 85 year old female; a patellar component removed from a 63 year old male after 24 months due to a traumatic fracture of the patella; and

patellar and tibial components removed from a 66 year old female after 20 months due to pain.

Richards Tricon Knee Components
The 7 Tricon knee components included: a femoral component removed from a 52 year old man after 2 months due to fibrous ankylosis; femoral, patellar and tibial components retrieved after 9 months from a deceased 68 year old female; and femoral, patellar and tibial components retrieved after 2 weeks for avascularity from a 69 year old female.

METHODS

The removed components along with any surrounding tissue were immediately placed in a 10% buffered formalin solution. Following fixation, the specimens were dehydrated in graduated ethyl alcohol solutions and embedded in methylmethacrylate. Undecalcified histologic and corresponding microradiographic sections were then produced with the implants in place using diamond cutting and grinding techniques [13,14]. Serial sections were cut with a high speed diamond saw and mounted onto plexiglas slides. The sections were first ground to a thickness of 100 microns using a precision swivel head grinder and contact microradiographs produced. The sections were further ground to a thickness of 50 microns and stained with toluidine blue and basic fuchsin. The sections were then examined in transmitted and polarized light for the degree and type of tissue ingrowth, evidence of osteoclastic activity and adaptive remodelling in the underlying bone, as well as an overall evaluation of the tissue-prosthesis interface.

RESULTS

ILLUSTRATIVE CASE REPORTS

Patient Number 1: PCA Patellar Component

C.R. was a 64 year old female who underwent a total knee replacement for osteoarthritis. Post-operative radiographs showed excellent bone cuts and device alignment, however pain persisted. At 11 months post-operative, she had a revision with a cemented PCA patellar component for adhesive capsulitis, inadequate knee flexion and pain. At surgery the patella component was judged to be firmly fixed, which confirmed the preoperative radiographic evaluation of no implant loosening.

Histologic sections from the device revealed a dense fibrous tissue layer separating the porous-coated implant from the underlying bone. No bone was observed either directly adjacent to or within the porous coating structure. Marked resorption of the underlying bone was also observed with numerous areas of osetoclastic activity. Although the fibrous tissue was in general acellular, in some areas the porous structure was infiltrated with macrophages and multinucleated foreign body giant cells.

Patient Number 2: PCA Femoral and Tibial Components

M.H. was a 73 year old male who had a noncemented PCA total knee implanted in March 1983 for degenerative joint disease. He had an uneventful post-operative course, however, in April 1984 he fell and sustained a fracture of the patella, and underwent a partial patellectomy. Subsequently he developed a late wound infection (coagulase negative staph), which required removal of the knee components in November 1984, 19 months after the original surgery.

Upon histologic examination, bony ingrowth was around the tips of the fixation pegs of the femoral and tibial components. On the tibial component mature, mineralized bony ingrowth was seen penetrating to the substrate on the lateral peg. The tibial tray and femoral condyle areas contained only fibrous tissue ingrowth. It is interesting to note that although this device was removed for a late infectious process, some areas of apparently healthy bone ingrowth were observed.

Patient Number 3: PCA Tibial Component

M.C. was a 79 year old female who had a total knee replacement for degenerative joint disease, and who was revised 9 months later for recurrent subluxation and instability although radiographically all were apparently well fixed. At revision surgery all components were solidly fixed, and the tibial component was removed with an osteotome. On gross examination the device was completely covered with bony tissue. The patient was revised with a thicker, noncemented PCA tibial component.

In all histologic sections, the porous coating was separated from the underlying bone by a fibrous tissue layer approximately 0.5 to 1.5 mm thick, in spite of the macroscopic and radiographic appearance of bone ingrowth. There were a few areas of bony adaptation to the surface, but in no area was a direct bone-implant interface observed. These results were suprising given the normal appearance of the underlying bone and since the device was firmly fixed, as evidenced by the difficulty in removing the component in surgery.

Patient Number 4: Tricon-M Femoral, Patellar and Tibial Components

M.E. was a 68 year old female who died 9 months after her total knee replacement from a subdural hemotoma. The patient had a history of degenerative joint disease and of breast tumor metastases to the knee, but the patient had no knee symptoms at the time of death.

Histologic examination of the femoral component revealed only isolated areas of bony adaptation to the large size porous coating, with the vast majority of the implant surface covered with a thick, acellular fibrous tissue varying in thickness up to 2 mm. Numerous areas of osteoclastic activity were observed in the underlying bone, indicating further resorption and increasing fibrous layer thickness. Interestingly some bony ingrowth was observed in the small pore size

material on the femoral component, which is designed only for fibrous tissue attachment. Examination of the patellar and tibial components also showed only fibrous tissue infiltrating the porous coating, with occasional areas of some bony adaptation to the porous coating, but no ingrowth.

DISCUSSION

The majority of the histologic sections revealed the presence a dense fibrous tissue layer separating the implant surface from the underlying bone. The fibrous tissue layer was in general less than 0.5 mm thick but in various locations ranged to approximately 1.5-2.5 mm thick. In many cases the fibrous tissue extended completely to the solid metal backing substrate through the porous coating. The fibrous layer was in general acellular; however tissue was found in a number of cases within the porous structure which contained macrophages and multinucleated foreign body giant cells. Although no component included in this study was removed due to loosening, the bone present at the fibrous tissue interface often showed marked resorption with numerous areas of osteoclastic activity.

Several of the longer-term PCA specimens showed bony ingrowth around the fixation pegs, even extending to the metal substrate. However, the femoral condyle and tibial tray areas of the same devices showed no bony ingrowth. In the Tricon devices, no bone ingrowth was found in the large pore size coating. The small pore size material on the anterior and posterior flanges of the Tricon femoral components are not intended to accept bone ingrowth, but in one case bone ingrowth was observed.

With time further bone resorption and an increase in fibrous layer thickness might be expected in a number of these cases. While the lack of extensive bone ingrowth did not contribute to the system failure in any of these cases, it is possible that progressive loosening may have occurred at a later date. While there is evidence that bone ingrowth will occur in porous coated knee components, the pattern of ingrowth and the results observed to date do not reveal a consistent pattern. Further histologic examination of retrieved specimens will hopefully elucidate the requirements for bone ingrowth and successful implant fixation.

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AN EVALUATION OF THE IN VIVO PERFORMANCE OF 250 INTERNAL FIXATION PLATES

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The *in vivo* performance of 250 retrieved internal fixation plates was evaluated by determining the corrosion characteristics and metallurgical properties of the implants and correlating these with clinical results. Screw-plate interface corrosion and screw surface corrosion were graded, and Rockwell hardness, grain size, thin inclusion content, and heavy inclusion content measurements were made. Upon examination, 89 % of all plates exhibited some degree of interface crevice corrosion, and 88 % of all screws exhibited some degree of surface corrosion. Statistical analysis revealed significant correlations between the corrosion gradings and metallurgical parameters studied. Stricter manufacturing standards for metallurgical properties could serve to enhance corrosion resistance and improve the *in vivo* performance of these devices. The routine removal of all fixation plates after fracture healing would reduce the occurrence of complications, such as breakage, loosening and implant related pain.

INTRODUCTION

The use of stainless steel plates for the internal fixation of fractures is widely accepted and advocated. Stainless steel's popularity is primarily because it can be easily shaped in the operating room by the surgeon. A major disadvantage of the material in comparison to other metals is its decreased strength and decreased fatigue resistance [1]. Reports of failure due to corrosion of stainless steel implants are also common [2-8]. Host response to metal implants has also been extensively studied. There have been reports of malignant tumors at the site of bone plating in canines, felines and man [7,9,10]. Localized inflammatory response and tissue necrosis have also been associated with stainless steel internal fixation plates [4,7,8,11]. It has recently been suggested that the type 316L stainless steel currently used for orthopaedic implants is not of sufficiently high metallurgical quality for its purpose, and that stricter manufacturing standards should be employed [1,3,5].

MATERIALS AND METHODS

To retrospectively assess the *in vivo* performance of stainless steel internal fixation plates, a group of 250 retrieved devices were clinically and metallurgically analyzed. The devices included 169 upper and lower extremity bone plates, 59 Richards type hip screw-plates and 22 Jewett type hip nail-plates, which were removed from patients between the years 1977 and 1985. All devices were recovered from Tulane University Medical School affiliated hospitals as part of an ongoing orthopaedic implant retrieval and analysis program.

All hardware inserted in patients at affiliated institutions is followed until its time of removal, at which time the actual implant is recovered. Data from this program revealed that from January of 1977 through August of 1985, approximately 1160 internal fixation plates were inserted, at a rate of approximately 129 per year. Thus, approximately one-third of the inserted implants have been removed. All implants were reported by the manufacturers to be fabricated from ASTM F138-76 Grade 2 (low carbon) stainless steel.

The clinical performance, corrosion characteristics and metallurgical properties of all devices were determined and analyses were performed according to ASTM Standard F561-78. Medical histories for each patient were obtained, including all clinical findings at both the times of implant insertion and removal. Whenever possible, one of the investigators was present at the retrieval of the implant.

Following surgical removal, each implant was ultrasonically cleaned in a 3% hydrogen peroxide bath and brushed with a mild detergent. Screws were kept in corresponding order with the respective screw-holes in order that screw and plate interface corrosion could be compared. All devices were photographed and examined stereomicroscopically for gross evidence of mechanical damage and corrosion. The screw-plate interface corrosion of each plate was graded on a scale of 0-5, where 0 represented no visible surface degradation at 60X magnification and 5 represented very severe surface degradation visible without magnification. The average score for a plate's individual screw holes served as the interface corrosion score for the entire plate. Similarly, individual screws were graded on the same scale and scores were averaged. The scores obtained were used in subsequent analysis.

Following microscopic corrosion evaluations, each plate was sectioned to isolate a 2 cm metal specimen. Individual specimens were mounted, polished to a mirror image and electrolytically etched for the determination of microstructural characteristics. Rockwell hardness, grain size, thin inclusion content, and heavy inclusion content measurements were made for each sample utilizing ASTM standard techniques. Clinical and metallurgical data were compiled and entered into a Decsystem 2060 digital computer and analyzed using BMDP Statistical Programs.

RESULTS

The average age at insertion for all patients with internal fixation devices was 39.3 years (range: 5 - 91 years). Those patients with bone plates averaged 33.9 years at insertion, with a range from 5 to 77 years. The mean age at insertion for

patients with Richards type screw-plates was 51.3 years (range: 20 - 80 years), while patients with Jewett type devices averaged 51.5 years at insertion, and ranged from 10 to 92 years.

The review of clinical data for all retrieved devices revealed that the majority of the implants were inserted due to trauma (48.4 %). Nonunion (2.4 %), malunion (2.0 %), congenital deformity (1.2 %) and dislocation and pathologic fracture (1.2 %) were the remaining representative diagnoses at insertion. The insertion diagnosis was unknown for 39.2 % of the devices examined. As might be expected, the vast majority of the hip plate devices (75.3%) were inserted for treatment of trauma.

The majority (50.4 %) of all devices were removed due to symptomatic causes. Only 28.0 % of all plates were removed on a routine asymptomatic basis, and 21.6 % of the devices were removed for unknown reasons. Implant related pain was the primary symptomatic removal reason, as observed in 34.1 % of the cases studied. This was followed by nonunion or malunion (21.4 %), infection (17.5 %), loosening (10.3%), breakage of the device (7.9 %) or other reasons such as avascular necrosis or refracture (7.9 %). Interestingly, of the Richards screw-plates only 9 were routinely removed (15.3 %), and of the Jewett nailplates only 1 device (4.5 %) was routinely removed.

The average time in situ for all plates examined was 26.3 months (range: 1 - 192 months). The bone plates averaged 27.0 months in situ (range: 1 - 192 months), the Richards screw-plates averaged 19.2 months (range: 1 - 96 months) and the Jewett nail-plates averaged 37.4 months in situ (range: 2 - 156 months).

Upon examination, 88.9 % of all implants exhibited some degree of interface crevice corrosion and 88.2 % exhibited screw surface corrosion. Statistical analysis performed on all devices revealed significant correlations between the corrosion variables studied and the metallurgical properties of the plates. Devices with inferior metallurgical properties exhibited significantly greater degrees of both interface crevice corrosion and screw surface corrosion. Devices with no corrosion (score = 0.0) were compared to devices considered highly corroded (score $\geq$ 2.0). Devices considered to have severe interface crevice corrosion had significantly greater screw surface corrosion ($P \leq 0.00005$) and significantly larger grains ($P \leq 0.003$) when compared to devices with no crevice corrosion. Similarly, devices with severe screw corrosion (score $\geq$ 2.0) exhibited significantly greater crevice corrosion ($P \leq 0.00005$), significantly higher thin inclusion content ($P \leq 0.068$), and significantly larger grains ($P \leq 0.025$) when compared to devices with no screw surface corrosion. A similar comparison was made of devices which had remained in situ 24 months or less and devices which were in place longer than 24 months. This showed significantly higher interface crevice corrosion scores ($P \leq 0.0004$) and screw surface corrosion scores ($P \leq 0.005$) for the devices which remained in situ longer.

Regression analysis also demonstrated significant correlations between grain size and corrosion scores, and other metallurgical properties. Both interface crevice corrosion and screw surface corrosion were found to increase with increasing grain size ($P \leq 0.005$). Note that an increase in grain size is reflected by a smaller grain size number, and that a decrease in grain size is represented by a larger grain size number. Thus corrosion was found to decrease with increasing grain size number. As expected interface crevice corrosion was highly correlated with screw surface corrosion ($P \leq 0.00003$). The thin inclusion content of the implant material was found to correlate with heavy inclusion content ($P \leq 0.00005$) as well as grain size ($P \leq 0.0001$). Further, the degree of interface crevice corrosion and screw sufrace corrosion were each found to increase significantly with time in situ ($P \leq 0.0003$).

Implant breakage was the reason for removal of 10 implants. The broken devices consisted of 6 bone plates, 2 Richards type screw-plates and 2 Jewett type nail-plates. When comparing the broken devices to the unbroken devices, there were no significant differences between the two for either corrosion scores or metallurgical characteristics. The broken devices were within ASTM specifications for metallurgical parameters with the exception of a single Jewett type device which was out of current standards for grain size number (4.5).

Metallurgical specifications for thin inclusions, heavy inclusions or grain size were not met in 25 of the 250 devices examined, and several devices were out of specifications for more than one of these parameters. For thin inclusion content 15 devices were out of specification (content $\leq$ 1.5) and an additional 17 devices had thin inclusion content of 1.5. Heavy inclusion content specifications (content $\leq$ 1.0) were not met by 12 devices, with another 30 devices having heavy inclusion content of 1.0. Grain size specification (grain size number $\geq$ 5.0) was not met by 3 devices, and 2 additional devices had grain sizes of 5.0.

DISCUSSION

Based on the results from this study there are clear indications that in many cases internal fixation devices are considered permanent implants. This is in spite of recommendations that these devices be removed after fracture healing has occurred, unless there are specific contraindications. Routine asymptomatic removals represented only 28% of the removals (70 devices), and of these 14 were removed after 24 months. Further, it was also found that reason for removal was independent of time in situ.

The tendency to use stainless steel fixation devices as permanent implants becomes more important when one considers the corrosion characteristics of these devices. Stainless steel is the material of choice not because of its corrosion resistance, but because the devices can be custom shaped in the operating room. The degree of corrosion observed in this study is

consistent with previously published reports [1,12-14]. Of the 250 devices studied 89% exhibited some degree of corrosion and both screw surface corrosion and interface crevice corrosion were observed to increase with time in situ.

Corrosion scores were found to correlate with the metallurgical characteristics of the implant material. Screw surface corrosion and interface crevice corrosion were each found to correlate with grain size. In each case larger material grain size correlated with increasing degree of corrosion. Further, grain size and thin inclusion content were similarly correlated; large grain size material was found to have more thin inclusions, also increasing the tendency for corrosion. It was also found that plates with greater thin inclusion content also had higher heavy inclusion content as well. Non-metallic inclusions are forms of material inhomogeneity and are documented causes for corrosion and corrosion fatigue [13].

The incidence of complications from the use of internal fixation devices may be reduced if retrieval is performed routinely whenever possible, and removing the devices earlier would also tend to reduce the amount of corrosion present. This may not be true, however in devices fabricated from metallurgically inferior material. Current ASTM standard F138-76 was not met by 25 (10%) of the devices studied. Thin inclusion and heavy inclusion content specifications were not met by 23 of the 25 devices; 2 devices were out of specification for grain size. The number of cause related removals at all time periods, and the extensive corrosion of the devices both indicate that stricter standards for manufacturing seem warranted.

From the clinical and metallurgical data examined in this study, the use of stainless steel fixation devices as permanent implants is not justified. The authors data suggest that stricter manufacturing standards would improve the clinical performance by reducing the incidence of corrosion of these devices. Furthermore stricter standards would insure that these implants are manufactured from metallurgically cleaner material and thereby reduce corrosion present at all time periods in situ. Finally, the routine removal of all internal fixation devices after fracture healing has occurred is suggested whenever possible.

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AUTOMATIC RANGE OF MOTION MEASUREMENTS FROM VIDEO IMAGES

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ABSTRACT

Obtaining accurate three dimensional angle information is important in the investigation of the range of motion in the cervical spine (neck). An accurate, quick, inexpensive method, coupling a video camera and system of light emitting diodes (LEDs), has been developed. The method allows three dimensional angles from the two dimensional image obtained by the video camera. A computer program which simulates the errors in the system was used to help determine the optimum LED fixture design and camera placement. Correction factors were calculated and when applied to the two dimensional data gave a maximum error of less than two degrees when compared to the actual angle.

INTRODUCTION

Acquiring the three rotation angles (roll, pitch, yaw) of a rigid body, using a fast video digitizing system will improve data acquisition techniques for investigation of cervical range of motion. Obtaining three dimensional information strictly from a two dimensional projection is impossible, however by constructing a geometric model of the system the errors can be predicted. A correction factor, composed of the predicted errors and small non-predicted errors, was obtained and when applied to the two dimensional data gave a maximum error of less than two degrees.

METHOD

The technique of obtaining the angles uses two bars which have light emitting diodes (LEDs) spaced three inches apart. These two bars are mounted orthogonally. One bar is pointed directly at the camera, such that its two LEDs appear as one dot on the screen. This bar bisects the second bar and thus guarantees that the second bar is parallel to the camera's image plane. This initial position is used to obtain the length of the bar in screen coordinates. As the fixture rotates the LEDs which originally appeared as one dot on the screen will begin to separate. The ratio of the length between these two LEDs and the calibration length were used in the following equation to obtain the angle of rotation of the fixture.

$$\psi' = \text{ARC COS} (L1/L0) \qquad \text{(eq. 1)}$$

where L1 = length between two LEDs when fixture is at angle

L0 = calibration length between two LEDs

This equation assumes that the fixture is constrained so that only rotation about one of the LEDs is allowed and the fixture is in line with the center of the camera lens.

Since a head is free to move in space and cannot be accurately evaluated using this assumption, a computer program was constructed to simulate the system and evaluate the errors due to any of several variables encountered in obtaining three dimensional information from two dimensional images. The variables included in the error evaluation were distance from the camera, length between LEDs, distance of the LEDs from the center of the image, and the other two angles not being measured. The program was then used to evaluate and obtain an optimum LED fixture design and the optimum camera position.

A rigid body moving freely in three dimensional space requires that all the variables be allowed to change between positions. The two dimensional coordinates, for the projection of bar "L", in the XZ- plane (Figure 1) are given in the following equations:

X coordinate of projection point LED1 is

$$x = (D*X) / (D - Y) \qquad \text{(eq. 2)}$$

Z coordinate of projection point LED1 is

$$z = (D*Z) / (D - Y) \qquad \text{(eq. 3)}$$

where the positions are defined as:

D = Camera distance from fixture
X = Horizontal offset from the image center
Y = Change in distance toward the camera
Z = Vertical offset from the image center
L = The actual length between LEDs
ϕ = Pitch of the fixture (degrees)
ψ = Yaw of the fixture (degrees)

If LED1 is then located at position (X,Y,Z)

Then LED2 is then located at position

$$X' = X + L*COS(\phi)*SIN(\psi)$$
$$Y' = Y + L*COS(\phi)*COS(\psi)$$
$$Z' = Z + L*SIN(\phi)$$

The coordinates for the projection point of LED2 is obtained with equations similar to equations 2 and 3.

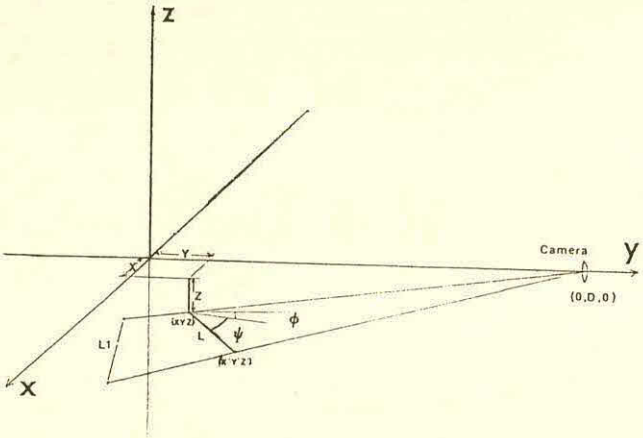

Figure 1. Two dimensional projection of bar "L".

The distance between the two projection points is calculated and this distance is one of the distances used in the ratio which determines the angle ψ'. The angular error was then calculated as the difference between the actual angle ψ and the approximation obtained from equation 1.

The effect of varying camera distance from the fixture was evaluated over a range of 103 to 108 inches. The distance of the LED image from the center of the screen was varied over a five inch range and the errors were calculated. The length between LEDs was varied from 2 to 7 inches and the effect of this change was evaluated. The pitch angle and roll angle were varied over a range of 0 to 45 degrees to examine the errors introduced.

At small angles the angular error was found to be more sensitive to changes in distance from the image center than changes in distance from the camera. Therefore it was desirable to keep the two LEDs, being monitored, as close to the screen's center as possible. The length between LEDs effects the error, however the optimum length is governed more by the limits of the computer monitor as the fixture travels through the ranges of motion. The pitch and roll angles were found to increase the error but if kept less than 5 degrees the error could be constrained to less than a degree. At larger angles (angles greater than 45 degrees), alternative fixture designs, incorporating more LEDs could be used to minimize the unavoidable errors.

The LED fixture previously described was mounted on a goniometer. A video camera linked to an Apple IIe micro computer was used to obtain the two dimensional image of the LED fixture as the goniometer was rotated through a range of 0 to 45 degrees in increments of 5 degrees (Figure 2). The calculated angle can be represented by the following equation:

$$\psi' = \psi + \epsilon(\psi) \qquad \text{(eq. 4)}$$

where ψ' is the calculated angle from equation 1

ψ is the actual fixture angle

$\epsilon(\psi)$ is the angle error including predicted and non-predicted error

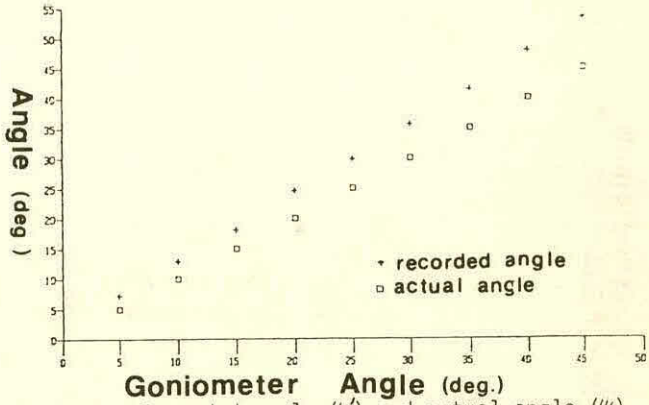

Figure 2. Recorded angle (ψ') and actual angle (ψ) vs. goniometer angle.

The predicted error, from the computer simulation, was then plotted against the actual angle and a linear function was fit to the data to introduce a correction factor. Figure 3 shows the results after correction with the following equation from predicted errors (maximum error 5.0 degrees).

$$\alpha = (\psi' - .342) / 1.063 \qquad \text{(eq. 5)}$$

where α is the predicted angle which includes correction for predicted errors

ψ' is the calculated angle from equation 1

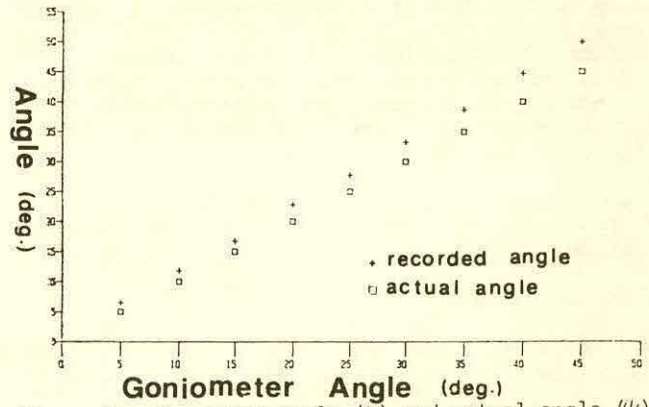

Figure 3. Corrected angle (α) and actual angle (ψ) vs. goniometer angle.

It was observed that non-predicted errors still remained in the angle calculation with correction. The actual error was then plotted against the actual angle and a linear function was fit to the data to introduce a new correction

factor. Figure 4 shows the results after correction with the following equation utilizing the new correction factor (maximum error 1.0 degrees).

$$\alpha' = (\psi' - 1.169) / 1.158$$

where α' is the predicted angle which includes correction for non-predicted errors

ψ' is the calculated angle from equation 1

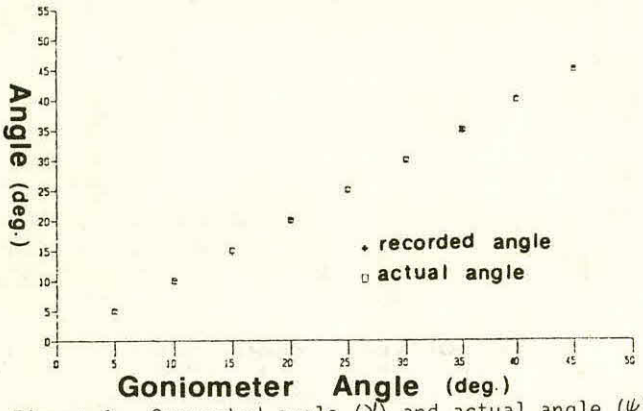

Figure 4. Corrected angle (α') and actual angle (ψ) vs. goniometer angle.

SUMMARY

The errors encountered when using two dimensional images to obtain three dimensional angle information have been investigated. Investigation of the error sensitivity to changes in different environmental parameters gave valuable information for the design of a system with minimum error. Minimizing the distance that the LEDs are from the image center and making angle measurements from LEDs that are oriented such that their bar is pointed directly at the camera lens will give minimum errors.

After optimization of the design and environmental parameters, the actual angle was obtained (with good correlation) using the calculated angle and the correction factor obtained for the actual errors.

This system of hardware and software will be used in a range of motion package for the cervical spine.

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THE ACCURACY OF A QUICK MULTIPLE MUSCLE GROUP TESTING PROTOCOL

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ABSTRACT

A quick test (QT) protocol has been developed to allow the rapid testing of a number of muscle groups in order to profile body strength. The maximum muscle strength using a standard test (ST) protocol was obtained from the best of three five second trials with a 1 minute rest between trials and no more than 4 muscles tested per day. The QT protocol allows only a 5 second rest between trials. Using the QT protocol and testing 13 muscle groups in one session vs. 5 sessions with the ST protocol resulted in an average reduction for the QT values of 4%.

INTRODUCTION

For many years the need and use of quantitative muscle strength testing has been reported. Despite this, routine use of quantitative muscle testing does not occur in rehabilitation medicine. This is in part due to the lack of appropriate and easy to use equipment and the time required to perform the testing. This report describes the results of research performed to reduce the testing time on a large group of different muscles. Our long range goal is to develop a clinically useful muscle testing system which can accurately test 15-20 muscle groups in less than one hour.

A commonly used protocol for maximal isometric muscle testing requires the subject to perform three 5 second contractions with a 1 minute rest between each contraction with the largest force value considered the maximum strength (6, 5, 8, 12). Studies have used rest periods from 30 seconds to 4 minutes (3, 10, 2). In these studies no more than 3 or 4 muscle groups are tested in one session. If many muscle groups are tested with this type of protocol, it would require a significant amount of time.

The purpose of this study is to compare the maximum muscle strength values obtained testing 13 muscle groups in one session with a protocol requiring 5 sessions.

METHODS

Using the standard test (ST) method the subject contracts maximally for a duration of 5 seconds with a timed 1 minute rest between maximal efforts. Three contractions at voluntary maximum levels are performed and the best of the three is considered the isometric maximum strength for that muscle group. The quick test (QT) method varies from the standard method in that there is only a 5 second rest between maximal efforts and all the muscle groups were measured in one session with a typical duration of 1 hour and 30 minutes for the complete session.

Prior to participation in this investigation each subject signed an informed consent in accordance with the guidelines established by the Committee for the Use of Human Subjects in Research at the University of Minnesota. Eleven normal college-age female physical therapy students participated in the study. The average height, weight and age of the group is given in Table 1. Prior to participating, all students underwent a medical history evaluation to determine any pathology contra-indicating performing strength testing.

Table 1.

Description of subjects: Accuracy of Quick Test Protocol.

N = 11 women

	Mean	±	Standard Deviation
1. Height (inches)	65.3		1.7
2. Weight (pounds)	130.6		12.7
3. Age (years)	23.5		3.6

Ten of the women were right hand dominant and one of the women was left hand dominant.

For the ST protocol the subjects visited the laboratory on five different occasions. The movements tested on each of the five visits were grouped as follows:

1. Ankle plantar flexion.
2. Shoulder flexion, shoulder abduction, and shoulder external rotation.
3. Shoulder extension, shoulder adduction, shoulder internal rotation and elbow extension.
4. Handgrip.
5. Hip flexion.

The order of the sessions was randomized for each subject by drawing the session number from a hat. The order that the strength tests were performed within each test session was standardized. Only the dominant side was tested using the ST method.

Strength testing using the QT protocol was performed using the same body positions and stabilization as the ST method. However, both dominant and nondominant sides were tested using the QT method whereas only the dominant side was measured using the ST method. This resulted in 19 tests being administered in one session for the QT but only 13 comparisons between the ST and QT protocol. The order of the muscles tested is the same as the order the results are presented in Table 2. Hettinger (4) reported that a training simulus due to a maximum isometric contraction held for one to two seconds would be completely dissipated in 14 days. Therefore, the QT was performed longer than 14 days after the ST.

Muscle Groups Tested	ST Best of 3 lbs. ± SD (% bw ± SD)		QT Best of 3 lbs. ± SD (% bw ± SD)		QT as % Δ from ST max
Rt. Hip Flexion	45.5 (34.3%	6.4 4.0)	45.2 (34.6%	6.2 4.0)	-.55
Lt. Hip Flexion	46.6 (35.4%	6.0 4.3)	46.2 (35.3%	5.8 3.7)	-.97
Rt. Plantarflexion	312.6 (239.3%	59.7 37.4)	297.9 (229.6%	57.4 40.7)	-4.69
Lt. Plantarflexion	315.0 (241.4%	59.3 41.6)	296.7 (228.5%	49.6 35.6)	-5.81
Rt. Handgrip	80.5 (61.2%	10.5 6.7)	81.4 (62.5%	12.3 7.2)	+1.14
Lt. Handgrip	78.9 (60.0%	8.1 3.9)	77.2 (59.3%	10.9 5.8)	-2.53
Shoulder Extension (dom)	52.9 (39.9%	9.4 6.4)	50.4 (38.6%	8.4 5.3)	-4.62
Shoulder Adduction (dom)	61.3 (45.9%	9.7 4.5)	58.0 (44.5%	9.1 6.0)	-5.37
Shoulder Internal Rot. (dom)	29.8 (22.4%	5.7 4.1)	29.0 (22.1%	6.4 4.5)	-2.88
Elbow Extension (dom)	28.0 (20.9%	5.9 3.7)	26.6 (20.2%	6.0 4.1)	-5.13
Shoulder Flexion (dom)	38.7 (29.2%	3.7 2.7)	34.8 (26.6%	4.4 3.1)	-10.27
Shoulder Abduction (dom)	63.7 (30.6%	8.0 4.6)	35.9 (27.2%	7.0 5.0)	-11.46
Shoulder External Rot. (dom)	26.3 (19.7%	4.4 3.1)	26.5 (20.3%	3.8 2.3)	+.76

Table 2. Strength testing results from Standard Test protocol and Quick Test Protocol.

For these tests all joints were positioned at 90° except the hand which followed Mundale's (8) method. Following familiarization with the test procedure consisting of three submaximal contractions, three tests of maximal volitional contraction each lasting 5 seconds were performed with a timed 1 minute rest between each test.

The subjects were asked to make a rapid maximal contraction with strong verbal encouragement given during the duration of the contraction. The best of the 3 contractions was considered the maximal isometric strength for the muscle group tested.

The force measurements were made using load cells in a computerized measurement system previously described by Patterson et al. (9).

RESULTS

The results of the maximal strength testing for both ST and QT protocols are given in Table 2. The data are presented giving the maximum force from the best of three contractions. The QT values were on the average 4% lower than the standard test. Using a paired T-test only 2 muscle groups showed a statistically significant difference between the QT and the ST protocols at $p < .05$. The correlation of the difference between the ST and QT values with the sequence of the test in QT protocol was $r = -.423$ and not significantly different from 0 at $p < .05$. The coefficient of variation ([SD/mean] x 100%) of the average force for the ST and QT protocols were 16.0% and 16.5% respectively. Normalizing the forces by body weight results in a coefficient of variation for the ST and QT protocols of 13.5% and 14.3% respectively.

The percent reduction in average maximal force comparing the best of one, two, or three repetitions for both ST and QT protocols is shown in Figure 1.

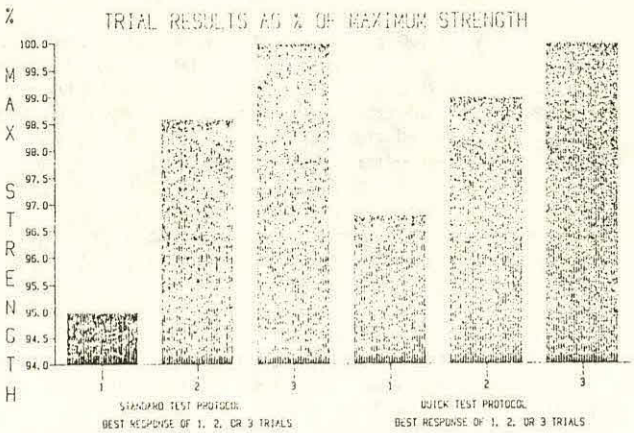

Figure 1. Trial results of the three repetitions presented as a percentage of the maximal strength.

DISCUSSION

The decrease of 4% in the quick test results are small compared to the average coefficient of variation (CV) in the force of 16.5%. This is slightly less than the average of 18% reported by Gunnar (13) when he tested 22 muscle groups in 19 to 20 year old military conscripts. Normalizing the force values by body weight reduces the coefficient of variation from approximately 16% to 14% for both protocols, which is similar to

previously reported values of 12 to 14% by Beasley (1). The variability of the quick test force measurement is comparable to the standard tests, therefore, the use of the quick test protocol may not significantly reduce the accuracy of repeated measurements when following a patient's progress even though they may be slightly lower in value. This would be important for following a given patient. Wiles (12) has reported a coefficient of variation on repeated measurements of strength from 4.5% to 14% in normals. Using the best of two repeated contractions instead of three results in less than 2% error for either protocol suggests that little would be gained using three repetitions.

The switching of the setups associated with using stabilization equipment similar to the Beasley methodology (1) limited the speed of performing the testing to 1 hour and 30 minutes. Currently under development is a testing chair that will allow rapid changing of the stabilization of the different muscle groups and thereby reduce the testing time, it is estimated, to less than one hour.

The results of this study in normals would suggest that the QT protocol could be used with only 2 repetitions to test multiple muscle groups without serious error. This would make it practical to make measurements to profile the muscle strength throughout the body in a given individual in one measurement session.

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15.3

ULTRASOUND MEASUREMENT OF TRUNK MUSCLE ATROPHY PRODUCED BY LUMBAR SPINE ORTHOSES

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ABSTRACT

In order to determine whether extended use of lumbar spine orthoses causes trunk muscle atrophy, nine healthy, adult, male volunteers were instructed to wear a lumbar corset continuously during waking hours for a period of three weeks. Compliance was measured using a custom-fabricated temperature-activated sensing device. Ultrasonic scans were taken prior to brace wear, immediately after removal of the brace, and three weeks later as a follow-up. Transverse scans of both extensor and flexor groups were taken at the L3 level. Sagittal scans of the extensors were taken 3.0 cm from the midline. Measurements were recorded for the length, width and area of the rectus abdominis; width of the oblique group; and depth of the erector spinae muscles at the L1 through L4 levels. In general, no significant change was found in these muscle dimensions between any of the time points. This suggests that atrophy of the trunk muscles during corset wear is not a significant problem.

INTRODUCTION

Lumbar corsets are widely prescribed to treat low back pain, although it is commonly believed that corset wear may cause muscle atrophy. However, no direct test of this hypothesis has ever been performed. This is partly because no reliable, noninvasive technique was previously available to assess changes in muscle dimensions in vivo. For example, studies have shown that circumference measurements of the thigh are poor indicators of changes in quadriceps muscle bulk, even if allowances are made for skinfold thickness (1,2,3). Muscle biopsies, in addition to being invasive, have limited usefulness because major assumptions must be made concerning the number of fibers in the muscle and the extent to which biopsies of tiny areas can indicate changes in whole muscles (4). EMG's are safe, but are subject to a variety of errors. CT scans can provide superior resolution of muscles, but have serious limitations because of radiation exposure (5). In contrast, ultrasound offers a safe, noninvasive method of taking reliable measurements of trunk muscle dimensions.

METHODS

Nine adult male volunteers (ages 18-29) with no history of low back disorders wore a Camp International Inc. Lumbosacral Support for a period of about three weeks. The subjects were instructed to wear the brace constantly during waking hours. Diaries kept by the subjects indicate that they wore their braces an average of 12 hours a day (±1.7 hours). In addition, compliance was monitored using a body-temperature activated timing device fitted into the corset. Muscle dimensions were recorded using a Picker #80LDI ultrasonic scanner with a 5 MHz transducer on short internal focus (focal zone of 3-5 cm). Images were taken of all subjects at three different time points: immediately prior to brace wear, immediately after brace removal, and three weeks later as a follow-up.

Transverse scans of the trunk extensor and flexor groups were performed at the level of the third lumbar vertebra. Sagittal scans were taken of the extensor muscles at various distances from the midline (3.0 cm from the midline provided the best delineation). These scans were then processed in three steps. First, significant anatomical features were manually traced from the transverse scans of the trunk flexors and the sagittal scans of the extensors. Second, a coordinate system was established on the transverse scan of the flexors to allow explicit definition of the muscle dimensions. Third, various muscle dimensions were determined using a manually controlled x-y digitizer for the areas and a millimeter ruler for the lengths and widths.

The following dimensions were used: length, width, and area for the rectus abdominis (RA); width of the oblique group (OG = external oblique, internal oblique, and transversus abdominis combined); and width of the erector spinae (ES) on the sagittal scans at the L1-L4 levels. Here, length is defined as the maximum linear dimension of the rectus on the cross-sectional slice. The rectus widths were measured along the perpendicular bisector of the lengths. These dimensions were chosen on the basis of a study to determine measurement reproducibility of ultrasound (6). The predictions of relative error (shown in the last column of tables 1 and 2) were made according to the following criteria: 10 subjects, 2 time points, 1 8x10 photo-enlargement per scan, and 1 measurement of a particular muscle dimension per photo-enlargement, with the difference detection threshold chosen at the 95% confidence level. All anthropometric data are scaled to life size.

RESULTS

Table 1 shows the mean and standard deviation for each measured muscle dimension at time 1 (pre-brace) and time 2 (post-brace). The percent change between the two time points, significance of difference, and relative error of measurement is also given. Table 2 presents the data for time 2 (post-brace) and time 3 (follow-up). Changes in muscle dimensions over time were evaluated using a matched pairs t-test.

CONCLUSIONS

1) In general, no significant atrophy occurred in the abdominal or erector spinae muscles during 3 weeks of lumbar corset wear or in 3 follow-up weeks in 9 healthy individuals.

2) This implies dimensional changes of no more than 9 to 23%, the reproducibility error of the technique for the various muscle measurements used (6).

3) Extension of this study to low back pain patients is presently under investigation, since extrapolation of results from this normal population is not necessarily valid.

Table 1. Muscle Size For Time 1 and 2 (cm or cm^2)

DIMENSION	TIME 1 MEAN	TIME 1 S.D.	TIME 2 MEAN	TIME 2 S.D.	PERCENT CHANGE	SIGNIF. OF DIFFERENCE	RELATIVE ERROR OF MEASURE.**
RA Length	7.1	±0.8	6.8	±0.9	-4.4%	N.S.	12%
RA Width	1.3	±0.4	1.4	±0.4	7.1%	N.S.	23%
RA Area	7.2	±2.4	7.0	±1.6	-2.9%	N.S.	23%
OG Width	2.9	±0.5	2.8	±0.4	-3.4%	N.S.	9%
L1 ES Depth	5.3	±0.9	5.3	±0.7	0.0%	N.S.	10%
L2 ES Depth	5.4	±1.0	5.4	±0.7	0.0%	N.S.	9%
L3 ES Depth	5.7	±0.9	5.6	±0.7	-1.8%	N.S.	13%
L4 ES Depth	5.8	±1.1	5.7	±0.7	-1.7%	N.S.	13%

Table 2. Muscle Size For Time 2 and 3 (cm or cm^2)

DIMENSION	TIME 2 MEAN	TIME 2 S.D.	TIME 3 MEAN	TIME 3 S.D.	PERCENT CHANGE	SIGNIF. OF DIFFERENCE	RELATIVE ERROR OF MEASURE.**
RA Length	6.8	±0.9	7.1	±1.1	4.2%	N.S.	12%
RA Width	1.4	±0.4	1.1	±0.4	-27.3%	0.05	23%
RA Area	7.0	±1.6	6.6	±2.0	-5.7%	N.S.	23%
OG Width	2.8	±0.4	2.9	±0.3	3.6%	N.S.	9%
L1 ES Depth	5.3	±0.7	5.0	±0.4	-5.7%	N.S.	10%
L2 ES Depth	5.4	±0.7	5.4	±0.6	0.0%	N.S.	9%
L3 ES Depth	5.6	±0.7	5.5	±0.7	-1.8%	N.S.	13%
L4 ES Depth	5.7	±0.7	5.7	±1.0	0.0%	N.S.	13%

** established in preliminary study (3).

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DESIGN AND ANALYSIS OF A THREE-AXIS ELECTRICAL GONIOMETER TO MEASURE TRUNK RANGE OF MOTION IN THE WORKSITE

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ABSTRACT

A goniometer has been designed which continuously measures trunk range of motion in the worksite environment. The device is harnessed to a worker who remains unconstrained while performing tasks. The harnessing consists of two plates which attach posteriorly to the sacrum and mid-scapular region via adjustable velcro straps. Attached to the sacral plate are three electrical potentiometers that align with the physiologic axes of flexion/extension, lateral bend and axial rotation. The potentiometers' outputs are stored in real time on a portable data recorder for digital processing in the laboratory. To evaluate the device, trunk range of motion for six males was compared to measurements from a clinical hand-held goniometer and from planar photography. A linear correlation of the goniometer to the other two techniques yielded r^2 values of at least .95 for all rotational axes of motion.

INTRODUCTION

The posture a worker assumes while performing work tasks may contribute significantly to his/her exposure to low back pain. Trunk range of motion is an important component of postures that may prove detrimental over a period of time. Instruments to evaluate trunk range of motion include the inclinometer (1), hand-held plastic goniometer, flexible rules, x-rays and three-dimensional spatial digitizers (4,5). Inclinometers, hand-held goniometers, flexible rules and x-rays require hand recording data and are only good for spot measurements. Therefore, these methods are only useful for gross, static flexion-extension analysis and are thus impratical means of monitoring trunk motions in the workplace.

Three-dimensional spatial digitizers are perhaps the best means of providing 3-D dynamic information and are gaining use in the worksite environment. However, they are either constrained to a small envelope of space or rely on a consistent field of view to monitor a worker's range of motion.

Because of this, few studies have been conducted on actual or required trunk range of motion for workers in a worksite environment. Nordin (2) measured continuous trunk flexion/extension movement of several workers with a harnessed pendulum goniometer. Similar work has been reported by O'Brien and Paradise (3) who used back surface strain gauges. We have designed and built an instrument that can monitor trunk motion in the work place. Comparative measurements with a clinical hand-held goniometer and with planar photography were made.

METHODS

Device Description

The goniometer is an electromechanical device which is harnessed to a subject via two plates. One is situated on the mid-scapular region and centered on the thoracic spine and the other centered on the sacrum. The plates are joined by a mechanical linkage which allows continuous movement between the sacral and mid-scapular plates. This makes the system nonconstraining for the worker to wear. The sacral plate houses three rotational potentiometers, the axes of which are aligned with three physiologic planes: (a) sagittal (flexion-extension); (b) coronal (lateral bending); (c) transverse (for axial rotation). In addition, the sacral plate houses a voltage supply (9-volt transistor battery) and a voltage regulator which provides an excitation voltage of one volt to each potentiometer. The voltage-divided signals of each potentiometer (linearly proportional to angular trunk movements) are interfaced with a portable FM tape recorder, also strapped to the worker, providing continuous time-based recording.

Method of Analysis

The goniometer was compared with two established techniques for measuring trunk range of motion: (a) a hand-held transparent plastic goniometer used by clinicians, and (b) the tracking of anatomical reference points by the use of planar photography. The three techniques are referred to as "goniometer," "clinical," and "photo" in the following discussion. Six male subjects, ranging in height from 170 to 193 cm, in weight from 55 to 91 kg, and in age from 20 to 32 years, were tested in flexion, lateral bend and rotation using simultaneous measurements from all three methods. The goniometer was strapped to the subject, after which the subject stood in a harness which constrained the lower body but allowed free movement of the upper body including rotation of the pelvis.

For flexion, skin markers were placed on the acromium (AC), the ASIS, and the greater trochanter (GT). Since the pelvis was free to rotate, the sacral plate of the goniometer also rotated in flexion. This rotation was subtracted from the photographically-determined flexion angles (AC-ASIS line) by monitoring markers placed on the sacral plate itself. This procedure emphasizes that the goniometer measures motion of the upper body relative to the sacrum and not to an absolute reference such as the vertical line. For lateral bend, skin markers were placed on the pubic symphysis and on the mid-sternum. For axial rotation, markers were placed directly on the goniometer to measure the motion of the mid-scapular plate relative to the sacral plate.

For the photographic technique, a ceiling-mounted camera was used to measure rotation and a tripod-mounted camera to measure flexion and lateral bend. For the goniometer, the analog voltages from the three potentiometers were clocked into an analog-to-digital (A/D) converter controlled by a microcomputer. Whenever a "reading" was taken, the cameras and the computer were hand-triggered simultaneously. The digitized voltage readings were processed in software and converted to angular displacements. The user was also prompted by the computer to enter the reading of the clinical hand-held goniometer, as part of a data file representing an individual subject performing one set of angular movements.

For each set of angular movements, the test protocol was as follows. First, the subject was instructed to stand comfortably erect and a reading representing zero angular position for all three measurement techniques was taken. Next, the subject was instructed to move to a new angular position by a researcher who visually aligned the axes of the hand-held clinical goniometer with the markers described for the photographic measurement technique. This process was iterative. Once the correct angular position was assumed, a reading was taken. Although this required the subject to stay motionless for a few seconds, repeatability for this process from previous tests was estimated as ±2 degrees. Using 10-degree increments, readings were taken in this manner to generate data sets of -20 to 60 degrees of flexion (extension is negative), -30 to 30 degrees of lateral bend, and -30 to 30 degrees of axial rotation. In some cases, a subject could not reach an extreme angular position and the researcher adjusted the clinical goniometer to record the actual position attained.

RESULTS AND DISCUSSION

For each subject, good agreement was found between the goniometer and the other two measurement techniques. We therefore combined data from the six subjects for each of the three axes of rotation (54 points for flexion; 42 points for lateral bend and axial rotation). Table 1 shows separate linear regressions of the clinical and photographic techniques (X) versus the goniometer (Y). Coefficients of determination (r^2) of at least .95 were obtained in all cases. The regression slope varied by as much as 22% (.82 to 1.22) from the ideal value of 1.00 which would imply equivalence between two measurement techniques.

The photographic (X) versus goniometric (Y) data and their linear regressions (solid line) for flexion, lateral bend and axial rotation are plotted in Figures 1a, 1b and 1c respectively. The relatively more scattered plot for lateral bend, versus flexion and rotation, is reflected in its lower r^2 value (.95 versus .98 and .99).

The standard deviation of the differences between a goniometer reading and the corresponding photographic/clinical reading is presented in Table 2. This standard deviation is a matched-pair statistic, representing the average discrepancy between measurement techniques.

The goniometric versus clinical readings for flexion produced the largest standard deviations (6.3 degrees). This occurs partly because the clinical measurement does not include pelvic rotation which is accounted for in the goniometer's movable sacral plate. This also explains the shallow regression slope (.82) in Table 1.

In addition to experimental error, the three measurement techniques may not agree because each relies on the angular shift of a line connecting only two points on the body. If these points are not the same for two techniques, a relative rotation between the points introduces a discrepancy. For example, we believe that the convexity of the flexion curve in Figure 1a stems from an added thoracic kyphosis at large flexion angles. This implies a forward rotation of the acromium (photographic point) relative to the mid-scapular region (goniometer point). Therefore, photographic measurements are higher than goniometric measurements at large angles of flexion. The more fundamental problem is that only two points are used to measure the motion of the upper body which is made up of multiple flexible linkages. In this sense, no single measurement technique is superior.

Table 1. REGRESSION ANALYSIS OF GONIOMETER

		FLEXION (N=54)	LAT BEND (N=42)	ROTATION (N=42)
GONIOMETER (Y) vs. PHOTO (X)	R-SQ.	0.98	0.95	0.99
	SLOPE	0.92	0.82	1.22
GONIOMETER (Y) vs. CLINICAL (X)	R-SQ.	0.96	0.97	0.99
	SLOPE	0.82	0.83	1.13

Table 2. STANDARD DEVIATION OF MATCHED PAIRS

MATCHED PAIR (degrees)	FLEXION (N=54)	LAT BEND (N=42)	ROTATION (N=42)
GONIOMETER - PHOTO	3.8	5.2	4.5
GONIOMETER - CLINICAL	6.3	4.5	3.4

Figure 1a FLEXION

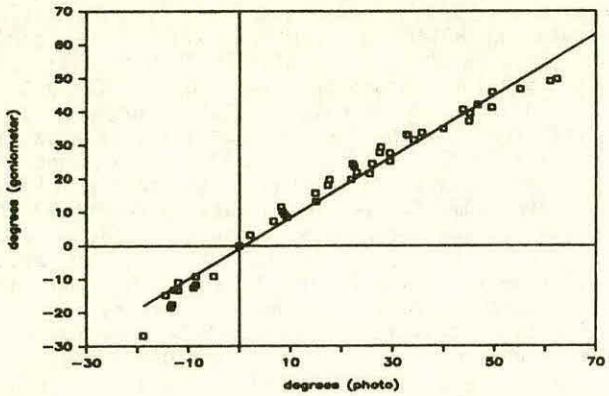


Figure 1b LATERAL BEND

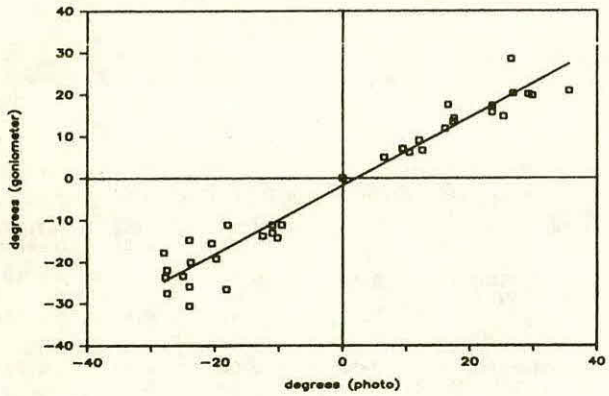
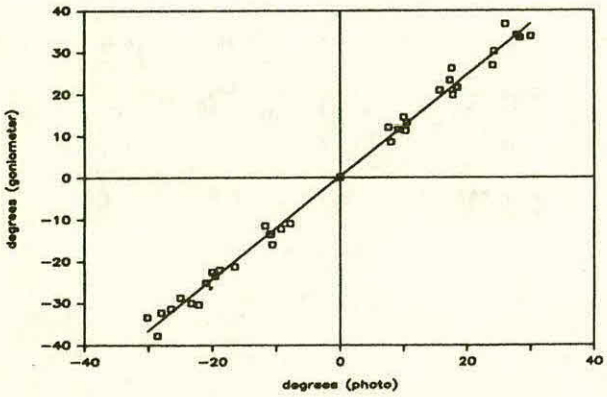
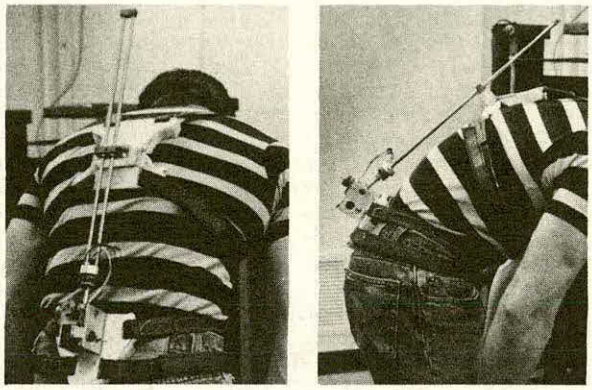


Figure 1c ROTATION



CONCLUSION

1. We have designed a goniometer, worn by the subject, for continuous measurement of trunk flexion, lateral bend and axial rotation.
2. We have compared this instrument to a clinical hand-held goniometer and to photographic measurements of trunk range of motion and have obtained r^2 values of at least .95.
3. This instrument will be connected to a portable FM tape recorder and will provide time history data of trunk range of motion required for working tasks.



THREE-AXIS TRUNK GONIOMETER

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REPRODUCIBILITY OF THE TRUNK FLEXION AND EXTENSION COMPONENTS OF THE MINNESOTA SYSTEM FOR ISOMETRIC MUSCLE STRENGTH

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ABSTRACT

The purpose of this study was to test the reliability of the trunk flexion and extension components of the Minnesota System for Isometric Muscle Strength (SIMS). This strength testing system is computer controlled and includes convenient and effective patient stabilization and load cell placement systems for trunk strength testing. Five subjects were tested on three separate days. Intra-test and inter-test reliability coefficients were very good ($r > 0.90$).

INTRODUCTION

Strength measurements are needed for assessing the functional implications of injury, disease, and disuse. Trunk strength especially needs to be assessed on patients with chronic low back pain, and needs to be included in functional and diagnostic assessments of any patients with musculoskeletal or neuromuscular diseases which cause generalized or localized weakness. Previous devices developed for measuring trunk strength have not adequately stabilized the pelvis, or are not capable of assessing other muscle groups (1,2,3). A device is needed which is capable of measuring multiple muscle groups quickly, efficiently, and accurately. The purpose of this study is to test the reliability of the trunk flexion and extension components of the recently developed Minnesota System for Isometric Muscle Strength (SIMS).

BACKGROUND

During the last two years the University of Minnesota Rehabilitation Engineering Center has been developing the multiple muscle group strength testing system. SIMS is computerized, includes subject stabilization systems and methods for quickly moving load cells. It is necessary to stabilize subjects to isolate muscle or muscle group actions and to prevent forces from being dissipated by movement of the subject away from the load cell. Stabilization is also necessary to prevent the subject from converting an isometric (static) contraction into an eccentric (lengthening) contraction. This system was designed to test only isometric muscle contractions; however, repetitive isometric contractions can be readily monitored and controlled by the computer system for testing endurance. Computer software has been developed for controlling strength testing, endurance testing, and strength training. This system is designed to test the following isolated muscle functions: neck flexion, extension, and side bending; trunk flexion, extension, and side bending; hip flexion and extension; knee flexion and extension; and ankle plantar and dorsi-flexion. Recent improvements on this design will allow the same chair based system to be used to test additional muscle functions, for example: shoulder flexion, extension, abduction, and adduction; elbow flexion and extension; hip abduction and adduction; and handgrip and pinch.

METHODS

Strength Testing System

The signal conditioning and computer software components have been described previously (4). Signal generation and conditioning were provided by Interface Model SM-500/SM-1000 load cells, load cell amplifiers, an analog to digital converter (Interactive Structures Model AI-13) and an Apple IIe computer. The software for strength testing (4) was used to calculate peak force and peak torque.

SIMS includes a chair based apparatus which was used to stabilize the subject and to mount and position load cells. This chair consisted of a framework which supported a seat which could be moved forward and backward, a footrest which could be adjusted up and down, strap and pad stabilization devices and tracks for the adjustable placement of load cells and/or stabilization pads (see Figure 1).

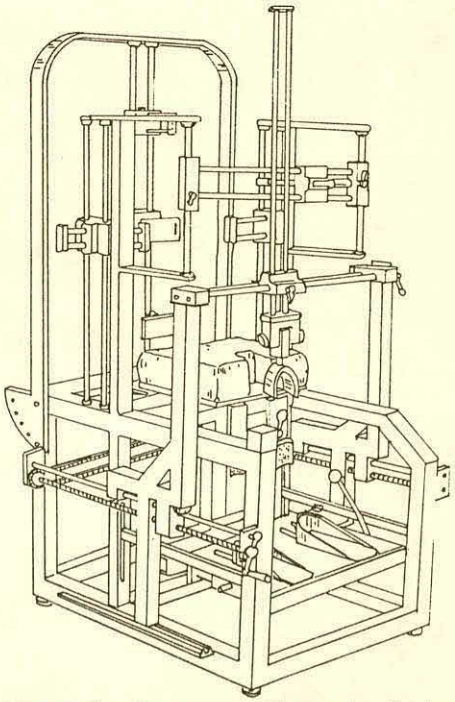

Figure 1. Prototype Minnesota System for Isometric Muscle Strength stabilization chair

The trunk flexion/extension component used only the posterior and anterior tracks. The posterior track was used to position the sacral stabilization pad and to position the load cell and patient contact pad for testing trunk extension force (see Figure 2). The anterior track was used to position the load cell for trunk flexion (see Figure 3). A webbed belt was attached to the seat platform and positioned on or below the anterior superior iliac spines to provide a stabilizing force in a downward and backward direction into the seat and the pelvic pad. Straps were also used to stabilize the thighs and the feet (see Figures 2 and 3).

Procedure for Data Collection

Five normal young adults were tested during three separate sessions which were conducted at approximately 48 hour intervals. Two maximal contractions of the trunk flexors and extensors were obtained during each of the three sessions. During each session the subjects were seated in the chair with full comfortable support of the thighs; and with the pelvis, thighs and feet firmly stabilized. The order of testing of the flexors and extensors was randomized. The load cell contact pad was centered at the level of the spinous process of the 4th thoracic vertebra for testing extensor force and centered at the Angle of Louis (the junction of the manubrium and the body of the sternum) for testing flexor force. The subjects were instructed to push as hard as possible against the load cell when hearing a tone from the computer and verbal commands. Intense verbal commands were given as follows,

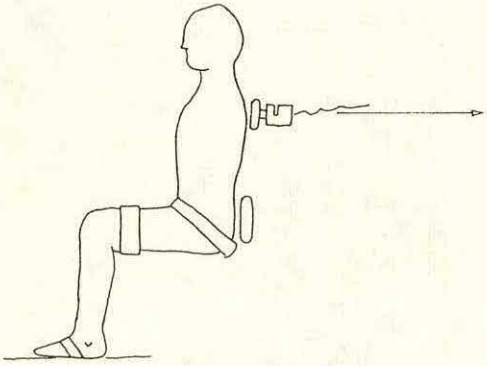

Figure 2. Patient stabilization and load cell placement for trunk extension

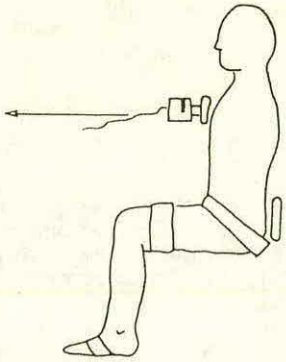

Figure 3. Patient stabilization and load cell placement for trunk flexion

"Get ready...push...push...push...." Verbal commands were given during the entire 5 sec duration of contraction.

RESULTS

The results are given in Table 1. No consistent pattern of change was observed. Intra-test reliability (the correlation between Trial 1 and Trial 2) ranged from 0.91 to 0.99 for peak flexion and extension forces, respectively (see Table 1). Inter-test reliability coefficients (intra-class correlation of sessions 1, 2, & 3) ranged from 0.92 to 0.95. Trunk strength measurements were reliable for both intra and inter-test situations. The mean percent error for Trial 1 vs. Trial 2 ranged from 0.2 to 2.5% and the mean absolute error ranged from 1.1 to 5.5%.

DISCUSSION

Intra-test reliability for measurements of the strength of trunk flexors and extensors with SIMS are similar to the results of previous studies for the flexors and better than previous results for testing the extensors (2). Since the extensors generate roughly twice the force of the flexors, the effectiveness and the comfort of the stabilization system are most critical for good reliability when testing the extensors.

If the subject is not consistently held in the desired neutral sitting position, recorded forces will not be reliable. If the subject is able to move into a more optimal lengthening position for the muscle being tested, the forces will increase. Also if the subject is able to use a more powerful muscle to convert the type of contraction to a lengthening contraction, the force will increase considerably. If the pelvis is not adequately stabilized, the hip extensors can produce this effect by tilting the pelvis. This apparatus appears to minimize these effects.

TABLE 1. REPRODUCIBILITY OF MEASUREMENTS OF THE STRENGTH OF TRUNK FLEXORS AND EXTENSORS: ISOMETRIC CONTRACTIONS, N = 5

	Session 1					Session 2					Session 3				
	Trial 1		Trial 2		Intra-test	Trial 1		Trial 2		Intra-test	Trial 1		Trial 2		Intra-test
Situations	Mean	SD	Mean	SD	r	Mean	SD	Mean	SD	r	Mean	SD	Mean	SD	r
Trunk Flexors Peak Force lb.	86.68	18.13	87.59	24.26	.96	88.99	19.49	87.28	17.41	.91	76.69	13.59	78.10	16.17	.96
Peak Torque ft.-lb.	93.42	22.07	94.06	25.86	.94	90.89	22.56	88.96	19.69	.93	80.55	18.88	82.05	21.26	.97
Trunk Extensor Peak Force lb.	195.50	77.86	189.50	67.27	.99	185.56	42.75	186.48	43.53	.99	183.15	46.12	178.70	42.25	.97
Peak Torque ft.-lb.	239.26	107.77	231.52	93.82	.99	221.59	62.18	222.72	63.21	.99	219.65	66.64	213.95	59.56	.98

SUMMARY AND CONCLUSIONS

The trunk flexion and extension components of the Minnesota System for Isometric Muscle Strength (SIMS) provided highly reliable intra and inter-test results when testing normal young adults.

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PERFORMANCE OF THE LASER SCANNING SYSTEM FOR 3-D MEASUREMENT OF HUMAN BODY SEGMENTAL MOTION

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ABSTRACT

An advanced method of tracking three-dimensional body motion has been developed in order to improve upon the accuracy and bandwidth of existing systems. It uses three rotating planes of laser light, fixed and moving photovoltaic diode targets, and a pipe-lined architecture of analog and digital electronics to locate multiple targets in an eight cubic meter volume. Data collection rates are a function of the laser scan rotation speed and are currently selectable up to 480 hz. Tested system performance at a 480 hz data rate includes a resolution of 0.8 mm (0.03 inches), a repeatability of ± 0.635 mm (± 0.025 inches), and an accuracy of ± 2.0 mm (± 0.08 inches) with all results applicable at the 95% level of confidence along each coordinate direction. The present accuracy in reducing position data to angular body orientation is ± 0.5 degrees. Moving targets can be tracked at speeds exceeding 2 m/s (typical of ankle motion) with signal integrity tested but not limited to 25 hz motions.

INTRODUCTION

There has been a need to accurately track body motion in many scientific and medical fields for quite some time. In the last three decades, rapid advances in technology have led to the development of a variety of techniques capable of achieving this goal. One application area is the study of human motion in order to develop better quantitative methods of evaluating human performance and understanding pathological kinesiology.

One particular aspect of human motion under study with broad neuro-musculoskeletal applications is gait function. Many instrumentation systems have been developed for measuring lower extremity motion including systems based on photogrammetry, electrogoniometry, sonic triangulation, accelerometry, and electro-optical phenomena. During the last decade, particular attention has been focused on electro-optical techniques. Within this group are several systems currently marketed and used in gait laboratories: VICOM, SELSPLOT, CODA-3, and the United Technologies (UDT) systems. Each system quotes comparable performance characteristics sufficient for the purpose of collecting human motion data, but each has its own idiosyncracies inherent to their fundamental technology, which limit their utility.

In an attempt to provide a cost effective, high performance approach, the system described here has been under development with limited support for several years. Our present prototype has undergone extensive testing and calibration as a means for body motion measurement. This paper reports on our most recent results.

LASER SCANNING SYSTEM

A technique has been developed and adapted for the purpose of three-dimensional gait analysis (1,2). The founding concept is that three planes with differing normal vectors intersect at a point in three-dimensional space. Using three or more targets per rigid body, the position and orientation of the body can be calculated. This information ultimately defines angles and axes of rotation for the joints observed.

To generate the planes of light, Helium-Neon lasers are fitted with a lens arrangement which allows adjustment of the dispersion angle and the focal distance. Each laser is directed at a rotating mirror, which sweeps the plane through the target volume at a constant angular velocity. The mirror is an octagonal prism attached to a 60 hz reluctance synchronous motor. This allows for a high data collection frequency, 480 hz, with off-the-shelf components. Although the multi-faceted mirror provides a desirable sampling rate, the reflective surface is at a finite distance from the mechanical axis of rotation. This displacement causes the optical axis of rotation, that of the light plane, to move with respect to the mirror's angular orientation. As the location of these axes are significant variables in the system, the modelling of this motion was found to be important in order to maintain the accuracy. Thus, rather than considering these terms negligible, a detailed model was derived from differential geometry and implemented in software.

In the previous version of the system, the axis of rotation of each plane was arbitrarily oriented with the axis position and orientation computed based on an open form iterative solution (3). The solution time, however, was prohibitively slow. To eliminate the numerical solution and thus greatly speed up the computation phase of the measurement, the axis of rotation for each laser was set orthogonal to each other and to the global coordinate system. The orthogonality of the axes of motion ensures a closed form solution to the equations. The process of orienting each laser is an interactive one using measurement feedback. The result yields two planes with vertical axes and a third with a horizontal axis. The solution is thus reduced to two 2-dimensional calculations per three-dimensional point.

As indicated earlier, the location of each axis of rotation must be determined in order to effectively use the system. Other parameters that must be identified are the location of the initial and final reference photodetectors for each laser. To locate the positions of the six fixed reference diode targets and a local coordinate frame for each moving axis of rotation, a ZEISS precision measurement system was used. This measurement device consists of two theodolites interfaced to a computer. Once the theodolites are calibrated to

provide their locations with respect to a global coordinate frame in the room, points can be located with respect to the defined frame with accuracy and repeatability better than 0.003 inches. To do the calibration, four or more points of known location are needed. These points were located on an optical bench (2 m x 2 m) which straddles the target volume during the calibration process. This removable test bench is used to define the global coordinate system and the ZEISS calibration points which were located with a six-foot vernier caliper. With the references permanently attached to the floor and ceiling and the axes of rotation oriented, the three mechanical axes of rotation and the six reference detector locations may be measured. With all system parameters defined, the data generated by the system can be reduced to three-dimensional points.

With all dependent variables determined, the information needed in order to calculate x,y,z points includes the independent variables, i.e., the angles of rotation swept by each laser from the initial reference point to the target point for each plane. This is done by converting the elapsed time from the reference trigger to the target trigger to an angle. This time is determined and stored on the data acquisition board. The signals from the analog electronics act as interrupts on the data board whenever a plane passes a target. The current clock value, which is reset whenever a plane is at an initial reference, is then stored in the appropriate register to be polled by the host CPU. This polling arrangement is only a temporary one; a newer system memory maps the register for faster performance. After the data is stored in the host, the timing data can be converted to x,y,z points.

CALIBRATION

To determine if the data generated by the system is accurate, a detailed calibration was done. The process involved moving a vertically-oriented, linear array of eight targets through the target volume. At each discrete location, data was collected by the laser scanning system and the locations of each target was recorded by the ZEISS system. In all, 616 locations were sampled within the target volume. Comparisons of the two methods did initially show a large, regular disagreement (2.5 to 3 inches) between the two methods. A thorough sensitivity analysis on the reduction equations showed that the sensitivity of the results to the fixed system parameters were all less than unity, but the sensitivity to changes in the scan angles was more than 100. Using the theodolite data and the fixed system parameters, an error analysis was performed on several models relating the timing data and the measured coordinated displacement. The results of these analyses showed that the errors could be represented by a phase lag in the timing data. A possible source of this error is a capacitive effect associated with the photo detection.

By making the necessary software adjustment, the error was removed. After this correction, the error between the two systems dropped to a worst case magnitude of 2.0 mm (.08 inches) along each of the 3 orthogonal axes or 3.5 mm (.14 inches) along the 3-D vector between the measured and actual points in space. These results were achieved at 480 hz data acquisition rates. It should, however, be noted that during the calibration process, heavy construction was underway in the vicinity of the laboratory. This could not be avoided due to the inflexibility in rescheduling the loan of the computerized theodolite system. The results of the calibration nevertheless demonstrated that this first prototype system could perform even under unusually poor circumstances within an acceptable tolerance for human motion study.

DYNAMIC TESTING

Two additional tests were implemented to test the system's performance. The first was a single pendulum placing the eight targets on a 1.75 meter bar; the resulting low frequency motion covered an entire planar section of the target volume in an oscillatory pattern. Data was collected at frequencies from 60 to 480 hz and showed, as expected, that there were no local discontinuities or global warping within the target volume.

The second test was to determine performance under small displacement and high frequency conditions. A single target was attached to the end of an aluminum rod, whose other end was mounted on the shaft of a closed loop position controlled DC servomotor. The resulting radial distance to the target was 6.6 cm (2.6 inches). The proportional controller applied a saw tooth oscillation to the target tipped rod at a specified frequency. This resulted in a constant angular velocity during two phases of each oscillation. Incorporated in the feedback loop was a high resolution encoder, which provided an accurate angular displacement measurement independent of the laser scanning data.

The target was driven at frequencies from 0.5 to 25 hz; both the target and the encoder were monitored. Position data in cartesian format from the scanning system was reduced to a polar format by a cartesian to polar conversion using circular least squares fit to define the circle center. With both sets of data in the same format, comparisons could be made. Fourier signal analysis showed that both sets of data matched in amplitude and phase at the excitation frequency, but the encoder data began to display a slightly lower amplitude at low frequencies as the excitation frequency increased. The angular motions were compared against each other. At low frequencies, there was no deviation, but as the frequency increased, slight differences occurred with the magnitude of the laser scanning angle starting to increase with respect to the encoder angle. A cartesian plot of the target position shows regular deviations from the expected circular path which can be attributed to bending. This corroborates the frequency domain results since bending would increase the low frequency magnitude of the laser scanning data only.

From this test, two observations can be made. The system has the ability to record motions at

frequencies far above those present in human gait. The second is that the system can record higher frequency dynamic deflections which may make it a useful tool for measuring and characterizing soft tissue motion.

RELATIVE ORIENTATION BETWEEN BODIES

One of the motivations in developing this system is to measure the 3-D rotational and translational displacements with respect to each other. This information can be used to determine the 3-dimensional joint rotations about a moving axis of rotation and can be obtained based on the premise that three or more points define the position and orientation of that body. There are many ways of reducing a set of position data to orientation information, one of which is the SCHUT algorithm (4,5,6).

This is a position averaging technique which gives the relative change in position between two bodies. The method used in our system is quite similar, but instead of using position averaging, it averages the orientation to reduce error and to determine changes in orientation. The results from the two methods agree within one degree in most of the cases tested.

To check the accuracy in determining these angles, a single degree-of-freedom jig was designed with each of the two links holding a cluster of four targets fixed with respect to each other. By displacing the jig throughout the measurement volume with the hinged joint angle fixed, the measured/computed orientation angle can be compared to the actual angle. This was done for a range of angles for each of the three body planes. The results of this test placed the error of this measurement at $\pm$ 0.5 degree at the 95% confidence level. This result represented the accuracy in predicting the orientation angle throughout the entire target volume for each of the three body planes.

Calculation of the "instantaneous" axis of rotation of the joint in question is possible. This measurement can be performed on the system using the reduction algorithms (i.e., SCHUT), but is not entirely stable due to the nature of the computation. This is especially a problem with bodies oriented 180° relative to each other. A new and different method of calculating the orientation between two bodies has been developed based on duality theorems and spatial kinematics (7). This method bypasses the instability conditions in other algorithms and thus provides a more accurate result for the rotation angles and the axis of rotation.

CONCLUSION

The system described provides a fast and accurate method to record human locomotion parameters. The tests reported here were conducted on the initial prototype pending the imminent completion of a newer system. Earlier prototypes were only capable of tracking eight targets; thus data is available only for one joint. The system displays good dynamic bandwidth and acceptable performance. Data collected for 1.75 seconds at 240 hz for 2 limb segments (4 targets per limb) can be reduced to 3-D relative orientation or joint angles on our existing prototype in approximately 20 seconds on an Intel 8086 based system (with no attempt at optimizing the code). A preliminary test with an array processor has cut this figure almost in half. Significantly better performance is planned for the future.

ACKNOWLEGEMENTS

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REAL-TIME ANIMATION OF HUMAN WALKING FOR THE EVALUATION OF PATHOLOGICAL GAIT

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ABSTRACT

A system has been developed for graphically animating human motion during gait. This animation will be directly linked to biomechanical data collected in a gait laboratory or to mathematical models. The results facilitate improved understanding of the significance of the many degrees of freedom associated with each of the joints.

INTRODUCTION

Interest in computer generated humanoid figures began in the late fifties when Boeing created a static silhouette of a signal officer on an aircraft carrier which was used in order to study the viewing fields of pilots in cockpits (7). As computer graphic technology developed in the latter sixties and early seventies, more sophisticated models followed (1,7). Research and development into computer animated images was oriented toward facilitating real time motion capability without sacrificing on detail. Often, the motivation for this work was to study the effects of anthropometric constraints associated with the man-machine interface on the design of new products.

DATA DRIVEN ANIMATION

Due to the frequent difficulties associated with interpreting the significance of biomechanical data collected in the gait lab, researchers have been drawn toward the visual medium of computer graphics. Our objective, similar to others (7), is to integrated real time animation of full three-dimensional walking humanoid figures with a biomechanical data acquisition system. It is, however, critical that the data collected be an accurate representation of the three-dimensional motion, ergo the effort in developing a new laser scanning device for tracking human motion (6). The resulting system will provide clinicians with the ability to "see" the correlation between the collected biomechanical data and the actual human motion. An automatically generated humanoid figure can "walk" by playing back the data collected from the lab. The advantage of such a system includes the ability to examine the gait from every viewing angle, to zoom in, back off, or to stop the motion. One can "play back" the data without calling a patient back into the lab. Unlike film or videotape, the result can be viewed from any desirable perspective, and can include the "removal" of extremities, such as the arms, if the line of sight needs to be improved. The resulting animation will make it possible to visualize the significance of the timing, of the joint motion, of the three-dimensional angular relationships, and of the foot/floor contact. All these contribute in some measure to the "gestalt" of walking.

The system has the ability to superimpose "normal" gait on a "patient's" gait in order to visualize differences. Similarly, one can animate two pathologies simultaneously in order to differentiate between the resulting gait patterns. Another ramification of superposition is the comparison of mathematical models of walking with actual human walking by playing back the appropriate animated humanoids (one representing actual human generated data and the other controlled by the model). In this way, it is possible to refine the equations until the model controlled humanoid motion is an accurate representation of human walking. Improved mathematical representations which incorporate the effects of different treatment modalities can become a basis for experimentation. Different treatment plans can be tested and compared on the model prior to testing on actual patients. The system, therefore, serves as a testbed for computer aided design of treatment strategies.

Figures 1a and 1b show a subject with a hypothesized above knee prosthesis superimposed on a "normal" gait pattern at a particular instant in the cycle. Figure 1a illustrates the effect of the prosthesis on both the vertical and horizontal displacement of the body center of gravity (C.G.). Figure 1b shows the same frame, but with both bodies alligned to a common C.G.; as a result, the difference in joint angle at the knee can be perceived. Table 1 describes the independent degrees of freedom modeled at present and others which will be incorporated in future versions.

SOFT TISSUE DYNAMIC EFFECTS

A data driven graphic animation system for walking has additional implications. All existing measurement systems track the gross motion of the body segments at their surface. Present technology does not allow for in vivo measurement of direct skeletal motion due to soft tissue motion. Soft tissue dynamics are a function of muscle contraction, muscle mass, and its distribution, the deformation properties of adipose tissue, body segmental position, and their location relative to gravity. The results are such that there may be gross errors in predicting the internal motion of the skeletal structure (7). These can become significant in understanding pathologies such as degenerative joint disease where knowledge of the exact motion of the skeletal links can be useful in predicting the impact of the damaged cartilage on gait.

In order to improve our understanding of this problem, we have begun to explore the inconsistencies between the gross measurements of body segmental motion during walking and the integrity of the kinematic skeletal model driven by measurement data. Without considering soft tissue dynamics, the motion data can lead to predictions of segmental motion which are

incompatible with the anatomy and the biomechanics of gait. Through the use of an accurate computer generated graphic animation, inconsistencies in the data at various stages of the gait cycle can be determined.

FEATURES OF HUMANOID ANIMATION SOFTWARE

The animation software runs on a VAX 11/780 under the VMS operating system, coupled with an Evans and Sutherland PS300 color vector refresh graphics workstation. The humanoid data base for visualization and subsequent animation is based on a software package, MnCELL, which had been previously developed for simulating machine motions (4). MnCELL enables the user to see the animation from any angle or distance, to accelerate or decelerate the motion, or to stop it at any particular point in time. The software can, furthermore, remove visually hidden lines to produce realistic "solid" images.

Unlike the humanoid animation systems developed elsewhere (1,7, E&S's ADAM), this package allows the experimenter or clinician to define a complete humanoid figure proportional to the anthropometric measurements. In addition, the user has complete control over a number of functional details of the particular humanoid. For example, the investigator may alter any number of degrees of freedom associated with one or both of the knees to study joint prosthetics and their effects (Figures 1a and 1b). Furthermore, a joint or body segment may be "tagged" by changing its shape or color in order to better identify particular phenomena.

Figure 2 shows a walking subject viewed from an arbitrary perspective while at the same time tracking the sagittal knee angle trajectory. The cursor moves along the graph in correspondence with the knee motion. On a color display, the graph is the same color as the "tagged" knee joint. It is possible to have many more graphs tracking any number of joint or limb motion trajectories displayed during animations. An exploded view of the full gait cycle can then be generated to scale along the walkway in order to visualize particular events which occur rapidly. To illustrate this, the complete gait cycle in Figure 3 has been expanded along the time scale as shown in Figure 4.

Note that in order to facilitate real time animation on a vector refresh system, no attempt is made to remove the lines not normally seen in viewing solid objects. This, however, can cause confusion when printing out "snapshots" of the animation. Hidden line elimination software removes this artifact to provide more realistic imagery on the hardcopy. We are now looking at raster based technology in order to improve our capabilities further.

ACKNOWLEDGEMENTS

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TABLE 1

Body Segments	Independent Degrees Of Freedom: Presently Implemented:	Independent Degrees Of Freedom: Will Expand As Needed:
lower extremity phalanges	SR	*
ankle	SR	VR, FR
knee	SR	VR, FR, rVT, rST, rFT
hip	SR, FR	VR
pelvis[1]	aVT, aFT, FR	aST, VR, SR
thorax	FR	VR, SR
shoulder	SR	*
elbow	SR	*

[1] body coordinate frame

Key:
VR = Rotation about an axis which is coincident to a vertical body axis
SR = Rotation about an axis normal to the Sagittal plane
FR = Rotation about the Fore-aft axis
VT = Translation along the Vertical axis
ST = Translation along the medial-lateral axis (i.e., the axis normal to the Sagittal plane)
FT = Translation along the Fore-aft axis
* = No additional degrees of freedom needed
a = absolute motion in space
r = relative motion with respect to an adjacent body segment.

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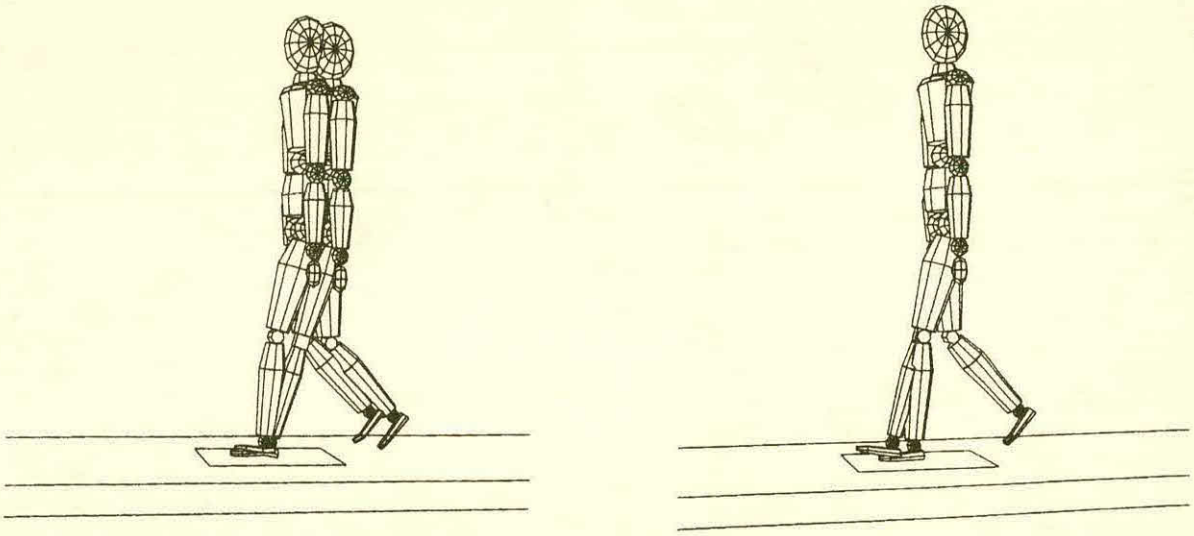

Figure 1: Gait for normal subject and above knee amputee can be superimposed in time and space by:
A: aligning the gait cycle B: by aligning body centers of gravity.

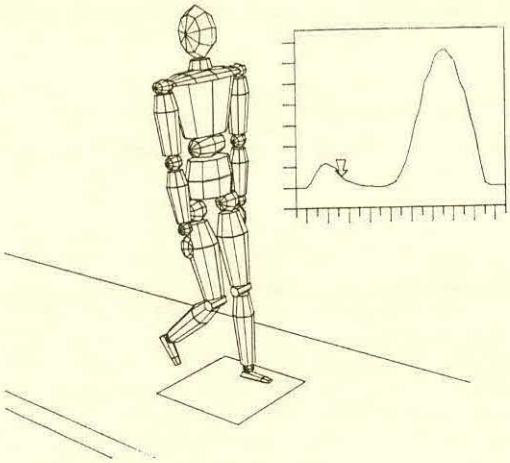

Figure 2: Sagittal knee angle trajectory corresponding to left knee of subject (color coding not shown)

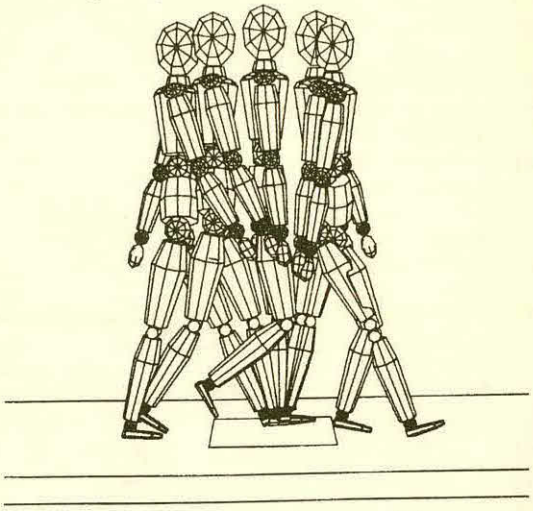

Figure 3: Walking cycle with proper time/distance scaling

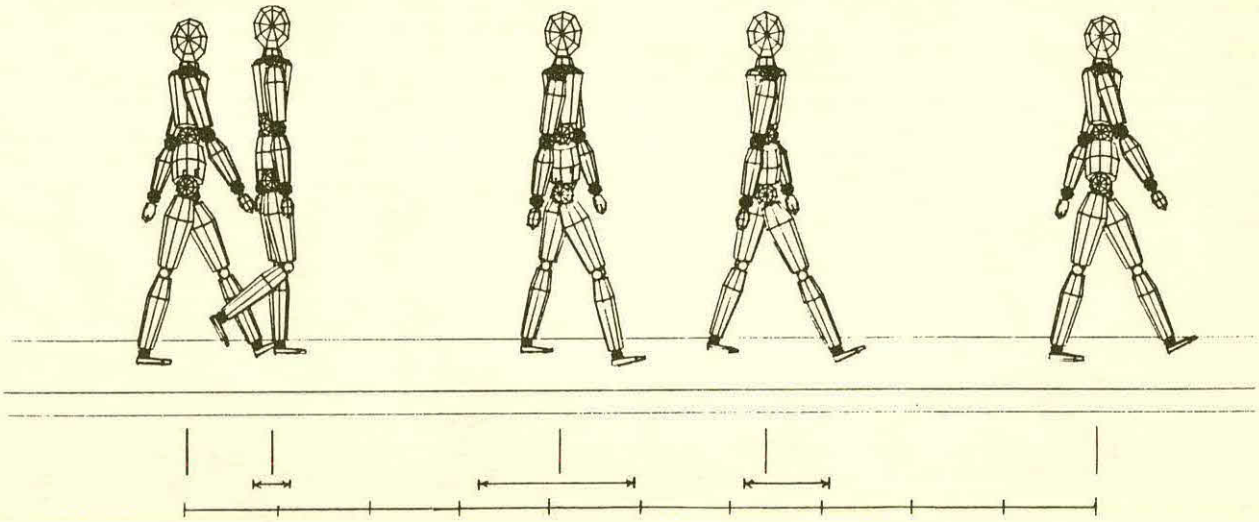

Figure 4: Walking cycle "exploded" along a time scale

A TRAINING AND REFERENCE MANUAL FOR AN INTERACTIVE ROBOTIC AID

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ABSTRACT

The first generation Veterans Administration/Stanford University Robotic Aid was developed to give individuals with high level spinal cord injuries increased control through the ability to manipulate their environment. The training and reference manual to that robotic system is discussed in this paper. This manual is best described as a historical, instructional and informational guide to the Robotic Aid. While it does not presume to set a standard for developers of other robotic systems or rehabilitative devices, it will be used in the production of the manual for the second generation VA/SU system.

INTRODUCTION

The training and reference manual has evolved with the Robotic Aid Project. It began as a small set of training exercises and grew to include instructions ranging from how to start up the system hardware and improve command recognition, to suggestions for troubleshooting and information on the technical aspects of the system. The many different versions of the manual reflect well the changes that have occurred in the five years of first generation development. These changes are most evident in the current name of the vocabulary (VFIVE is the fifth rendition of the command set) and the addition of graphics to the text. Its readers include technical and clinical researchers, experienced and first-time users of the Robotic Aid.

Work on the training and reference manual was suspended in 1984 with the funding of the proposal *Development and Evaluation of an Advanced Manipulation Aid for the Severely Disabled* [1] and the commencement of work on the second generation system– a mobile robot with increase sensory capabilities, dialog management and strategic motion planning facilities, and alternate control interfaces. The current edition of the manual will be used as a foundation for instructions on, and guidelines for, the system under development.

BACKGROUND

Clinical evaluation of the first generation VA/SU robotic system began in October 1981: formal training of disabled individuals began in 1982. In an effort to provide trainers and users of the robotic aid with standardized instructions for optimum usage of the device, a training and reference manual was developed by members of the clinical evaluation team. The purpose of this manual was to communicate the features and intent of the robotic aid, to familiarize the naïve user with the system, and to present a format for training.[2]

The need for directed training on this system is pronounced. The Robotic Aid is an interactive system requiring both a human operator and machine intelligence, and is intended for use in an unstructured human living environment. As this system does not operate in the same context as an industrial robot that is programmed to do repeated tasks in a controlled environment, the user must be able to "perform the highest supervisory function by deciding what task is actually carried out" and "explicitly guide the robot's motion using his own sensory and decision making powers".[3] This necessitates a certain level of control expertise and requires that users spend a certain amount of time with the robot. Not only must individuals be familiar with the commands to which the robot responds, but they must also be able to anticipate the motions that those commands evoke.

METHODS AND MATERIALS

The format of the manual is designed to introduce the user to the robot, give instruction on voice training to allow for optimum recognition accuracy and lead the user, step by step, through exercises to learn the different features of the Robotic Aid (see figure 1). Detailed information regarding system procedures, possible problems, error recovery (see figure 2), command architecture and technical requirements of the system are available in support. Computer and human generated graphics are included to assist the user in real-time control of the robot (see figures 3 and 4). Conventions for communication between the user and the robot are established early on: words that are typed in by the user are highlighted in **THIS FONT** while words to be spoken appear like this, "**HELLO**". This manual benefits greatly from the flexibility of TEX, the typesetting emulator developed by Donald Knuth.[4] In addition to providing multiple fonts, TEX makes available distinctive characters that act as visual cues to the user, i.e., ↩ translates as "Hit the RETURN key on the keyboard".

The final version of the training and reference manual is produced on a QMS lasergraphix printer using TEX and GrafTEX, a library of TEX-compatible graphic subroutines created at the Rehabilitation Research and Development Center. TEX and GrafTEX are flexible media that provide crisp and concise, but

Your Response (what you say)	Robot Response (what it says)	Notes
1. READY		
2. ATTENTION		
3. ZERO	"Zero. Are you sure?"	Confirm with "**YES**".
4. PARK	"Park. Are you sure?"	Confirm with "**YES**".
5. BYEBYE	"Byebye. Are you sure?"	
6. YES	"See you later!"	At this point the arm can no longer be accessed by your voice.

Figure 1. Sample training exercise.

PROBLEM (What the robot thought you said)	SOLUTION (What you say to recover)	RESULTS (What should happen)
A change from the mode you were in (ex: from PILOT to WORLD).	Say the name of the mode you were in (PILOT).	Robot goes back to the previous mode, PILOT. (**P** underlined on the Argus display).

Figure 2. Sample instructions on recovery from misrecognized commands.

easy to modify, text, tables and graphics and the laser printer allows for inexpensive, quickly produced, high-quality documents. Editing occurs on-line so a new edition can be 'published' within an hour.

RESULTS

This manual has been used, in some form or another, in the training of over 100 individuals. Many of the tasks, designed to test mastery of particular command sequences, are based on needs expressed by initial users of the system (e.g.,getting a drink of water or picking up a book and placing it on a shelf). Feedback has been an important part of the development and evaluation cycle. Direct results of user feedback include: a voice management program with options for training, updating and saving a vocabulary; and on-line instruction for recovery from a fatal robot error. (The arm will turn itself off when it collides with another obstacle.) In both cases, user autonomy has been increased permitting greater interaction with the robot without greater intervention by a supervisor or attendant.

DISCUSSION

Development of the manual for the second generation advanced manipulation aid is in its preliminary stages. One format being considered is the integration of instructions into the robotic aid interface like an on-line 'help' facility or using interactive video technology. A master copy of the manual would be updated regularly so that a state of the project report would always be available. A user could submit a request for a hardcopy simply by saying, "PRINT". Another option includes a perforated *cheat sheet* that would summarize instructions on how to start up the system and invoke particular programs or give hints to recover from command misrecognitions. It would serve as a quick reference for experienced users and eliminate the need for the printed manual as an encyclopedia.

Short term efforts will result in graphics produced solely by computer. Engineers at the RR&D Center have developed a link between the Apple Macintosh and the Digital VAX 11-780, used now in manual production. Once this link is refined and proven reliable, free form, non-geometric graphics will be possible.

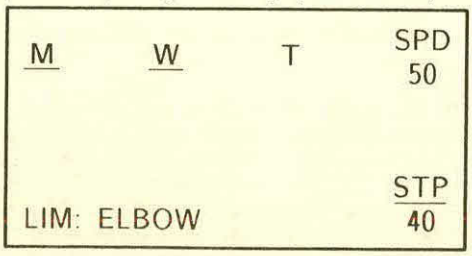

Figure 3. Graphic representation of the Robotic Aid status display. It currently shows that the elbow joint of the arm is at a limit stop.

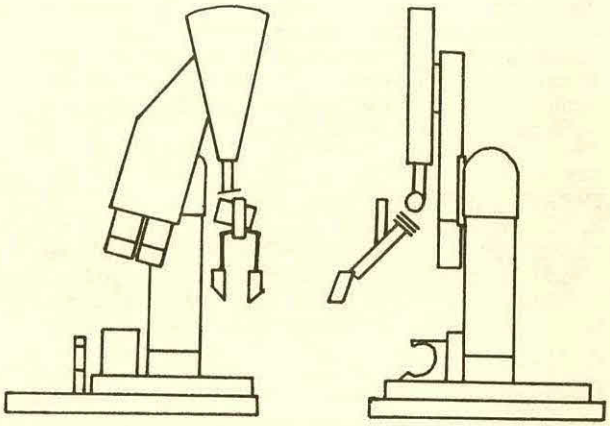

Figure 4. Pen and ink drawing of the robotic arm. It shows the user what a limit stop on the elbow looks like.

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Copies of the current *The Robotic Aid Training Manual* (246 pages) are available, for the cost of reproduction, from the author.

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AUGMENTATIVE MANIPULATION

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ABSTRACT

Advancing technology is increasing the number and types of manipulation aids available to the person whose disability involves upper-extremity impairment. Achievement of effective manipulation depends upon an understanding of the attitudes and characteristics of the user, the characteristics of the aids and their functional relationships, and the properties of the manipulated environments. The term "augmentative manipulation" is proposed to identify a holistic approach to supplementing manipulation skills.

INTRODUCTION

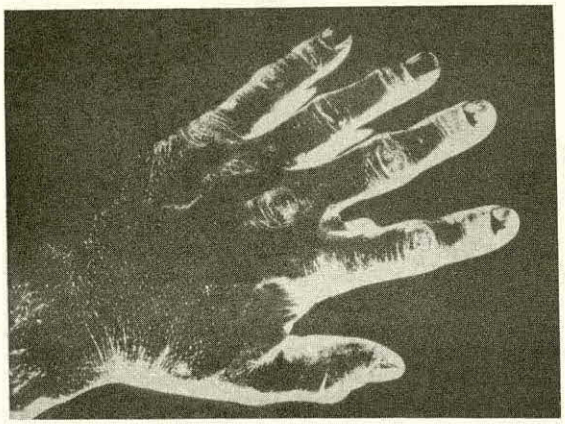
At one point in the movie *Blade Runner*, an android, one of several designed for space exploration, is challenged by a genetic engineer to demonstrate his skill at chess. The android scoffs in reply, "We're not computers, Sebastian, we're physical."

Like that android, we, as human beings, have been designed, whether by evolution or divine intervention, as physical beings. We learn by manipulating the physical world, and later, as our thinking gains in sophistication, to use our physical manipulations as metaphors to achieve insight into abstract concepts. We use the environment as a tool to act on ourselves—physically, mentally, aesthetically. We shape ourselves by shaping our environment. The more control we have over manipulating our environment, the greater are our opportunities for reshaping ourselves.

The principal media through which the physically intact person manipulates the environment are the hands. From dealing with objects, to accessing information, to maintaining personal hygiene, the hands serve as a versatile physical and sensory extension of the mind. So closely is the control of our hands coupled to our minds, that we are rarely aware of the subtle manipulations we perform as we carry out a task. When working through our hands, we almost never think about our hands, but rather about the work "at hand".

In the case of a disability affecting manual skills, much of a person's rehabilitation is centered on substituting or compensating for the loss of physical and sensory abilities mediated by the hands. A variety of devices, primarily used in occupational therapy, have been developed to assist in specific manual tasks (such as hygiene, dressing, meal preparation, and writing) where some residual hand/arm function remains. The total or severe impairment of hand function, however, requires a range of assistive devices because the disability more broadly affects manipulation of the physical environment. Some devices are needed to extend the individual's intact sensory abilities into the environment—to "couple" the user to the objects being acted upon. Some are needed for action at a distance since severe physical impairment frequently includes impairment of mobility. Some devices are needed to facilitate tasks requiring continuous interaction between the user and the object or objects being manipulated. Other devices are needed for momentary tasks which do not require close involvement of the individual.

The following sections will describe some of the basic characteristics of manipulation aids, offer comments on factors which influence their relative utility, and finally propose that a holistic approach, emphasizing the functional relationships between the user, the manipulation aids, and the environment, be fostered in the supplementation of manipulation skills.



The challenge of augmentative manipulation: to supplement or replace the impaired physical and sensory abilities of the upper extremity by a set of complementary manipulation aids and manipulation strategies.

The delineation of aids and of factors influencing their utility is not intended to be all encompassing, but is presented to promote an awareness of the relative functions of different aids and to provoke thought about the design and application of manipulation aids.

MANIPULATION AIDS

Manipulation aids can be divided into two categories; those that are task specific and those that more generally assist in manipulation. The former category would includes aids for eating, for personal hygiene, for dressing—aids which perform a specific function within the context of a task, but which have no function outside that context. In the category of general purpose aids, I include hand orthoses, mouthsticks, headpointers, environmental control systems, upper-extremity prostheses, remote manipulators and robotic devices, and other persons in the environment—manipulation aids which each can be used for a variety of tasks.

In this discussion, the concern is with general purpose manipulation aids.

Hand orthoses stabilize the hand and wrist for pushing and pulling objects close to the body or for holding down an object that is being manipulated. The orthoses also provides a platform for the attachment of task specific tools, such as eating utensils or writing implements. And, if powered by body movement or external actuators, a hand orthoses can provide basic prehension.

Mouthsticks and headpointers, like hand orthoses, can be used to push objects about which are close to the body. They are

especially useful for activating push-button devices (such as keypads of electronic calculators, telephones, and appliances; keyboards of typewriters or computers; elevator buttons; and push-button environmental control systems). With interchangeable tips, they can also be used for writing or artwork. Some commercial and custom mouthsticks and headpointers also provide a mechanism for tip prehension, for grasping and positioning relatively small and lightweight objects.

Environmental control systems serve as interfaces for the control of a variety of electrical and electronic devices. They provide action at a distance. For the individual who may be relatively immobile at times or whose energy is better spent than maneuvering from one appliance to another, the environmental control system extends the range of control.

Upper-extremity prostheses do not have direct application to the motor-impaired individual without an amputation; however, many concepts developed in the control and design of prostheses may be relevant. Much of the work in prosthetics has dealt with substituting control sites to compensate for the amputated joints. And many schemes have been developed to expand a small number of control sites into a greater number of control signals for the operation of multi-joint prostheses. The motor-impaired individual is in a similar situation requiring both substitution of control sites to compensate for the affected joints and an expansion of the remaining control sites or movements to increase the repertoire of function. Upper-extremity amputation also results in significant sensory deficits which have been the focus of extensive research on sensory substitution and feedback. This body of work is already available to those dealing with the deficits of motor/sensory impairments, such as result from injury to the spinal cord.

Remote manipulators and robotic devices hold the promise of a general manipulation instrument. They can provide action at a distance. They can be used in unstructured environments or carry out pre-programmed tasks within highly structured environments. They can be used to manipulate larger and heavier objects than an individual could handle with body-actuated manipulation aids. And they can be provided with the ability to accommodate certain characteristics of the environment, supplementing the commands of the user. The key to the fulfillment of this promise is the control interface between the user and the manipulator/robot.

The most versatile manipulation aid is, unsurprisingly, another person. People are generally readily available. A sufficient number of them are responsive to requests for assistance. And enough of those have adequate intelligence to carry out carefully worded instructions. In addition, the environment and objects within the environment are peculiarly suited to manipulation by people. Regardless of whatever other manipulation aids an individual may have available, emphasis should be placed on mastering the ability to direct another person to carry out any task that may be required. Having this skill is both a precaution against the failure of an assistive device and a necessity since many tasks are impractical or impossible for a severely disabled person to perform.

COMMENTS

Locus of Control and Control "Coupling"

One aspect when considering the relative role of a manipulation aid is the perceived locus of control, i.e. whether the aid is perceived as an extension of the body or as a remote effector controlled by, but not closely coupled to the user. The degree of coupling affects the mental loading of the manipulation aid, i.e. the amount of mental effort expended in controlling the aid. Hand orthoses, mouthsticks, and headpointers are examples of closely coupled manipulation aids. Because of their intimate mechanical coupling to the body, they serve as physical and sensory extensions of the body. In the case of a hand orthosis, forces exerted on the hand are perceived by the sensory apparatus of the most distal intact joint. For the mouthstick, the skin and proprioceptive senses of the mouth and neck are extended to the tip of the mouthstick much as the sensory apparatus of a blind individual's hand and arm is extended to the tip of a cane. The sensory apparatus of the head and neck are extended through the headpointer.

The wealth of sensory information available through these body-extending manipulation aids greatly improves the control of these aids. The user has a mechanical sense of the effect of his or her actions on the objects being manipulated and does not need to rely purely on visual inspection. Little mental effort is expended in the control of the aid, allowing the user to concentrate on the task.

Switch or voice-operated manipulation aids, on the other hand, may not take full advantage of intact physical control abilities or of intact proprioceptive senses. Because the aid is mechanically uncoupled from the user, control of the aid requires closer attention by the user and external supplemental feedback, such as command confirmation, to insure proper transmission of control commands. The mental loading is higher when the user is uncoupled form the manipulation aid because the user's attention must be divided between control of the aid and performance of the task. The aid, in effect, becomes an intermediate object to be manipulated.

Continuous or Momentary Interaction

The influence, on performance, of coupling through the control interface will depend to a great extent on whether the interaction is continuous or momentary. In the fields of human factors and control, it is generally considered more effective for machines requiring moment-to-moment adjustments to be controlled as extensions of the operator. Examples would include driving a car, piloting an aircraft, or handling nuclear materials or chemicals with remote manipulators. The operator is able to respond more quickly to errors in control or changes in the environment and is able to more accurately grade the response in proportion to the need because of the feedback afforded though the mechanical coupling.

Likewise for the motorically impaired individual, tasks requiring a series of manipulations, as in setting up books and notes for studying or in preparing a meal, might be performed more quickly and efficiently if the controller of the manipulation aid extends the physical control and sensory abilities of the individual. If, however, the task is momentary, requiring only brief attention from the user, such as turning an appliance *on* or *off* or initiating a pre-programmed set of actions of a robotic device, than the advantage of a physiologically-coupled controller is less. Other types of controllers (switch-type or voice-activated) may serve just as well for these situations.

Influence of Control on Tasks Selected

Although remote manipulators and robotic devices may have the mechanical characteristics to carry out a variety of tasks, the types of tasks mediated through these devices are likely to be chosen based on the type of control used. As has been learned from the design of multi-function upper-extremity prostheses, control needs to be natural, or intuitive, and the system be highly responsive to that control; otherwise, the mental effort expended in the control of the device in a continuously interactive manner overshadows the benefit of the task performed through the device.

If the control interface of a remote manipulator or robotic device provides the physically impaired individual a natural extension of intact physical and sensory abilities, the individual will be more likely to use the device interactively in novel situations and unstructured environments. If, on the other hand, the control interface is such that a significant degree of planning and attention to the actions of the device are necessary for successful use, the individual is more likely to use the device for repetitive tasks which can be pre-programmed and which are carried out in a highly structured environment.

View and Viewpoint

Other factors, such as the view of the object being manipulated will influence utility of a manipulation aid for certain tasks. In upper-extremity prosthetics, hook-like terminal devices are frequently recommended over hand-like devices for individuals with bilateral limb loss. Although the hook-like devices offer a greater variety of prehension patterns, they also, because of their slender frame-like construction, present the user with less visual obstruction of the work area and of objects being manipulated.

The tips of mouthsticks and headpointers offer little visual obstruction. But a hand in an orthosis may obscure the work area because of the restricted range of approach to the work resulting from the impairment of wrist and elbow function. As with the terminal devices of the bilateral amputee, the orthotically supported hand cannot be easily repositioned to approach the work from a different angle. Instead, an extension to the orthosis may be needed in order to set the hand back from the immediate work area, but maintain a mechanical link between the objects being manipulated and the individual.

Remote manipulators and robotic devices may be positioned so as to place objects in clear view while they are manipulated. But whereas the user of a hand orthosis, mouthstick or headpointer has a body-centered viewpoint from which to manipulate, the user of a remote manipulator or robotic device may have to adopt the perspective of the manipulator or robot when issuing movement commands. (This type of problem is encountered in the operation of radio controlled cars. When the car is traveling toward you, the car's movements are opposite those of the control stick, and you must adopt the car's perspective, opposite of your own, as you turn it right or left.)

Other Considerations

Additional factors affecting the use of a manipulation aid include: transportability (an easily transported device can be used in many environments); set-up time and effort (if an individual requires assistance in setting up the manipulation aid, such as donning a hand orthosis or arranging objects within the reach of a remote manipulator, the individual might be more inclined to simply ask the one assisting to perform the intended task); the degree of structure in the environment (regardless of the type of manipulation aid used, the more structured the environment is with respect to the physical abilities of the individual, the more efficient the individual will be within that environment—the properly structured environment itself can be an aid to manipulation); training (manipulation aids, even as seemingly simple as a mouthstick, require time and practice to be used deftly); down-time and repair (an aid which is susceptible to frequent breakdown impedes the acquisition of skill and contributes to frustration and a sense of dependency); use of a powered mobility aid (a mobility aid, such as a powered wheelchair, enables the person to independently and appropriately position him or herself with respect to the objects being manipulated); seating and positioning (stable support of the body is particularly important for efficient use of body-extending manipulation aids).

SUMMARY AND PROPOSAL

Manipulation of the physical environment is a multi-faceted problem. Technological advances are providing new systems and techniques, increasing the options for supplementing manipulation skills. More versatile mouthstick and headpointer systems are becoming available. Traditional orthotic support of the hand and wrist is gaining from advances in lighter weight, more durable materials and in body and externally powered actuators. Functional neuromuscular stimulation of the hand and wrist may eventually complement the use of orthoses. "Universal" environmental control systems, capable of learning the control codes of remotely actuated electronic appliances, are appearing in the commercial sector. Remote manipulators and robotic devices are being developed experimentally and commercially, specifically as aids to assist in manipulation by motorically impaired individuals.

However, no one manipulation aid will be most efficient for all tasks. Just as the individual who has a speech impairment may use, in combination, gestures, signs, vocalizations, a passive letter board, and an electronic communication aid, depending upon the circumstances and nature of the message to be communicated, the individual who has a manipulation impairment should be able to draw from a variety of manipulation aids, selecting the one most appropriate to the physical constraints of the task and degree of personal involvement with the task.

The effective use of these devices depends on careful assessment of the user's abilities, the characteristics of the environments the user will need to manipulate, and the characteristics of the manipulation aids. Psychosocial issues will play a significant role, particularly in the early phase of intervention. The attitude of the individual toward reliance on assistive devices, especially in the case of impairment resulting from traumatic injury, must be taken into account. The appropriate time to encourage the use of a particular device should be chosen in the context of the person's interest in achieving the task. Procedures should be developed to respond to changes in the individual's attitude over time and experience.

In short, the successful supplementation of manipulation skills can be best achieved through a synthesis of intervention strategies, drawing upon technical advances and a sensitivity to the individual's self-concept and manipulation needs.

Language reflects thinking, but also influences thinking. The term "augmentative communication" defined and encouraged an approach to communication in terms of its function and how the function of communication might be facilitated by a synthesis of strategies and assistive devices. To foster like thinking with respect to the function of manipulation and the synthesis of a variety of approaches to achieving manipulation, I suggest the term "augmentative manipulation". The focus of augmentative manipulation being not on the hardware, but on the actor and the act, and on the facilitation of action by a set of complementary manipulation aids and manipulation strategies.

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DEVELOPMENT AND USE OF A ROBOTIC ARM SYSTEM WITH VERY YOUNG, DEVELOPMENTALLY DELAYED CHILDREN

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Abstract

A robotic arm system has been developed and used to foster environmental interaction in very young developkentally delayed children. The system includes hardware and software for training and playing back movements, and collecting and analyzing data. Children as young as 5-6 mo. (MA) will use the arm as a tool.

Introduction

Handicapped children often display a lack of responsiveness to their environment. This is the result of an inability to interact with the environment rather than an inherent characteristic of the individual. It may be avoided if early intervention takes place.

Cognitive development is dependent on early physical interaction with the environment. Sophisticated cognitive and language abilities are all dependent on early consistent interactions with both people and objects. Physically disabled infants frequently cannot interact with their environment in a meaningful manner. Consequently, they may not develop cognitive and social skills and be responsive, interactive individuals.

The goal of this project is to provide a computer- controlled manipulative system which will increase the control disabled children have over objects and social events in their environment.

Robot System Hardware

We have carried out a preliminary study using a robotic arm (the MiniMover-5,(1) with a group of developmentally delayed and able-bodied young children. The anatomy of the robot arm consists of 5 main structures: a stationary base, a body, an upper arm, a forearm, and a 2-fingered gripper. Not only is this robot arm anthropomorphic in its structure (1/2 adult human scale), it also moves like a human arm in its articulation. The arm can rotate at its base, extend and flex at both its shoulder and elbow, pitch and roll at the wrist, and, open and close at its gripper (hand).

Two ways of teaching the robot to execute a desired movement are included in the system: teaching-by-text (the operator uses textual commands) and teaching-through-guidance (the operator uses a guidance unit to lead the robot along the desired path). During the guidance, the path data are stored so that the arm can later play back the movement.

The guidance unit is a 4"x4"x6" box that houses a joystick, 4-buttons, and interface circuitry. The joystick is a four DOF discrete- output device. Movements include: forward/backward, left/right, up/down, & clockwise-twist/anti- clockwise-twist. The first three pairs of joystick activations will move the robot arm in the direction of activation. Twisting will cause the gripper to close/open. Pitch and roll of the gripper are controlled by of two pairs of buttons at the four corners of the unit. A tool coordinate system is employed.

Robot Arm Control Software

We have chosen FORTH (2) for this development since it has many of the desireable characteristics of a robotic programming language (3). These include extensibility, structured language techniques, unrestricted variable names, English-like syntax, and extensive subroutine nesting. FORTH is also able to manipulate transform equations, and it can be interpreted as well as compiled. It is also faster than compiled BASIC. The most important characteristic of FORTH is its extensibility. For example, the implementation at present can allow one to define actions such as REACH and PICKUP:

```
: REACH 200 FORWARD 150 DOWN ;
: PICKUP CLOSE-GRIPPER 100 UP ;
```

When REACH is called, either from the keyboard or within a program, the arm will move 2 inches forward, and then 1.5 inches down. When PICKUP is called, the gripper will close, followed by the raising of the arm for 1 inch. Also, a new word REACH-AND-GRAB can be constructed as a combination of previously defined words:

```
: REACH-AND-GRAB REACH PICKUP ;
```

When REACH-AND-GRAB is called, the entire movement described above will be executed by the arm. The following is a list of currently available 1-D movement directives and their syntax.

Forth Word	Syntax
FORWARD	n FORWARD
BACKWARD	n BACKWARD
UP	n UP
DOWN	n DOWN

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LEFT	n LEFT	
RIGHT	n RIGHT	
FLIP-UP	m FLIP-UP	(+P)
FLIP-DOWN	m FLIP-DOWN	(-P)
ROT-CW	m ROT-CW	(+R)
ROT-CCW	m ROT-CCW	(-R)
OPEN	n OPEN	
CLOSE	n OPEN	
CLOSE-GRIPPER	CLOSE-GRIPPER	

n=inchesX100 m=degrees P=pitch R=roll For example, to move the arm left 2.5 in. n=250. and the correct command is: 250 LEFT.

Our system allows a therapist to teach the robot arm to execute movements (the trajectory through which the arm traces, including both the displacement and orientation of the gripper) which are relevant to an individual child. These movements may then be stored and recalled later in appropriate situations.

Based on a careful evaluation (4) we have found this system to be easy for a therapist to use, and also capable of producing movements which the therapist feels are appropriate to a child.

The robotic arm control software of the system consists of the following principle modules: the guidance module which makes possible teaching through guidance, the edit module which allows editing of a movement after it has been taught, the playback module which replays a taught movement, a management module which allows parameter changes, and a directory module which allows changes between system functions such as training the arm, playing back a movement, recording data or using other contigencies such as toys or graphics.

The teach-by-guidance method is implemented by the guidance unit together with the guidance module written in FORTH. This guidance module repeats the following cycle until terminated by the ESC key: it first reads and translates the input, calculates the robot parameters (backward transformation) and stores them into a data array. Finally, the motor driver is called to move the arm. Within the guidance mode, the therapist can use the arrow keys on the Apple computer keyboard to trace back and re-teach that portion of the movement. In tracing operations, a forward trigonometric transformation is used to convert the vector in joint coordinates into a vector in tool coordinates so that updating of tool coordinates is possible.

A movement editor is also included so that a previously taught movement can be modified and given a new name. This is useful when a slight variation (e.g. opening the gripper a little more for a larger object) of an original movement needs to be made. With this editor, a segment of a movement can be deleted, replaced, appended, or inserted. The playback module replays a taught movement when activated from the keyboard (for therapist) or from the baby-switch (for the infant). The baby playback routine allows interaction between the baby and the system, thus playing a central role in contingency intervention programs. At present, the switch activation can be set in a one-hit or a continous mode, each controlled by different software routines. In the one-hit mode, an assembly routine QUIK-MOVE-ARM plays back the entire movement from the beginning to the end without interruptions. In the continous mode, a routine written in FORTH reads the switch and moves to the next position if it is activated.

The management module allows the therapist to change parameters of interest. These parameters include the name of a movement, its speed, the increment size, and the activation mode of the baby switch.

The directory module handles the saving and retrieving of data between the disk and the RAM memory. To the user, movements are indexed by their names and this module is responsible for keeping track of what movements exist on the diskette, and where they are located.

Experimental Control Software

Our experimental system also has provision for labeling a button or key as corresponding to each of the observable (but not directly detectable) behaviors. The observer will press the appropriate key when a behavior is noted (e.g. directing eye gaze to the object being controlled or to the screen, expressing fear, interest, boredom, etc.). The computer also has a clock, and it automatically records the time of occurence along with the coded behavior. At the end of an experimental session, the computer can be used to combine the manually entered behaviors with those directly sensed (e.g. switch activations, robotic arm movement actions controlled) and display the matrix of behaviors. A hard-copy of these data may be obtained using a printer.

The test control module coordinates all the functions pertinent to robot arm movement, switch detection, and data storage and processing for an experimental session with each subject. Choices of visual graphics or toys (including a tape recorder) as alternatives to robotic arm movement are also included. The switch detection module provides a software interface to the switches connected to the system. switches. After configuration by the test control module, the switch detection module monitors switch closure times and durations.

The data collection and processing module stores the data format and raw data to disk for later retrieval. Running averages and totals for the data can be accessed and displayed for such items as switch closures, switch closure durations, and patterns of switch activation and observed behaviors during the course of the interaction with the subject.

The data display module accesses the data stored on disk following data collection and allows the preparation of a variety of reports. These reports may take the form of statistical analyses and plots. Either video or printed data formats may be used.

Evaluation with Young Children

The present robotic arm system has been used in a preliminary study with 7 very young (CA < 36 months) developmentally delayed children in an infant development program and 3 able-bodied chidren matched in chronological age. Based on interview data from the program director and parents, we began with a period of familiarization during which we played with the child, and determined what their typical responses were to things they like, dislike, are fearful of and are bored with. These were coded for data collection using the system. We then used either a battery operated puppy or cassette tape recorder which was activated by a switch. This allowed us to establish whether cause and effect between the switch activation and toy movement was understood and to determine the best anatomic site and switch for each child.

For all these children, the hand/arm was the best anatomic site. Most used the Zygo tread switch and one used the Zygo leaf switch attached to a piece of yarn. The latter switch was used because it was similar to a task in which the child used the yarn to retrieve a toy.

After we had established reliable switch cause and effect, we trained the robotic arm to make movements which the program director and parent thought would be of interest to the child. The child was then presented with the switch and asked to hit the switch to see the arm move. Continuous and single press modes were both used.

We found that the continuous mode, used with a movement which had a payoff to the child, was the most effective. For example, we trained the arm to pick up a cracker which was out of the child's reach and bring it close enough to be taken with their hand. Each of the children demonstrated understanding of the cause and effect between switch activation and arm movement by first looking at the cracker, then looking at the switch and pressing it, then looking back at the arm as it moved. Sometimes they would reach for the cracker before it was close enough to grab. When this happened, they all reactivated the switch to bring the cracker closer, and then attempted to reach it. This sequence continued until they could reach the cracker.

The degree to which this type of interaciton occurred was called the correspondence criterion for interaction. We also determined the degree to which a sequence was repeated (the repeatibility criterion).Based on an analysis of these two criteria, we found that children with a developmental age greater than 5 to 6 mo. would interact with the robotic arm system as a tool.

Conslusion

This preliminary study established that the children were interested in the arm, were not frightened by it, and were bored unless the arm did something "useful" for them. The use of continuous switch activation to complete the movement proved to be valuable in determining if the child understood that the arm would eventually bring the desired object within his/her reach. All of the children and their parents were very interested in the experiments.

Aknowledgements

American Microscan loaned the Mini-Mover 5. Microbot provided the Apple to Minimover interface, and the Placer Infant Development Program (Roseveille, Ca) assisted witht he clinical trials.

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SMALL ROBOT ARM IN THE WORKPLACE TO AID IN THE EMPLOYMENT OF SEVERELY PHYSICALLY DISABLED PERSONS

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ABSTRACT

The Rehabilitation Engineering Center in Wichita, Kansas, has, as its emphasis, research in the area of finding means to place severely physically disabled persons in the workplace through the use of technology. A project to investigate the use of small robotic arms to assist the disabled at work has been underway for three years at the time of this writing. Several types of devices were investigated and evaluated prior to purchase. Two such types representing two distinct operating and programming criteria were purchased and have been in use on the job. Descriptions of the robotic arms, examples of workstations chosen, and relative merits of device types are presented and discussed.

INTRODUCTION

The Rehabilitation Engineering Center (R.E.C.) in Wichita is federally funded by the National Institute of Handicapped Research and is under the sponsorship of the Cerebral Palsy Research Foundation of Kansas, Inc., with a cooperative relationship with the College of Engineering at Wichita State University. Center Industries in Wichita is a company which employs disabled persons in the "real world" environment of non-sheltered employment. Center Industries has been the laboratory at which research activity has been carried out in the area of using small robotic arms to aid the severely physically disabled worker on the job. Considerable research has been carried out in the choosing of robot devices and of identifying suitable workstations to be operated using a robotic arm. Price considerations, of course, were a most significant factor. Considerable information has been gathered and the Center has experienced success to this point in the project.

FIGURE 1: TINNING OPERATION BY HAND

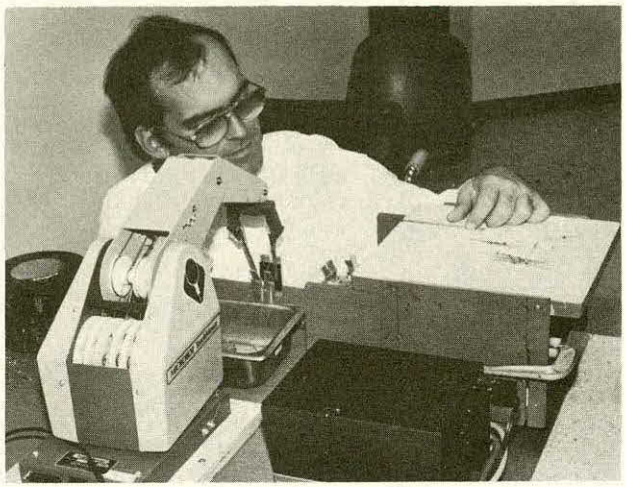

FIGURE 2: TINNING OPERATION USING ROBOTIC ARM BY MICROBOT, INC.[1]

METHODS

Choice of Robotic Arm Devices

The project objectives called for the choice of a small robotic arm in the $2,000 to $4,000 category. It was hoped that if a suitable device could be found in this price range that it would be affordable to the employer and, perhaps, even to the employee as a tool. Several sources were found that offered such devices. However, all devices in the desired price range were intended for classroom use in the teaching of robotics to students and none were considered by their manufacturer to be of industrial quality. That is, they were not intended to be used on a continuous basis on the job. To satisfy budget constraints in the first year of

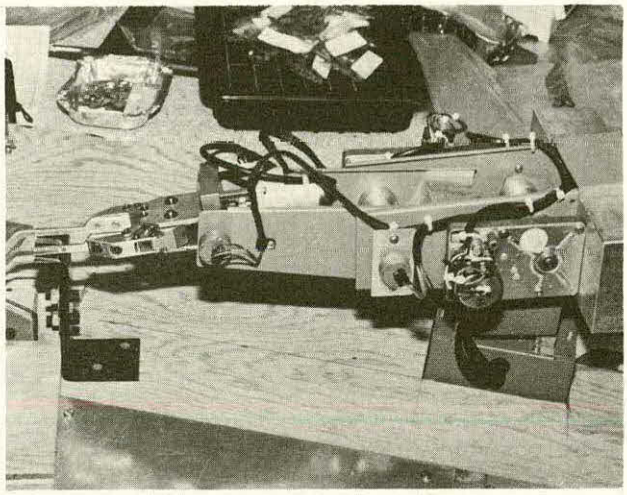

FIGURE 3: "ARMATROL" BY FEEDBACK, INC.[2]

of the five-year project a teaching robotic arm manufactured by Microbot, Inc.[1], was chosen and purchased. The price was less than $3,000 and it can be seen in Figure 2. In the third year of the project a robotic arm of different operating technology and programming procedure was purchased. This device is manufactured by Feedback, Inc.[2], and can be seen in Figure 3.

Choice of Workstations

The purpose of the project was to choose a workstation which would be suitable for robotic application in the area of manipulation of workpieces, but, would also require the support of a worker in the area of quality control inspection and indexing of workpieces for the start of each operating cycle. The worker should also be able to stop the operating cycle should problems arise. It was anticipated that a severely physically disabled person could then work at this station utilizing the robotic arm to perform the precise manipulations of small parts while, at the same time, exercising quality control (inspection) and the indexing of parts for the operating cycles.

A contract with the Boeing Company in Wichita calls for solder tinning of the electrical leads of small components prior to insertion into circuit boards. The operation requires the picking up of the part, dipping the lead in a liquid flux, then dipping the lead in molten solder (holding for three seconds), inspection of the lead for uniform solder coating, placing the part at the side for cooling in air, and, finally, placement of the part in an alcohol bath. The manipulation of these small parts in this operation was difficult, at best, to do for persons with unsure hands. One of these workstations was chosen for the first application of robotic arms in the research. The workstation as performed by an able-bodied worker by hand is shown in Figure 1. The workstation as set up with the Microbot[1] can be seen in Figure 2. Note that considerable support for the workstation in the way of fixtures was required to allow the worker who has cerebral palsy to properly index the parts and perform the other duties of the job. To date, this contract and activity at Center Industries

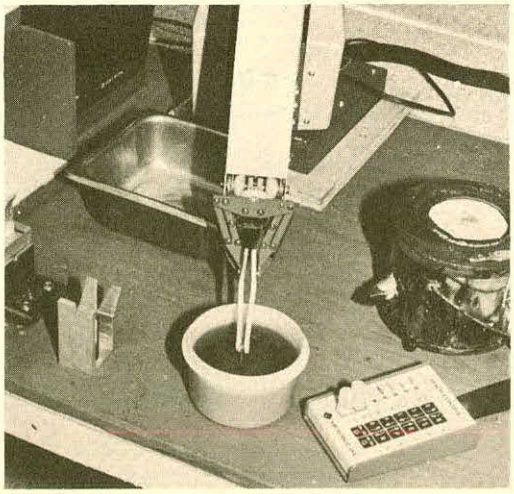

FIGURE 4: PROGRAMMING BY TEACH PENDANT

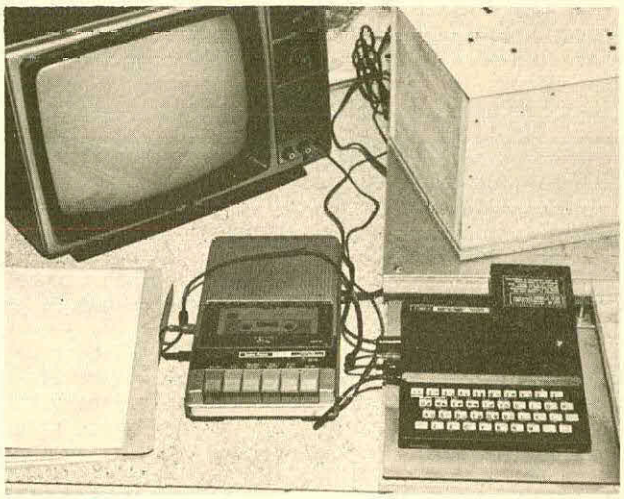

FIGURE 5: PROGRAMMING BY COMPUTER KEYBOARD

has provided the best choice for the opportunity to employ the use of small robotic arms in keeping with the original intent of the project. The robotic arm called the "Armatrol" by Feedback, Inc.[2], can be seen as it picks up the indexed part in Figure 3.

DISCUSSION

All robotic arms do not provide for the same level of articulation and each must be investigated as to particular need on the job. The model by Microbot, Inc.[1], called the "Teachmover" provides for shoulder, elbow, and wrist movements with the additional function of wrist "roll" or twisting of 360° each direction from center position. All of these movements can then be utilized as rotation takes place about the base. The "Armatrol" model by Feedback, Inc.[2], provides for all of the above motions except for "roll" of the wrist, which means that the workpiece cannot be "turned over" in the work cycle. Simple "picking" and "placing" tasks may not require rotation of the "arm" about a base. The research project goals were to investigate devices that provided for the most articulation so that fine motor tasks could be investigated. The workstations that were chosen require the capability of the fully articulated "arm". Therefore, a more thourough investigation of robotic arm capabilities was possible.

Robotic arms can be programmed to function in a repeatable cycle in three basic ways: 1. Use of a teaching pendent - a small console with push buttons to control the arm and enter programmed positions (see lower R.H. corner of Figure 4); 2. Entry of programming data through a computer keyboard requiring separtate computer and monitor (see Figure 5); 3. "Lead-by-the-nose" programming which requires the placing of the robot's "hand" manually and recording each placement by keyboard input. Methods 1 and 2 were utilized by the Microbot Teachmover and the Feedback Armatrol, respectively. Method 3 was not investigated by the project.

Costs

At the time of purchase, the Microbot Teachmover

that was first purchased cost $2,670 including a hard shell carrying case. An additional $870 was invested in a Radio Shack Model 100 computer for the purpose of recording programs for permanence that had been created using the teaching pendant. One year later, a Microbot Teachmover was purchased at a cost of $4,019 which had additional programming capacity plus the added feature of non-volatile memory. That is, the microprocessor would retain a program in its memory after power had been turned off, eliminating the necessity of "dumping" the program from the microprocessor to a cassette tape through the use of the computor.

The Feedback Armatrol was purchased at a cost of $1,890 including a 10% discount from the manufacturer. This purchase price included a Sinclair computer (see Figure 5). A black and white television set was purchased for $65 to use as a monitor (necessary to vilualize computer inputs).

CONCLUSIONS

At the time of this writing the research project has consumed three years of an originally proposed five year time period. Much has been learned and investigated. The following observations have been made by the research staff:

1. The small robotic arm is most definitely a useful tool which can be utilized to place a severely disabled person doing a complicated task.

2. The robotic arm not only is capable of performing fine motor tasks but acts as an excellent "pacer". It forces the worker to be "ready" for the next cycle of operation. Members of the research staff have found "pacing" to be a difficult concept for many to grasp without some external indicator.

3. Programming by the teaching pendant seems to be the preferred method when teaching the worker to program the device for a new cycle of operation. Computer programming of the devices requires knowledge of mathematical criteria which is beyond the majority of disabled blue collar workers.

4. Tasks can be found that require the human element for judgement in addition to the handling of workpieces by the robotic arm, eliminating the possiblity of a "token" employee.

5. A programmed repeatable cycle is the only form of use that a robotic arm can be productive on the job regardless of physical capabilities of the worker. Manipulation of the "arm" manually simply is too time consuming and inaccurate to be reasonable on the job, especially if production rates are required.

6. "Pick" and "place" tasks do not always require the articulation of a full "arm". Highly accurate devices are available with less than whole-arm capability.

7. In the day-to-day operation of a workstation at which a robotic arm is utilized it is most desirable that there be a known, or "stored", "home" position. That is, a position known to the microprocessor in the robot from which all operating steps can be referenced. The Microbot Teachmover does not have such a known position. Robotic technology to date uses two principles of mechanical operation; air or hydraulic pressure coupled with actuating cyclinders and electric motors. The two devices investigated are operated by electric motors. The Teachmover operates using electric stepper motors (the microprocessor simply counts the "steps" through which each of the operating motors rotate). The Armatrol utilizes servo motors coupled with encoders and offers the known "home" position. A prestored program will not operate the robotic arm through steps which must touch locations precisely if the reference point, or "home" position, is not precisely the same for each initial startup (i.e. at beginning of the day of operation).

8. The robotic arms that are sold for educational purposes and are the least expensive ($2,000 to $4,000 and even to $10,000) are not suited for the rigors of day-to-day continuous operation of a workstation. The mechanical and electrical components in these devices simply are not industrially rated for that kind of demand.

The author and other researchers on this project have concluded that the concept of employing severely physically disabled persons by the use of small robotic arms has been shown to be viable.

The Microbot[1] company now offers an industrially rated robotic arm called the "Alpha". The project has recently purchased one of these models (retail value $14,000). Sophistication of operation is at a much more technical level with approximately five times the programming capability. Staff is currently learning operating procedures of this device. It is anticipated that problems that have been encountered in the area of maintenance will be curtailed.

ACKNOWLEDGEMENTS

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FOOTNOTES

1. Microbot, Inc., 453-H Ravendale Drive
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DESIGN OF AN OMNIDIRECTIONAL MOBILE ROBOT AS A MANIPULATION AID FOR THE SEVERELY DISABLED

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ABSTRACT

The design of a robot intended for human-service applications, which is the case for the Rehabilitative Robotic Aid under development at the Palo Alto Veterans Administration and Stanford University, is under different constraints than either an industrial robot or a classic telemanipulator system. Rather, an **Interactive Robot System** has unique design requirements stemming from the need to perform certain operations autonomously, yet at all times be responsive to human intervention to redirect the progress of a task. This paper will investigate the design issues, report on the current status of the project, and offer our solutions and thoughts about future system enhancements for discussion.

INTRODUCTION

The Veterans Administration/Stanford University Robotic Aid Project (VA/SU RAP) is currently developing a mobile, omnidirectional vehicle with a 6-axis robot arm mounted on it. The three-wheeled base and the arm are outfitted with an array of sensory subsystems to allow the unit to interact in a nondestructive and useful way with the environment it is intended to operate in, that is to say a home or clinic where severely disabled individuals perceive the same needs to manipulate their personal space as the able-bodied, but would require human assistance to accomplish them. The RAP project seeks to develop a tool that can be commanded and controlled directly, by voice, by head-motion, and by residual limb movement, to perform certain typical manipulation tasks in such situations, and thus restore a significant measure of independence and privacy to the person who wishes to express himself thus.

BACKGROUND

The process leading to the design described below was initiated in 1979, with the start of the first phase of the VA/SU RAP. The four years of this phase saw the development of a tabletop Unimation PUMA-250 arm, enhanced by a host processor managing voice I/O units, diagnostic displays, a 6-axis joystick, and the robot arm itself[4]. The programming of the real-time aspects of the control of the arm was instrumental in guiding current software work[1].

Over the past two years of this phase, the system underwent clinical trials, during which the system evolved in response to exposure to quadriplegic users at the Palo Alto VA's Spinal Cord Injury (SCI) Center. Additionally, a user and training manual was produced[3], and studies were performed to assess the success and viability of the voice-recognition unit as a control device[8]. This effort, in conjunction with continuing technological development, has led to the current phase of work[5], which centers on the implementation of an omnidirectional mobile base, on the design of sensor systems, and on more sophisticated software to control the device. Within several months, we will implement comprehensive performance assessment procedures with members of the intended consumer population at the SCI Center. This paper will concentrate on the design criteria and sensor system development.

DESIGN ISSUES, CRITERIA, AND CHOICES

While proceeding through the various aspects of the design, one criterion should be reiterated: this is an interactive, semi-autonomous device that moves about human environments and is not only controlled by, but also in close proximity to, its principal inhabitants.

Mobile Base and Drive Train

The mobile base is intended to exist in a home or clinic environment, and must not itself be handicapped in maneuvering in such a space. A wheelchair, hampered by the inability to crab sideways, is at a major disadvantage in places such as elevators and kitchens. An omnidirectional design by W. La[3] has been implemented for our vehicle. It consists of three motorized wheels mounted at 120°, each carrying a circumferential ring of rollers [see Figure 1]. The fact that the wheels shafts are fixed to the base greatly simplifies the turning problem, in that there is no steering gear per se: the rollers, instrinsic to the wheel design, allow for any combination of moving and turning. To move in a certain direction with a certain rotation requires only that the three wheels be driven by their motors in a prescribed manner.

This configuration represents the elegant minimum of wheels and motors for true omnidirectional locomotion. In addition, no suspension is required to keep all the wheels on the ground, as would be the case with four or more. The stability of the system is not significantly compromised, with the tip-over angle being a reasonable 24°. This is due to the low center of gravity (10" from the ground), achieved by mounting the batteries and the drivetrain components as close to the ground as possible, and installing the relatively lightweight computer equipment in the upper compartment of the base.

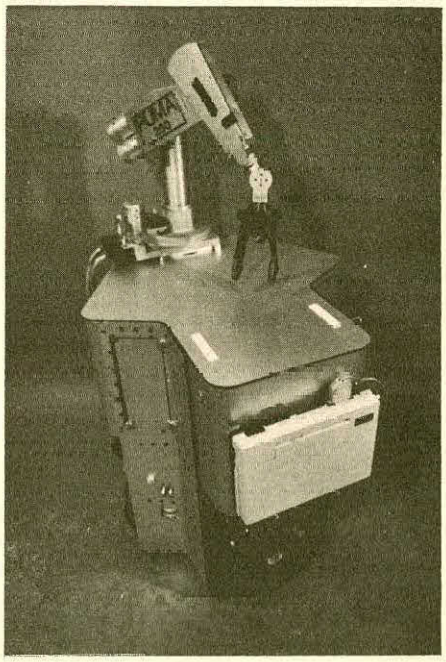

FIGURE 1. Mobile Base System.
A view of the three-wheeled mobile system with the PUMA-260 arm and the sensate hand. The base carries its own status display, shown in stowed position.

The low number of drive components serves to reduce the size of the base overall, and reduce the power requirements. The base has a 24" diameter and a 30" height (without the arm). The weight of 350 lbs. is due in large part to batteries (150 lbs.), but is consistent with the weight of an electric wheelchair with its occupant. Battery power is sufficient for three to six hours of autonomy, depending on the duty cycle.

The $\frac{1}{4}$HP motors assure a 1.5 mph maximum speed and a 12° climbing angle, sufficient to be compatible with human walking speeds and typical wheelchair ramps.

Robotic Arm

The arm we are using for this generation system is a Unimation PUMA-260 augmented with a force wrist and sensate hand. The arm will also acquire a "torso," or vertical and outward reach extender, to permit objects on the floor, on high shelves, and at the back of counters to be accessed. It was considered desirable to use a small arm, augmented in its reach, rather than a large fixed one, for reasons of safety and user acceptance. The manipulation of typical household objects rarely requires more than 5 pounds of force to be applied by the hand, robotic or human, so a larger and stronger robot arm was not mandated for this application.

Gripper and Force Sensor

The development of a smart sensate hand has been an area of research for several years[2,6], and the current design includes a set of 12 optical proximity sensors mounted on a one-motor parallel jaw gripper. Sensor readings are corrected in software for ambient lighting, and work is proceeding on the elimination of the effect of surface reflectivity. Algorithms are being established for automatic object centering and grasping and obstacle avoidance. The hand itself has a 10 lb. maximum grip, and is instrumented as to grip force and finger position.

The force sensor measures all six force and torque components of the load applied at the hand. The sensor is fitted into the wrist of the arm, and is optimized for the PUMA-260 and the hand being used. It will allow such procedures as operating appliance pushbuttons, turning doorknobs, and obstacle and collision detection.

CURRENT STATUS OF RESEARCH

At this point, the mobile base and the arm are fully operational, with the lowest level of control implemented. The on-board computer has access to all the relevant system states, and is programmed to transmit these to the control workstation on command. Omnidirectional motion has been implemented, as well as joint-interpolated and straight-line motion of the arm.

The software to accomplish the integration of the sensor signals, the commands from the host computer, and the outputs to the motor controllers themselves was written in DEC's MicroPowerPASCAL© by S. Michalowski, and is described in a companion paper in these Proceedings.

A first safety system has been installed, that being an instrumented 2-mph bumper. The bumper localizes the impact, and the controller can decide, for example, whether to push harder (to open a door), or to retreat (in the case of an unknown obstacle).

FUTURE DEVELOPMENTS

The design of the first base is being followed by the construction of four more units, all derivatives of the first one, but exhibiting newer technologies in terms of motor, drivetrain, and power amps. These units will serve not only as back-ups, but also as additional systems for path-planning, navigation, and sensor subsystem development. Below is a discussion of some technology-related projects currently in progress.

Ultrasonic Obstacle Avoidance System. An ultrasonic scanning system is being developed to establish a polar plot of the range from the base to the nearest obstacle. The system is expected to have a range of several inches to 50 feet.

Scanning Laser Absolute Ranging System. This system will employ fixed reflectors mounted at known locations in the mobile base's environment to extract absolute position information by triangulation.

Two-axis arm support. A "torso" is to be designed, as previously mentioned, to extend the manipulation range of the arm. As yet, this feature has not been implemented.

Docking fixture. A "live" dock will be designed to provide the base with a fixed position reference, as well as a battery charger and a direct link to the stationary computer.

CONCLUSION

This paper has concentrated on the design aspects of the system. Applications, computer architecture, and software design are discussed in companion papers presented at this conference.

The system under development promises to be a powerful test bed for research into the use of robotic technology as a mechanical manipulation aid for the severely disabled. The project has proven the feasibility of the technology, and now pushes on to establish its usefulness.

ACKNOWLEDGEMENTS

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Robot Piloting Techniques for Working in Unstructured Environments

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ABSTRACT - Although numerous projects have shown the feasibility of using robotic manipulators to aid the disabled, the tasks demonstrated have generally been preprogrammed and/or highly structured. A truly versatile manipulation aid for the disabled must permit the execution of novel tasks, i.e., those that have not been preprogrammed. Several strategies for piloting a robot using a three degree of freedom (d.o.f.) joystick have been designed and implemented on a laboratory system. Data collected in an evaluation of these piloting strategies with 26 able-bodied subjects are presented and discussed.

INTRODUCTION:

Rehabilitative Robotic Aids are remote manipulators designed to perform preprogrammed and interactive manipulative tasks for a disabled person, enabling them to affect their environment. The feasibility of such devices has been shown in numerous research and demonstration projects over the past few years [Corker et al., 1979; Leifer, 1981; Leblanc and Leifer, 1982]. Although rehabilitative robotics hold considerable promise for increased independence of disabled people, the utility of current robotic aids is extremely limited.

Most of these previously mentioned demonstrations concentrated on programs consisting of preprogrammed routines with simple control structures. While preprogrammed sequences are useful to carry out known tasks in a structured environment, they are unable to perform correctly in the presence of variability unanticipated at the time of programming. Artificial intelligence and sensing may allow the robot to adapt to slight changes in the workspace, but in general the user spends their time supervising the robot. The disabled user may provide basic flow control for the application program or interrupt the program in the event of gross changes that make it impossible for the robot to complete the task.

Interactive sequences, wherein a person controls the manipulator directly, make possible the execution of novel tasks in an unstructured environment. Candidate tasks for this type of control include correcting the robot during a preprogrammed task, teaching the robot new tasks (robot guiding), performing tasks that are too variable to be preprogrammed, and being creative (e.g., painting or drawing). Interactive control of a robotic aid will hereafter be referred to as piloting [Guittet et al., 1979; Van der Loos, 1983].

Robot piloting involves the use of discrete (e.g., switches) or proportional (e.g., potentiometer) inputs to control the robot arm in real time. These inputs are combined or multiplexed to afford the user control over all of the degrees of freedom of the manipulator. When a robot is being piloted by an able-bodied individual, it is often possible to utilize the motor abilities of the operator to control each joint simultaneously. This practice, referred to as telemanipulation, is often used to handle materials in hazardous environments. In the case of disabled operators, however, there are seldom enough control sites available for this method of control. In these cases a strategy must be devised for controlling the robot using the motoric input that the disabled person is capable of performing.

Six degrees of freedom must be specified to describe the position and orientation of a manipulator's end effector [Paul, 1981]. In cartesian coordinates the three primary d.o.f. are left/right (X), forward/backward (Y), and up/down (Z). When using this coordinate system to describe robot motion these three variables may describe the position of the wrist in space. Three additional d.o.f., wrist pitch, wrist yaw, and wrist roll are used to orient the hand. Although additional input may be necessary to specify how a robot having redundant (i.e., greater than six) d.o.f. is to reach a location, these six d.o.f. are sufficient to describe the orientation of the gripper in any location in the workspace.

For a disabled person, piloting presents several difficulties. First, there are frequently more outputs to be controlled than inputs available from the individual and some of these outputs may need to be controlled simultaneously. Second is that the coordinate system in which the robot is operating is separate from the user. The individual controlling the robot must continuously change his/her perspective to match that of the robot. For instance, when the robot is facing towards the user their left and right are opposite those of the robot.

A piloting strategy was desired that would remedy these deficiencies. One goal was to reduce the total amount of input needed to control the robot arm. Another was to have the inputs structured such that the user is not required to constantly shift their perspective in order to relate their inputs to the desired output.

A new piloting strategy devised for these purposes will be called **Pointing**. The pointing strategy segments the piloting task into functional groups in an attempt to achieve a more intuitive user interface. Pointing was implemented as a module of a program called PILOT. This program combines three piloting methods, pointing, tool coordinate and base coordinate, into an assessment tool that simultaneously pilots a robot using a chosen strategy and collects data on the performance of the operator. PILOT was used with 26 individuals to evaluate these three piloting systems.

IMPLEMENTATION:

The robot used in the laboratory evaluations is a five d.o.f. articulated geometry [Snyder, 1985] robot arm manufactured by Rhino Robots, Inc. [Shandu, 1982]. The Rhino lacks wrist yaw, a d.o.f. normally present on a robot of this geometry. The arm hangs inverted from a two d.o.f. slide base. This overhead gantry system gives the robot a total of seven d.o.f. and increases the amount of unobstructed workspace available for manipulation. The robot gripper has been modified with two high-output LEDs and focusing lenses mounted in the fingertips. The light from these LEDs are focused to indicate the direction that the hand is pointing.

The pilot control software is written in the C programming language and runs on an IBM AT™. The program communicates with the Rhino XR-2 controller through an RS-232 interface using the Rhino command format. The program may be adapted to drive similar robots (e.g., the Scorbot ER-III) by changing the input and output format.

The inputs used to control the robot in the evaluations were a three-dimensional proportional input and two switches. The proportional inputs consist of a two-dimensional joystick mounted on a one-dimensional slide base. The switches used are the pushbuttons present on the joystick. The use of standard peripheral devices allowed the software to read the inputs from a standard game adapter.

The Base and Tool Coordinate systems are cartesian coordinate systems located at the robot wrist, and the robot gripper respectively. Three d.o.f. of the joystick control robot motion Up/Down, Left/Right and Forwards/Backwards. By depressing one pushbutton, control is transferred to the joystick to control the wrist pitch and wrist yaw to orient the hand. Since the Rhino Robot is not equipped with wrist yaw, it was accomplished with a combination of waist rotation and motion of the two slide base axes. In this case the entire robot rotates about the wrist, producing the same result as the missing d.o.f.

For the pointing method, two of the three degrees of freedom of the joystick allow the user to aim the spot light sources in the fingertips at the location in the workspace that they desire the robot to move. The third d.o.f. of the joystick controls robot motion Forward/Backward along the axis of the gripper and beam of light. Once at the desired location, the user depresses the pushbutton. Now, two degrees of freedom of the joystick allow the user to orient the hand by pitching the wrist and yawing the robot about a point one inch in front of the fingertips, rather than at the wrist joint. The second pushbutton sends the robot to a home position which serves as a convenient place from which to point at a new location.

The piloting program gives the user real-time, continuous velocity control of the robot arm and joints. Differential inverse kinematic equations are used in the program to calculate the rotational or linear output of each robot joint based on the continuous stream of incremental differential positional commands from the joystick input [Crochetiere, 1982; Brady, 1983].

The piloting system uses an auxiliary clock to set a constant cycle time of 133 milliseconds. The piloting program integrates the differential positional commands for this cycle time and calculates the robot position for the end of the cycle. This process has the effect of simulating velocity control of the robot despite its true incremental operation.

The software monitors the position of each joint of the robot. A seven degree of freedom robot has a redundant degree of freedom. The extra d.o.f. of the robot is not used to make approaches to objects more flexible, rather, it is used to extend the range of motion. The software incorporates an algorithm to choose which joints to use to accomplish a motion so not to overextend joints or run into its mechanical limits. An example of this is when the arm is moving forward and becomes fully extended. At that point the slide-base takes over and maintains the forward motion.

EVALUATION:

The three piloting systems were tested on 26 able-bodied subjects to evaluate their performance piloting the robot through an obstacle course. Nine people tested base coordinate piloting, nine people tested pointing piloting, and eight people tested tool coordinate piloting.

The course was constructed on a three foot square table erected in the robot's workspace. Wooden obstacles were arrayed on this table in a set configuration of towers, pyramids and patterns. The block faces were marked to indicate a sequence of landing locations to which the subject was to pilot the robot arm. Once near the block face, the subject was required to touch the block with the robot fingertips normal to the face. The 22 block faces that the subject was required to navigate to were set at various heights and angles.

Each subject was tested independently using a single piloting system. After a short training session, the subject was asked to maneuver through the course twice. Time, distance traveled, overshoot, and joystick usage were monitored by the program and recorded.

RESULTS AND DISCUSSION:

The results presented in Table 1. are the averages over all the subjects in a given group. Results for the first trial and second trial for the course are listed separately to show the effect of learning over the two trials. The results presented represent:

1. The total time required for the subject to complete each of the trials.

2. The amount of time the subject spent <u>not</u> piloting (may be related to an initial cognitive load).

	POINTING SYSTEM		BASE COORDINATE		TOOL COORDINATE	
	TRIAL 1	TRIAL 2	TRIAL 1	TRIAL 2	TRIAL 1	TRIAL 2
TOTAL TIME	877.8 s	681.8 s	895.9 s	733.5 s	1050.9 s	824.6 s
IDLE TIME	204.5 s	121.5 s	206.7 s	135.3 s	249.8 s	146.3 s
ERROR X ARM	68.6 in	66.2 in	155.1 in	151.6 in	107.5 in	111.7 in
ERROR Z ARM	84.1 in	93.2 in	72.7 in	76.0 in	54.5 in	81.7 in
ERROR X BASE	62.4 in	4.3 in	73.1 in	65.4 in	66.6 in	51.2 in
ERROR Y BASE	24.4 in	11.3 in	35.9 in	29.5 in	43.6 in	40.1 in
CONTROL AXES USED	1.32	1.37	1.38	1.36	1.24	1.32
MODE CHANGES	71.9	54.0	81.4	62.0	81.8	57.6

Table 1. Data Summary from Piloting Evaluation

3. Error based on how much further the robot was piloted than was necessary to navigate the course.

4. The average number of joystick axes a person used simultaneously (only recorded during the time that they were piloting).

5. Pushbutton hits to change modes from positioning to hand orientation.

While a complete analysis of the data has not been performed, a cursory examination suggests that pointing is in some ways a superior strategy for robot piloting. Pointing appears to require less time and produce less error of certain types, especially those that result in the manipulator travelling a greater distance than necessary. Extended trials with subjects practicing all three systems showed that these observed advantages persist through the learning process.

An interesting observation can be made concerning the usage of multiple d.o.f. input devices. Subjects seldom simultaneously used the inputs available. The average joystick use during the evaluations was approximately 1.3 simultaneous axes. This suggests that the control required for piloting may be well suited to single inputs generally available from severely disabled clients and voice control systems [Sachs and Leifer, 1979].

CONCLUSION:

The Pointing system for robot piloting is attractive for providing real-time manipulator control for the disabled. It requires a type of input that is well suited for sequential presentation to disabled individuals having severe functional impairment. Initial testing suggests that task time is lower using piloting and that path and destination error is reduced. The experimental apparatus and software will be used to implement tasks requiring interactive sequences to gain additional experience and collect more data during the execution of actual tasks.

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EVALUATION OF THE APL/JHU ROBOT ARM WORK STATION*

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Abstract

A robot arm/work table system has been designed at APL/JHU and evaluated at VA Medical Centers as an assistive device for the high spinal cord injured person. The goal of this system is to provide some level of independence to the disabled individual who possesses little or no manipulative function. The robot arm is a 6 degree-of-freedom device which is controlled by a low cost computer with both individual joint control and up to 48 preprogrammed motion sequences. This paper will describe the overall system and discuss alternative control input modes for the highly disabled person. Controllability and safety are two important issues which have commanded a high priority in the design. Testing has been conducted with over 20 volunteer quadriplegics and results of this clinical evaluation will be highlighted.

Background

The APL/JHU robot arm/work table system has been designed to enable high level quadriplegics to execute certain manipulative tasks with little or no attendant assistance. This research program, has now reached a stage where a manufacturing prototype has been fabricated and is in the final stages of evaluation. This system is designed to provide manipulative capability for a variety of tasks for the high level quadriplegic. Such individuals, with little or no upper and lower limb functions represent one of the highest needs in rehabilitation engineering.

This work station has been designed with the specific goal of meeting total task accomplishment in complete safety and with little or no attendant assistance. The latest model incorporates a low cost computer (6809 microprocessor) with adequate flexibility to permit programming a number of useful manipulative tasks on a structured work station. The robot arm is a 6-degrees- of-freedom, computer controlled anthropomorphic limb. The individual degrees of freedom may (if the user so elects) be directly controlled by selection of the desired joint to be moved. Alternatively, the user may select preprogrammed motions to perform tasks where such structured trajectories can accomplish a task. Picking up a book and placing it in a reading stand or adjusting printer controls are examples of task compatibility as shown in Figure 1.

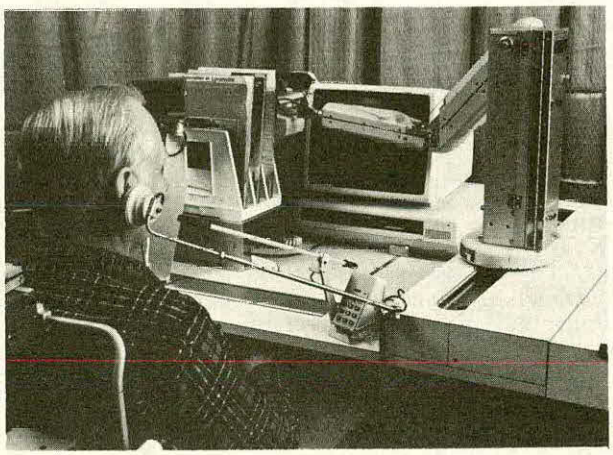

Figure 1
Workstation with Reading Materials and Computer Setup

Input Modes to the Robot

Throughout the course of this research program many control input methods were examined as possible input modes to the system. Safety in operation and ease of user control were important goals of the input device. Chin motion input was selected after evaluation of alternatives because of its positive control and good resolution capability. The chin motion sensor may be a dual purpose controller as part of the wheelchair controller or may be a stand-alone device mounted on the workstation. Small up and down chin motion provides proportional control of selection of individual joint motion or prestored motion sequences while a single pulse activated by slight rocking fore and aft causes an event to start or stop. For systems employing the dual mode wheelchair controller, steering control of the wheelchair is achieved by lateral motion of the chin controller. Torque of the motors, hence wheelchair motion, is controlled by how far the controller is depressed. The chin control apparatus is very small and obscures very little of the face. Once the user maneuvers in front of the workstation, an optical link transfers signals to the robot arm and the chin controller is now used to input the robot arm.

Individual users who have their own wheelchair controllers and choose not to use the APL/JHU dual mode wheelchair chin controllers may utilize a stand alone chin controller. This table mounted controller is operated in an identical manner as the wheelchair controller. For those individuals who can not input a chin controller, a sip & puff controller is available. Thus, the APL system addresses the needs of a wide range of users by providing alternative control methods of inputting the system.

The Workstation Concept

The workstation concept is based on placing components in fixed location on the work table such that the robot arm may use manual step-by-step motion or prestored computer controlled motion trajectories to carry out the desired functions. The system not only provides for basic manipulative functions within its 6 degrees of freedom motion, it has demonstrated that complex functional tasks such as self-feeding, handling a variety of reading materials, use of a telephone, and a computer system may be accomplished with minimal attendant assistance.

As an alternative to direct control of any single axis-of-motion, the user may select one of many prestored trajectories to accomplish a specific task. The following tasks, which are only a partial list, use prestored application programs in the current model of this robot system:

1. Move mouthstick or morse code keyer into position for use
2. Pick up the telephone and earphone hold it next to one's ear
3. Hang up the telephone
4. Pick up Kleenex tissue
5. Eat sandwhich from a plate

6. Eat with spoon in plate; and
7. Eat from a bowl
8. Turn the computer off-on
9. Input the computer printer controls for draft or letter quality print

An important tool used in conjunction with the robot arm is a mouthstick. Manipulative functions such as putting a magazine in place for reading are accomplished by the robot, which page turning is accomplished by use of the mouthstick.

The work station may be configured for the task environmental needs of individual users. A typical desk-work station layout was shown in Figure 1. Prestored trajectories may be readily preprogrammed to accomplish new tasks as needed. The programming keyboard is designed to be used by either the therapist or an experienced quadriplegic user. The functional keys define BASIC-like language element for easy specification of motion sequences. Commands exist for motion to a point, stimulation and sensing of external devices, jump to other motion sequences, and more.

Clinical Evaluation

An important tool in the evolution of a practical system design is the conduct of clinical testing with quadriplegic volunteers. Conceptual ideas which look good on paper may often be discarded after realistic day-to-day testing. Clinical testing can contribute much to overall system design, since it exercises the system under realistic environment. Since January 1983, two experimental robot arm work stations have been included in part time evaluation testing at VA Spinal Cord Injury Centers at Richmond,Virginia, Cleveland, Ohio, and Hines, Illinois. These tests have shown that the system concept has merit for certain highly disabled persons yet still needs some "fine tuning" make the system of practical value(1).

Involved in the clinical evaluation were 20 male quadriplegics between 21 and 60 years of age. Their levels of injury ranged from C-2 to C-5. Individual accumulation of time working with the equipment ranged from 1 hour to over 100 hours; Over 300 meals have been eaten by these individuals using the robot arm. The self feeding arrangement is being demonstrated by the design engineer in Figure 2.

Figure 2
Workstation Self-Feeding Arrangement

About one half of these individuals found the equipment gratifying to use, especially for self feeding There were no safety problems encountered throughout the test program. The most frequent and significant problems were:

1. Incompatibility of the system with a reclining user;
2. Inadequacy of this early version of the chin controller to compensate for posture changes in a wheelchair; and
3. Incompatibility of the work station with other wheelchair controllers

Since these clinical tests were conducted, a new chin controller with self adjusting features was designed to compensate for the patients' change in position during the day. A table mounted controller option was added to the system to allow the use of a broader range of existing wheelchairs. Additional tests are being conducted at a VA Medical Center to further evaluate these changes.

Manufacturing and Producibility

The APL robot arm work table system was selected by the VA for transition to a manufacturing prototype model. Such a model has been designed, constructed, and is undergoing acceptance evaluation testing. Upon completion of these tests, it is expected a small number of units will be ordered and placed in VA Medical Centers for long term utilization by quadriplegic patients in these centers.

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DEVELOPMENT OF A COMPUTER SYSTEM FOR ASSESSING LOW BACK PAIN DISABILITY

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ABSTRACT

A new method of collecting a comprehensive data base on clinic patients involves the use of a computerized questionnaire, answered directly by the patient. Information gathered in this manner can be used to tailor treatment and rehabilitation to individual patients. An approach to determining what information constitutes a useful comprehensive data base is presented, as well as several design criteria which should be considered in the development of computer data-gathering systems. Although these suggestions are offered in the specific context of low back pain disability, they are also applicable to the study of other disabling diseases.

INTRODUCTION

Low back pain (LBP) affects from forty to eighty percent of individuals at some time during their lives (5) and it has been estimated that fifty percent of these will experience recurring problems with the lower back (6). For some, low back pain is a minor, isolated inconvenience, having little effect on their daily lives. For others, the pain may be severe enough to impair their ability to work and carry on usual daily functions.

Because LBP is not a visible handicap, it is not easily measured. Furthermore, the causes of LBP are multifaceted, and consequently the course of LBP is unpredictable. However, the seriousness and pervasiveness of LBP becomes evident when one considers that there are currently 11.7 million LBP-impaired and 5.3 million LBP-disabled individuals in the United States (3). Evaluating LBP disability is confounded not only by the absence of any visible physical anomalies, but also by the many factors which influence it. Furthermore, several levels of LBP disability exist, and either disability or pain may vary in severity during the course of one's life.

The Vermont Rehabilitation Engineering Center is addressing this growing socioeconomic problem by studying which of the myriad of factors known to be important in the course of LBP are associated with disability. Our goal is to identify those factors most reliable in predicting which individuals with LBP will become disabled by their pain, and to use this information to apply appropriate treatments.

In order to achieve this goal, we have designed a microcomputer-based system which collects a comprehensive data base on a large number of patients and, given such information, predicts the likelihood of disability for each patient. The information gathered can be valuable to both treatment and rehabilitation. Additionally, the information from all patients can be studied collectively to predict trends or answer questions of a statistical nature pertaining to the etiology of LBP.

Computers have traditionally been used for data manipulation and analysis, as well as for programming and directing machinery or devices. Today, computers are being used more frequently in clinical settings to collect and record essential information about patients. Often a clinician questions the patient and records the responses on the computer (2), or operates a computer to record biological data, for example, recording EMG to measure muscle strength and endurance (4). Developments in non-keyboard based interaction (e.g., the touchscreen and the mouse) allow for further time-saving applications. Thus, in a clinical setting, patients can be questioned directly by the computer. This application has several advantages over more traditional methods of data collection. Computer-administered questionnaires free up clinician time because the computer asks the questions, records the answers and summarizes the information. Also, the computer provides a sense of confidentiality. In this setting, patients are likely to answer questions of a personal or sensitive nature more honestly and without embarrassment.

METHODOLOGY

The computerized data collection system passed through three development phases before it was ready for clinical use (Figure 1). In the first phase, the content of the questionnaire was established. A list of relevant factors was derived through an extensive literature review and through discussions of a panel of experts in the field of low back pain. Each such factor represents a patient characteristic believed to affect LBP disability level, for example, psychological profile or number of previous hospitalizations. The second phase involved developing a questionnaire to obtain accurate information about each factor. For some factors, standard questionnaires or measurement tools were adopted; when standard tools were not available, appropriate questions were written to gather the necessary information. The third phase involved adapting this set of questions for computer use, considering several criteria which would ensure an effective transition from paper to computer. Immediately prior to clinical use, the computerized questionnaire prototype was administered to patients as well as health professionals and other experts. Their critiques were used to refine the prototype and arrive at a final version of the questionnaire for clinical use. However, refinements continue to be made, based on input from patients.

Questionnaire Content

It was hypothesized that identifying those factors most closely linked to LBP disability would enable

us to predict which individuals are at greatest risk for disability. A literature review was the first step toward compiling a list of pertinent factors. Next, a panel of six experts in low back pain research or related fields was convened. Using the list derived from the literature search, the "panel of experts" identified other potentially useful predictive factors and estimated the relative importance of each. The panel also suggested reliable ways to assess, or measure, these factors.

Through the literature review and panel discussions, we arrived at a list of 30 factors currently thought to be the most influential in determining disability resulting from low back pain. Factors are grouped into the broader categories of Occupational Factors, Psychosocial Factors, Injury Factors, Diagnostic Factors, Demographic Factors, Medical History, Health Behaviors, and Anthropometric Factors. (The complete list of 30 factors is available from the authors upon request.)

Questionnaire Development

In creating the computerized questionnaire, a foremost aim was to make the questionnaire clear, concise and easily understood. In this way patients would be less likely to answer erroneously due to misunderstandings. The panel helped in this stage of developing the questions by suggesting ways of measuring the factors. They either recommended a standard, well-known instrument or suggested possible phrasing. Sentences were short and to the point, and used words which are common in everyday language. Our guideline was to write questions which would be understandable at an eighth grade reading level.

Adaptations for Computer

Our patients come from varied educational and professional backgrounds, and not all have experience with computers. For this reason, the computer should be as nonthreatening as possible so that novice users feel as comfortable with the medium as experts.

To ensure an enjoyable experience for patients, we made our computerized questionnaire as "user-friendly" as possible. Here, user-friendly means clear, consistent and easy to use. In designing the computerized version of the questionnaire, the following requirements were deemed essential: visual clarity, an easy-to-use input device, pacing determined by patient, reduced operation time, and other modifications which could enhance user-friendliness. Suggested methods for achieving these ends are elaborated below.

Visual clarity. The screen layout and design should be clear, uncluttered and should prevent confusion. Additionally, consistency between screen displays should be maintained (1). This trains the patient, or computer user, to look at certain areas of the screen for various functions. For example, if the question is always placed in the upper left portion of the screen, and the responses are always to the right of the question, the user quickly learns to anticipate this spatial arrangement. Decreased confusion and frustration, and increased efficiency result. The user does not waste time sorting out what is being displayed, but rather can spend the time concentrating on what is being asked.

Easy input medium. Typing should be avoided, or kept to a minimum since many people do not type or are not familiar with a typewriter keyboard. To circumvent this problem, we used a touchscreen computer. The patient selects a response by touching the screen in a predefined area. The touch-sensitive areas are easily specified or relocated by the program, and no special tools are required to activate them. The patient may use a finger, pencil, or any other similar object to touch the screen.

Adaptations for self-pacing. Because people read and think at different rates, it is important to allow the patient to set the pace of the questionnaire. Questions that flash by too quickly and do not allow sufficient response time may prove frustrating and stressful for those inexperienced with computers. Moreover, if questions are presented too slowly, one tends to become impatient or bored. Either situation may result in missing or inaccurate data. The best solution is to allow the user to proceed at a comfortable pace by selecting when to advance to the next question.

Execution speed. Branching (passing over inappropriate questions) should be employed to minimize the amount of time required for answering. Execution speed can also be increased by selection of proper software.

User friendliness. Enjoyment of the questionnaire, as well as its ease of use, should be considered in all phases of design.

DISCUSSION AND CONCLUSIONS

We have used this system to obtain comprehensive information from 100 patients who were seen at a low back clinic. Based on our experience and comments from patients, we have concluded that this method of collecting data is quite effective.

At least one half of the patients had never used a computer or had limited computer experience. Despite this, none objected to using the computer. In fact, upon leaving, several spontaneously remarked that the computer experience had been enjoyable.

Although none of the patients objected to using the computer, most felt that the questionnaire was too long. The average time required to complete it was one and one-half hours. We have now refined the questionnaire so that the most recent version requires less than one hour to complete. It is hoped that this modification will increase patient compliance and allow more patients to participate.

Two major advantages of using a computer are that information collected is immediately available, and that errors in the data set are less likely to occur. Once a patient has completed the questionnaire, a report of the results can be printed out immediately and distributed to the attending

physician or included in the patient's medical chart. The computer readily calculates scale values, derived from responses to sets of questions, and compares them with average scale values. Such summarized information is useful to the clinician for determining a patient's status relative to the norm.

A second advantage to using a computer for data collection is that errors are reduced. The computer can be programmed to allow one or more responses from a list of possible alternatives, or to skip a question altogether. At the same time, the patient is prevented from modifying or accidentally omitting questions. Thus the computer helps ensure consistent responding among patients. Finally, data errors due to transcription are reduced because the computer generates a data file which is then immediately available for analysis.

We have used the computer to collect data on factors related to low back pain disability. However, this approach can be used in a variety of situations and with individuals from various educational and technical backgrounds.

ACKNOWLEDGEMENTS

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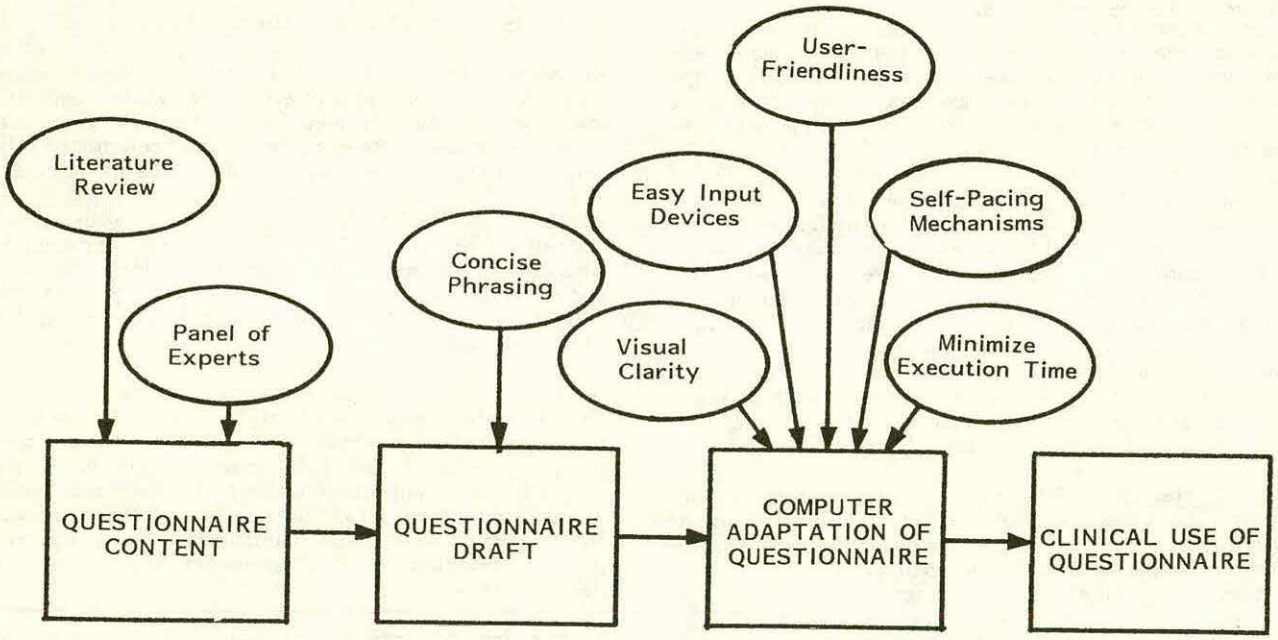

Figure 1. Development of a comprehensive computerized questionnaire.

AN ELECTRONIC, PROGRAMMABLE ACTIVITY TIMER TO PROMOTE RECREATIONAL EXERCISE IN PHYSICAL THERAPY

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ABSTRACT

Physical therapy is by its very nature an intimate, labour-intensive activity. To allow therapists more time to deal with new and/or problem cases, a new technological aid has usefully been employed for promoting routine exercises by incorporating these exercises in a stimulus-reward training system. We have designed, built and clinically tested a Programmable Activity Timer (PAT) which continues to provide rewards/encouragements/ reminders to patients engaged in locomotor exercise programmes and which therefore acts as a "surrogate" therapist allowing more cost-effective utilization of trained personnel. The device is therapist-programmable for: number of exercises in a sequence; duration of that sequence, and rest/reward period between sequences. By appropriate choice of input switches, (eg. range-of-motion, force) and output "reward" devices (eg. computer games, household devices, VTR's) a wide range of therapies and goals can be integrated and customised for the patient's needs. In addition, it incorporates a data-logging function enabling the therapist to assess patients' progress and review the treatment at any time.

INTRODUCTION

The escalation of trained staff costs, as well as greater awareness on the part of patients is placing increasing pressure on health professionals to re-examine many people-intensive programmes in Hospital Health Care.

There is a growing trend towards home health care especially for those with chronic conditions who need long-term management and therapy. Rehabilitation programmes are typically staff intensive. Physical therapy, both Physio- and Occupational is no exception. In the face of escalating patient numbers, earlier discharge from hospital and limited professional staff, we have managed, by the use of complementary technology to increase the cost effectiveness of our physical therapy programmes.

BACKGROUND

Our philosophy has been that most patients once they are familiar with their own exercise/motion therapy, are quite capable of carrying out these activities provided their interest can be stimulated and maintained. This is particularly true in the case of children for whom the ultimate aim of long-term therapy is not easy to comprehend.

Rather than emphasise repetitive regimented exercise sequences, we have attempted to "bury" these activities into a game or activity-of-daily-living task so that the exercise therapy becomes the means to the end, rather than the end in itself. The programmable activity timer (PAT) that we have developed, is designed to be as flexible as possible, so as to be used for a wide variety of therapeutic tasks, simply by choosing suitable input "switches", setting therapy goals on the device console and selecting the most attractive "reward" for successful completion of these tasks. A typical application is the encouragement of a paraplegic client in a wheelchair to become aware of the importance of regular lifting of the pelvis to promote circulation and minimise the danger of pressure sores. The activity which is attractive to this person might be watching TV. In this case the therapist would put a pressure-switch under the client's ischial tuberosities, and connect it to the unimodal input of the PAT. The TV receiver would be connected to the output which would be switched "on", for a selected period (typically 20 minutes) after the client had lifted him/herself sucessfully a preset number of times (eg. 3) within a preset period of time ((eg. 30 secs) for at least 10 seconds (say), per lift. Before the "reward" activity is cut off, a highly visible warning light turns on and/or an adjustable buzzer sounds. Initially, users of the PAT tend to wait until the last moment before doing their therapeutic sequence. However within a short time, they become "unconsciously aware" that the warning/cue indicator is on, and automatically do their exercises, without breaking the continuity of their TV programme and interestingly, in many cases, without being aware of having performed their exercises.

In a sense, the PAT provides behaviour modification in a subliminal way and acts as monitor of skin blood flow disruption just as a non-paraplegic person's skin pressure and pain receptors would normally.

MATERIALS AND METHODS

The PAT is a programmable, event-driven counter/ switch used to control the mains electrical power supply to tools, devices, or games used for physical therapy and activities of daily living.

Input

(1) One single-pole, jack-type socket for each of two normally-open microswitch inputs. (one is used for unimodal activities, both are used for bimodal).
(2) A 5 pin DIN socket, connected in parallel with the above, provides an alternative input.

Selector and test switches

(1) A mains power switch
(2) A RESET push-button to initialise the PAT with new preset control values and to zero the TOTAL MOVEMENTS counter/display.
(3) A DPDT switch to select UNIMODAL (a limit switch or BIMODAL (2 limit switches) mode.
(4) An overriding ON/OFF switch to remove reward device power in an emergency.
(5) A push-button REWARD DEVICE TEST switch to check correct operation of the device to be controlled by the PAT.

Preset Controls (Set up by therapist)

(1) A single-digit decimal thumb-wheel switch, labelled PRESET MOVEMENTS (1-9) indicating the number of movements in a therapeutic sequence.
(2) A single-digit decimal thumb-wheel switch labelled SECONDS PER MOVEMENT, indicating the time allowed per movement.
(3) A two-digit decimal thumb-wheel switch, labelled MINUTES REST, indicating the rest/reward period, (maximum 99 minutes).
(4) A potentiometer which is used to set the alarm/begin-exercises buzzer to acceptable level. The buzzer may be turned completely off.

Patient/Therapist feedback displays/indicators

(1) A LED to indicate that the device is powered up.
(2) A neon light to indicate mains power at the output socket.
(3) 2 LEDS indicating closure of the two input switches (these act as continuity testers to confirm successful connection and operation of the input device).
(4) A large highly visible "lighthouse" on the top of the PAT. While on, it indicates that movements are expected; off indicates successful completion of a sequence. It also flashes while "ON" to serve as a guide as to the expected repetition rate of each exercise in a given sequence.
(5) A single-digit numerical LED display, labelled MOVEMENTS REMAINING, which indicates by countdown the number of movements in a sequence which have still to be performed.
(6) A four-digit numerical LED display labelled TOTAL MOVEMENTS, which keeps a running total of all successful movements, providing both patient and therapist with a quantitative measure of activity and progress.

Output

A single, earthed, 15 amp shuttered MAINS POWER socket

OPERATION

To set up the system, the therapist selects the appropriate input for the client's needs. To date these have included simple single-switch/on-off/push-release movements (designated Unimodal in this text), and range-of-motion (2 switch or Bimodal) exercises each of which can be loaded or free. Naturally the loading, position, angle of application, and range-of-motion limit switches are the choice of the therapist. Using the latest industrial sensors it is possible to require the patient simply to interrupt a light beam to achieve the same switching action.

A device of obvious interest to the patient is selected as the "motivator", sewing machines, TV's, video games, and personal computers to name but a few.

Sequences of exercises, their repetition rate and duration as well as the interval between them are all under the control of the therapist and are set up for individual patients, as appropriate. The PAT counts the closures of the switches as "successes" and allows a preset time to elapse before resetting and indicating a restart condition to the patient. Should the patient perform the exercises within the selected time-span, a "reward" is given by switching on the power to the chosen motivational appliance for a preset "reward" period. If the activity is not successfully completed within the set time, a warning buzzer sounds to alert the patient. If this is repeatedly ignored the "reward" is switched off after a certain time interval and can only be re-enabled by completion of the deficient exercise sequence or by manual resetting by the therapist.

Since the device is intended to operate without the therapist necessarily being present, a counter of "TOTAL MOVEMENTS" for that patient/session has been incorporated. This counter can be reset at will by the therapist and therefore provides valuable clinical feedback of the level of activity, motivation and appropriateness of the particular therapy.

DISCUSSION

To date, two models of the most recent version of the PAT have been used continously in a typical clinical O.T. setting for 2½ years. Patients who have benefited include, cerebral palsy, polio, CVA, tendon/muscle contractures, arthritis, fractures and orthopaedic post-surgical. Motivation has been improved and therapy progress accelerated.

CONCLUSION

The University of Cape Town PAT has proved to be a valuable adjunct to conventional Occupation Therapy at Groote Schuur Hospital, and it has been well-received by patients and staff, who can now concentrate on "problem cases".

In future, using a similar philosophy but more modern technology we plan to produce a state-of-art microprocessor-based instrument, programmable from a keyboard which will hopefully perform even better and at lower cost.

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ABSTRACT

The ability to communicate is one of the most limiting losses imposed by severe physical disability. A communication system for the severely physically disabled patient should be easy to set up and operate, allow the patient flexibility of expression, and be low cost. We believe a properly designed Morse code communication system can fulfill these requirements.

Our system is based on the Commodore 64. The Program is contained in a cartridge, thereby eliminating the need to load the program. However, changes can still be made to custom fit the program to a particular patient. The system combines the advantages of a committed device with the flexibility of software.

INTRODUCTION

The ability to communicate is one of the most limiting losses imposed by severe physical disability. By definition, physical disability restricts reliable control of body parts, the restriction being proportional to the degree of the disability. Reliable motion may be so restricted as to be limited to the head, a single digit, or the blinking or movement of the eyes (1).

A communication system for the severely physically disabled verbal or non-verbal patient should fulfill the following requirements: a) minimum patient effort for operation, b) easily set up, c) independence from attendant assistance once set up, d) communication on a scratch pad display, e) hard copy capability, f) flexibility of expression beyond the capabilities of a simple typewriter; the patient should be able to create special characters or even illustrate text, g) use should be relatively independent of the patient's position, e.g., sitting or lying, h) low cost.

BACKGROUND

The automatic Morse code keying systems designed for high speed radio transmission are ideal for use by the physically disabled. Their use requires only reliable operation of a single pole double throw center off switch. One position controls dots the other dashes. The switch can be mounted in a hand splint as shown in Figure 1. A simple to and fro motion of the thumb controls the dots and dashes. The splint provides positioning of the thumb, freeing the patient from upper extremity or body positioning requirements. When the patient's hand is placed in the splint, control of the switch is independent of other motions of the body or extremity. The rate at which the dots and dashes are generated is automatic and, therefore, independent of the operator. The flow of information is CONTROLLED rather than

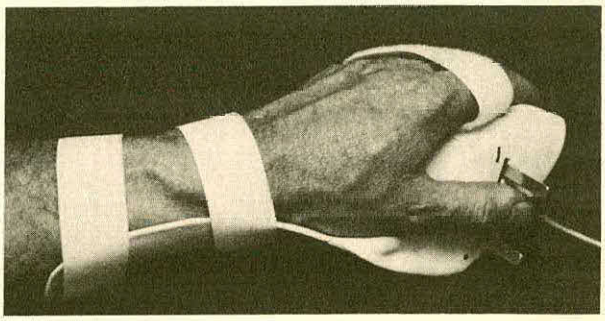

Figure 1

GENERATED as with mouth stick typing. The rate is set according to the abilities of the operator. Conversion of the Morse code to an understandable display of characters can be accomplisehd with a microprocessor based system. If desired, hard copy can be obtained.

We believe that a properly designed microprocessor based Morse code communication system can fulfill all of the previously stated requirements. Although good systems are available (2) (3) (4) (5), they are expensive and/or not flexible enough for many patients. There are also Morse code conversion programs available which are written for some of the popular computers. These have the advantage of being more affordable but the disadvantage that someone must load them into the computer. Many of the popular computers are also more costly than many patients can afford. The work of this presentation was prompted by our belief that there are no systems which fulfill all the criteria listed in the introduction.

MATERIALS AND METHODS

We based our system on the Commodore 64. Though costing less than $150, it has many of the features of higher priced computers as well as features not found in other computers. One major feature is a port into which program cartridges can be inserted (Fig. 2). When this feature is used, the program is up and running as soon as the computer is turned on. Our program is contained in such a cartridge, thereby eliminating the need for an attendant to load the program. If a proper power switch is available, the patient can be completely independent in using the program. The cartridge is programmed on the 64 using the "CARTRIDGE-MAKER-64" accessory[1]. Because of the cartridge, this program has the advantages of a committed device with the flexibility of software. Changes can be made to custom fit the program to a particular patient using the "CARTRIDGE-MARKER 64". For a basic communication system the only additional items to the 64 are the plug in cartridge, a switch tailored to the patient, a joystick and a TV set. Excluding the cost of the special switch and the TV set the cost of this system is under $200. To take full advantage of

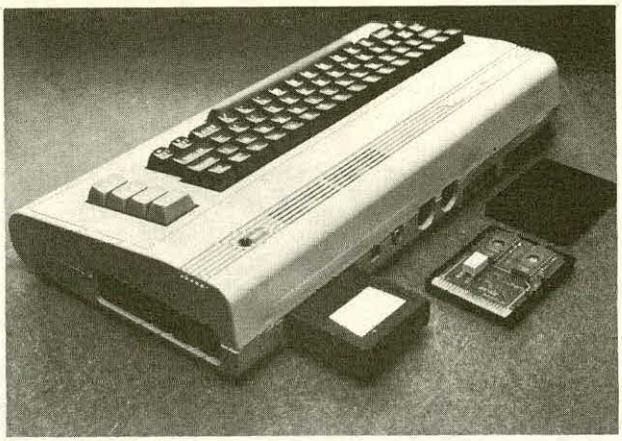

Figure 2

the system a disc drive and a printer are also required. This adds approximately $400 to the above cost.

The standard method of transmitting Morse code is to represent a dot with a short pulse and a dash with a long pulse of the same frequency. This causes problems when high transmission rates are desired. To overcome this difficulty, some of the early designers, such as Seamone and Schmeisser[2], devised a method whereby the dots and dashes are of equal length but the dots have a higher pitch than the dashes. This is the method used in our system.

Even though basic is available in this computer, it is too slow and uses up too much memory for this type of program. Therefore, the program is in machine language.

For the purposes of this program the Morse code is considered to be a series of binary numbers ranging in length from one digit through 6 digits. A dot is considered to be a zero and a dash a one. The information is stored in two registers, one, the code register, for the binary number itself and the other, the code length register, for the number of digits in the number (one through 6). Looking at the code register alone does not give sufficient information. For example 0000 0000 could be an E(.) an I(..) an S(...) or an H(....). Distinguishing among these requires looking at the code length register in which a 3, for example, identifies the letter S.

The code is converted into a character or instruction through two look up tables. The first table is divided into 6 sections. The beginning of each section is pointed to through the code length register. The number stored in that location is compared to the number in the code register. If there is no match, the next table location is compared and each successive location is compared until there is a match. The match supplies a pointer which selects the actual character or instruction from table two.

There are only 64 possible codes (characters or instructions) in each table. Through the use of the shift code, however, the number of print character codes is actually doubled. Although the table could be much longer, additional length would be of no advantage to this program since there is still space for 14 more characters or instructions.

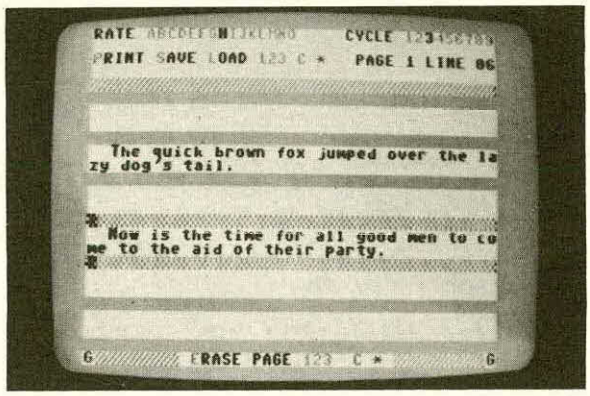

Figure 3

RESULTS

Through the combined use of Morse code and cursor manipulation via a joystick, the patient can perform the following tasks:

A) Easily set his own transmission rate (Fig.3 top line). This is not possible with most of the other systems.
B) Type up to three pages with 64 lines per page. As shown in Fig. 3, for the sake of clarity only six lines are displayed at a time and each line is separated by a colored bar. Each line is displayed as two 40 character lines but is printed as a continuous up to 80 character line. Information can only be transmitted and edited through one display line. This is the line between the two bars that begin with black boxes. The page line that the display line is on is shown in the PAGE NUMBER/LINE NUMBER register located on the third screen line down.
C) Set the line length from one character through 80 characters.
D) Edit the text by use of insert and delete commands.
E) Type all the characters on the standard typewriter keyboard. This is accomplished through the use of special codes which are not contained in the standard Morse code set. For example, upper and lower case characters can be typed through the use of a shift command code.
F) Print and or save individual pages (Fig. 3) third line down).
G) Load previously saved pages from a disc (Fig.3 third line down).
H) Enter a graphics mode (through the Gs on either end of the bottom line in Fig. 3). When the page is printed whatever appears on the screen is printed exactly as it appears on the screen. Special characters or designs which the printer cannot reproduce in the character mode can be reproduced in the graphics mode.
I) Combine the character mode and the graphics mode on the same page. This allows illustrations to be combined with the text.
J) Erase individual pages (Fig. 3 center of bottom line).
K) Line feed (line feed / carriage return code or

via the joystick.
L) Reverse line feed (joystick).
M) Access any line of any page quickly (combination of K and L).

COMMENT

What prompted our interest in the use of Morse code was what we recognized as the great disadvantages of mouth stick typing by the high quadriplegics. Such typing is cumbersome, slow and tiring. The patient must be carefully positioned in front of the keyboard. The mouth stick must be precisely placed over each key by the patient and then pressed.

The neck muscles and possibly the tongue perform the functions of the arms and hands respectively. The standard keyboard layout itself adds to the difficulties.

Instead of trying to physically duplicate the functions of one part of the body with those of another part, the present approach is to amplify any residual functions. The slightest motion of a digit can control a large assortment of devices through a simple, low force electrical switch as the interface with the system. The present system accomplishes versatile, yet struggle-free, communication despite extensive loss of motor control.

CONCLUSION

An inexpensive computer based Morse code communication system has been described. It combines the versatility and flexibility of software with the convenience of a hard wired committed system.

FOOTNOTES

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17.4

DEVELOPMENT OF A MICROPROCESSOR BASED KEYBOARD EMULATOR FOR IMPROVED ACCESSABILITY

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ABSTRACT

An Interface Unit is currently being developed at the Hugh MacMillan Medical Centre (HMMC) to enable physically disabled students who cannot use the keyboard and/or trackball due to a lack of fine motor control to access the Icon computer system, an Ontario Approved Educational Microcomputer.

This paper discusses the current status of a completed prototype Interface Unit and the development of a second prototype that includes a remote communication link. Future plans are also presented.

INTRODUCTION

The Icon computer is the first Ontario Approved Educational Microcomputer which will be resident in many of the Province's schools. This computer was specifically designed as an educational computer and consists of a fileserver and from one to 32 workstations linked via a coaxial cable network.

In 1985 Bill 82 (the equivalent of U.S. Public Law 94-142) became law stating that all individuals are to have equal opportunity to receive a public education (Wilson, 1983). Because the Icon computer had been developed during this time, it became necessary for the Ontario Ministry of Education to address the special needs of such disabled individuals relative to its use. Within this population, there are a significant number of students whose lack of fine motor control restricts, or prohibits, their using a standard computer keyboard.

A multi-disciplinary team comprising of experts in the technical, educational, occupational therapy fields, and associated consultants was formed at the HMMC to address the issue. Teachers and therapists play a major role in evaluating the use of the Icon 'as is' by disabled students to identify accessing problems. This paper focuses on the technical solution to accessing the computer through the development of an interface unit that permits a variety of input devices to be used.

THE PROBLEM

Many physically disabled students have the motivation and the cognitive ability to perform academically as well as able-bodied students. The main obstacle for most is their level of fine motor control which also restricts them in using standard computer keyboards. Such students may either have difficulty in controlling hand and arm movement or cannot produce sufficient forces to operate a standard keyboard.

For students who have movement of hands and arms, the speed and accuracy with which they can target on a single key are affected. They may miss the intended target key altogether or hit multiple keys simultaneously. Some students have body sites rather than single finger targetting ability, in which case keyboarding is extremely difficult.

For students who have very limited movement of arms and hands, the problem is that of being able to reach the keyboard. For wheelchair users, this problem is compounded by the difficulty of placing the student sufficiently close to the keyboard to access it.

Some students may not be able to control movement of their hands and arms in which case the activation of a single switch by another part of the body must be used.

The Icon computer provides a trackball which is located on the upper right hand corner. For most students with a physical disability, the trackball is difficult to operate, especially because of its location.

The Icon computer is a closed architecture machine which precludes the user or developer access to the internal hardware or software stucture. Interfacing to the Icon is thus restricted to an alternate keyboard input, one serial port, and two parallel ports.

Many alternate input devices to overcome access problems are available such as various sized keyboards and pointing devices, including light pens, joysticks, single or multiple switches, trackballs, and mice.

The main objective of the HMMC team is to develop an interface which will allow access to the Icon computer via such input devices by physically disabled students.

THE SOLUTION

As a result of the limited access to the Icon's architecture, a dedicated interface unit was developed that will support the alternate user input devices. The

Interface Unit is a microprocessor - based system which is interfaced to the Icon computer and effectively emulates the function of the Icon keyboard. The Interface Unit sends key codes to the Icon system's alternate keyboard input port. These key codes are identical to the Icon's keyboard codes which makes the Interface Unit virtually transparent to the system.

THE INTERFACE UNIT OVERVIEW

The Interface Unit has two physical connections to the Icon, (i) a connection to the alternate keyboard input for transmitting keyboard codes to the Icon; and (ii) a connection to the serial port for bidirectional communication between the Icon and the Interface Unit for the redefinition of keyboards. Two versions of the Interface Unit are currently being developed. Both versions are connected to the Icon as described above. The first version consists of a single unit which requires the user's input devices to be plugged into it in order to have access to the Icon as shown in Figure 1.

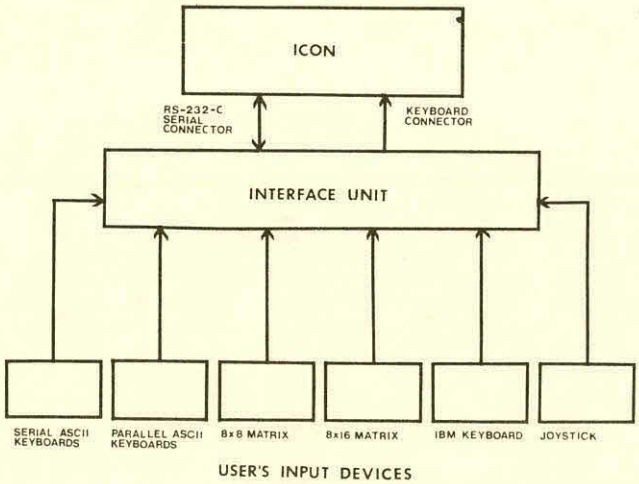

Figure 1. Block Diagram of First Version Interface Unit

The second version effectively divides the first version into two physically independent units that allow the interfacing of the Icon to the users' input devices via a remote bidirectional communication link.

The HMMC Interface Unit

The Interface Unit is a microprocessor - based system consisting of an NSC 800 as the central processing controller. Two parallel input/output devices are provided to allow for the interfacing of the input devices to the Icon. One serial UART device is provided to allow for the interfacing of serial input devices to the Icon. A second UART device is also provided and is connected to the Icon's serial port via an RS-232C link. This allows for bidirectional communication between the Icon and the Interface Unit. Its purpose is to provide the downloading of keyboard translation tables to the Interface Unit. These translation tables may contain single key translations or tokenized messages for each key closure. RAM is provided for storing the downloaded translation tables and ROM is provided for storing the control program. All components of the Interface Unit are of low power CMOS to reduce power consumption which is provided via the alternate keyboard input of the Icon.

All user input devices are unique in the type of connector used and the signals which they produce. The Interface Unit's main purpose is to accept input from these devices, convert the signals into Icon key codes, and send them to the Icon via the alternate keyboard input.

Remote Link Interface

This section addresses the interfacing of the Icon to the users and their input devices via a remote communication link. The transceiver, presently under development, will provide users with greater freedom of movement and more independence in the use of the Icon system by eliminating the wiring between the users' input devices and the Icon.

The Remote Interface consists of a Remote Interface Adapter and a Remote Interface Controller as shown in Figure 2.

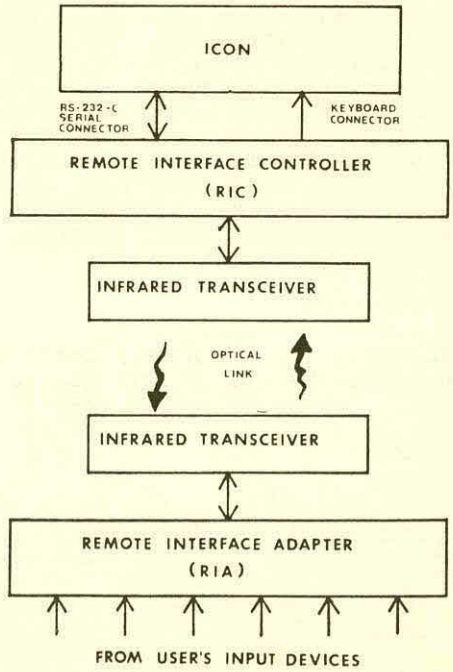

Figure 2. Block Diagram of Second Version Remote Link Interface

One unit, the Remote Interface Adapter (RIA), connects to a user input device and accepts data from it. The RIA sends the codes to the RIC via the infrared link using a protocol developed at HMMC. The RIC converts the codes to Icon key codes and sends them to the Icon. Handshaking exists between the RIA and the RIC to ensure proper transmission.

SUMMARY

The first version of the Interface Unit is finalized, packaged, and is currently undergoing clinical evaluation in the Hugh MacMillan School. Additional printed circuit boards have been produced for field testing in other institutions and for further development of the Remote Link Interface.

The clinical feedback received thus far has been very promising in that the Interface Unit is addressing many of the accessing problems encountered with the Icon. Changes to the Interface Unit will be made as the clinical evaluations continue. The Interface Unit will allow the following devices to be used with the Icon computer: Serial ASCII keyboards (RCA), Parallel ASCII keyboards (Apple II+, MOD, and Elementary MOD), IBM keyboard, Unicorn matrix keyboard, and King and Mini matrix keyboards.

FUTURE PLANS

Future plans include incorporating changes into the Interface Unit as clinical evaluations are completed. Completion of the Remote Interface Unit and a form of trackball emulation using the completed boards will be the base for the development of a final version of the Interface Unit which will incorporate any additional required functions. Trackball emulators will include an external trackball, 8-position joystick, proportional joystick, one-through-five switches, and an Apple II and/or IBM game port.

ACKNOWLEDGEMENTS

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One Finger Operation of the IBM Family of Personal Computers

Charles C. Lee
Gregg C. Vanderheiden
Arthur Rasmussen

ABSTRACT

Software programs to allow one finger operation of the IBM family of Personal Computers have been written by the Trace Center at the Waisman Center, University of Wisconsin. The programs remain resident in the computer once invoked. A different program is required for the IBM PC/XT, the IBM PCjr, and the IBM AT since each computer handles the keyboard a little differently. The programs provide many options to allow the user to configure the program to meet their specific needs and situation. Options include: tri-state shift keys, audible feedback on the states of the shift keys, manual turning on and off of the program, audible feedback when the program is manual turned on or off, user selectable delay time on the automatic repeat of the keys from 0.5 seconds to 60 seconds in 0.1 second intervals, automatic turning off of the program if a shift key and any other key is held down at the same time, and automatic turning off of the program if there is no activity at the keyboard for a user selectable time from 1 to 60 minutes in 1 minute intervals. All of the above options are set by software switches. Also, there are options to have the program be on at start-up, to show the status of all the options on the screen at start-up, and to un-install the program.

INTRODUCTION

Many handicapped individuals can only use a keyboard with one finger, a mouth stick, or a headstick. This introduces three problems. The first is the problem of holding down two keys at the same time. This problem exists when trying to hold down a shift key (Left Shift, Right Shift, Ctrl, and Alt, and the Fn for the PCjr), and another key.

The second problem is the automatic repeat of a key. Many handicapped individuals, especially those with cerebral palsy, are slower in their hand or finger movements when typing on a keyboard. The IBM computers automatically repeat the key if it is held down for more than 1/2 a second (delay time) and repeats the key at 10 characters per second (typematic rate). Yet it takes some individuals 2 second before being able to release the key which means that they would get as many as 10 extraneous characters.

There are programs which currently deal with either one of the problems, but not with both. These programs also do not deal with the third problem of what to do when handicapped and non-handicapped individuals use the same computer, since the tri-state shift keys and the increased delay time on the automatic repeat is not generally desirable for the non-handicapped individual. This poses a difficult problem when the non-handicapped person is not able to exit the current application program or reboot the computer.

IBM KEYBOARD OPERATION

The keyboards of the IBM family of personal computers operate in a similar but not identical fashion. Each key has a code which identifies it, called a scan code. When a key is depressed, the keyboard sends to the computer a down scan code of the key which was depressed. When the key is released, the keyboard sends to the computer an up scan code of the key released. If the key is held down for the set delay time, the keyboard sends down scan codes to the computer at the typematic rate. If a new key is pressed down when a old key is repeating, the old key's scan code is no longer sent to the computer. Instead, the new-key's scan code is sent to the computer. It is up to the computer to determine how to treat the information received from the keyboard. Every time the computer gets a scan code from the keyboard, it interrupts whatever it is doing, translates the scan code into a key and performs the appropriate action. The program that does this is called the keyboard interrupt routine. The interrupt routine keeps track of the state of the shift keys in a set memory location. The address to the interrupt routine is placed in an interrupt vector.

DESIGN GOALS

1. To allow one finger use of the shift keys.

2. To allow a selectable automatic repeat key delay time and to set the typematic rate accordingly.

3. To provide manual means to turn on and off the program for the user.

4. To provide an optional feature which would detect when a non-handicapped user was using the computer and turn off the first and second features listed above.

METHODOLOGY

The goals were accomplished by writing a new keyboard interrupt routine that would be executed instead of the original keyboard interrupt routine. The new routine then makes the keyboard behave differently by handling the scan codes differently. It still uses the original keyboard interrupt routine to translate the keys and to perform certain action (e.g., rebooting with CTRL-ALT-DEL). When the program is loaded, it changes the interrupt vector for the keyboard so that the new keyboard interrupt routine is executed instead of the original keyboard interrupt routine. It then installs the new keyboard interrupt routine into memory, where it remains resident, even when other programs are running. If the program is loaded a second time, it determines whether the new keyboard interrupt routine is resident; if so, it change only the options of the interrupt routine and does not install another copy of the routine in memory.

DESCRIPTION

Tri-State Shift

The program supports the option of a tri-state shift on the Right-Shift, Left-Shift, Ctrl and Alt keys (also the Fn key on the PCjr). Depressing and releasing a shift key once makes it active for the next non-shift key. Depressing and releasing a shift key a second time (before any non-shift key) locks the shift key in an activated state until unlocked. Depressing and releasing the shift key unlocks the key if it is in the locked state.

More than one shift key can be shifted or locked as long as all of the shifting or locking is done before depressing a non-shift key. Depressing a non-shift key changes all shift keys in the shifted state to a non-shifted state, but leaves those in the locked state alone.

Delayed Repeat

The delay repeat option allows the user to set the delay time for the automatic repeat. It can be set from 0.5 seconds to 60 seconds in 0.1 second intervals (on the AT it can be as low as 0.3 seconds). This will only be effective on the non-shift keys. The key will then automatically repeat at a rate of half the time of the delay. For example, if the delayed repeat time is set at 2 seconds, the key must be held down for 2 seconds before it will be automatically repeated. It will then repeat every second until the key is released or another key is depressed. The automatic repeat can be effectively disabled by setting the delayed repeat time to 60 seconds.

Manual On/Off

The program can always be manually turned on and off by the user. To turn on the program, the Left-Shift key must be depressed and released four times in a row. To turn off the program, the Right-Shift key must be depressed and released four times in a row. This allows the knowledgeable handicapped or non-handicapped user to turn the program on or off.

Automatic Turning Off

This option is very important if a non-handicapped and a handicapped individual use the same computer. It allows the one finger program to turn itself off so that it does not prohibit the non-handicapped user from effectively using the keyboard.

There are two ways the program can automatically turn off. Both are options. The first option is to have the program watch for a shift key and then any other key depressed simultaneously (something a one finger typist does not do). The program then turns itself off and the keyboard performs as it normally would.

The second way that it automatically turns itself off is when there is no activity at the keyboard for a set period of time (time out). The time out can be user selectable from 1 to 60 minutes in 1 minute intervals. This is useful in cases where the handicapped user leaves (without turning the routine off) and a non-handicapped user comes several minutes later to use the computer. The program will already have been turned off by the time the non-handicapped user begins.

Neither of these methods require any knowledge of the non-handicapped user about the special functioning of the one finger program.

Note: Even if the two key turn off option is not selected, the states of the shift key registers are set to their actual states if a shift key and another key are depressed at the same time. In this way, the keyboard will always act like a normal keyboard when two or more keys are held down.

Feedback

Feedback to the user is very important in any application. This is especially true when a single action, like pressing a key, can have more than one result. Audible feedback is provided via the speaker to indicate the new shift state when a shift key is depressed. A low and then a high beep is used to indicate the shift key is now in the shift state. A single high beep is used to indicate the shift key is in the locked state. A single low beep is used to indicate the shift key has been unlocked.

There is also audible feedback when the program is manually turned off or on. A low to high rising tone indicates that the program is now turned on, and a hi to low falling tone indicates that the program is now turned off.

The current status of all the options can be displayed when the program is run to give feedback to the user on the current set-up.

Uninstalling the Interrupt Routine

The keyboard interrupt routine can be uninstalled by selecting the uninstall option when running the program. The new keyboard interrupt routine will then no longer intercept the keyboard scan codes from the original keyboard interrupt routine. However, the memory space cannot be regained for use without rebooting.

Applications

This program is excellent for individuals who can type with only one finger or the equivalent (headsticks, mouthsticks, etc.). It is also useful for non-handicapped individuals, since there are many times that a non-handicapped individual has use of only one hand when at a computer (e.g., the other hand is holding down pages to a book or doing something else). The program is useful to those who are slow at releasing a key once it is depressed whether they need the shift feature or not.

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IBM Technical Reference AT

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A Multitasking Approach to Computer Access: Practical Considerations in Implementation

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ABSTRACT - Many benefits of computer access for the disabled have been well established. A variety of approaches for allowing microcomputer use have been proposed and demonstrated in recent years. The software-based multi-tasking approach has advantages in terms of cost, flexibility, and versatility compared to these other techniques. The evolution of microcomputer operating systems now permits the implementation of access strategies based on concurrent operation. The approach has been demonstrated with current technology and represents an alternative to access strategies requiring hardware modification or manufacturer cooperation.

INTRODUCTION:

Microcomputer use by the disabled is recognized as beneficial in educational, vocational and personal settings. However, these individuals frequently have difficulty with traditional user interfaces. The typewriter keyboard, for instance, requires a level of manual dexterity that often prevents its use by members of this population.

In recent years, the goal of providing microcomputer access has been addressed by members of the rehabilitation community. Many approaches have been suggested, each having associated advantages and weaknesses.

Custom Software - The custom software approach involves writing programs that circumvent the need for standard keyboard input [Nelson et al., 1981; Teoh and Gates, 1982]. The disabled person would use a special word processor, a special spreadsheet, a special database manager, and so on. Programs that have been written to date consist primarily of word processors [Smith and Graystone, 1981] and communication programs [Demasco and Foulds, 1982; Kelly and Ford, 1984].

The user interface can be specially designed to efficiently execute the program and effectively utilize the disabled person's abilities. However, the development cost of these programs must be borne by a smaller number of consumers than a program written for the non-impaired population. This is clearly a significant barrier to the development of software of this type since very few packages other than word processors are available.

Software Modification - Another method for providing access for the disabled involves the modification of commercially available software packages. Very few examples of this technique exist. When the program is written, a "patch" is provided to permit alternative input methods. This requires significant committment from software developers who already operate in a demanding marketplace [Vanderheiden, 1981].

Hardware Approach - A method frequently discussed in the rehabilitation community involves the design and use of custom hardware interfaces to standard microcomputers. This hardware typically "emulates" the keyboard of the microprocessor; entries from the hardware appear to the computer as if they originated at the keyboard [Rogers et al., 1982]. This permits the use of many commercially available software packages, but requires that keyboard emulation hardware be designed for the desired microcomputer.

The majority of this hardware has been directed at interfacing with the Apple II™ series of micros. One approach involves the use of a plug-in card for the Apple and supports a limited number of types of user interfaces [Schwejda et al., 1983]. Another approach used an inexpensive computer to process the user's input (e.g., a scanning program) and provides appropriate keyboard signals to the target computer [Korba et al., 1983; Nelson et al., 1984]. The trade-offs involved here are flexibility, cost and packaging.

The availability of a keyboard emulator obviously must follow the introduction of any new microcomputer; in fact, lag is often needed to determine if the computer's popularity warrants the emulator design and production effort. An additional complication is that a manufacturer will not necessarily keep the keyboard signal specification constant; the IBM PC® family has three different keyboard signal protocols.

While these techniques have provided the disabled with some degree of access to microcomputers, none has truly provided unrestricted access to commercial software and a flexible user interface. Trends in business and scientific uses of microcomputers have resulted in development tools that are applicable to access for the disabled. Personal computers have recently become sufficiently powerful to allow for the implementation of multi-tasking operating systems. These allow a user to run two or more programs at the same time. This trend towards "concurrent" operating systems on this class of computer represents an alternative to current access strategies.

MULTI-TASKING APPROACH:

Multi-tasking presents a unique opportunity in accessing a broad range of commercially available software. User interfaces can be implemented in software and used in conjunction with other applications. For example, a single-switch scanning program can be developed that would provide keyboard input to a spreadsheet program running concurrently on the computer. This technique will prove to have many benefits.

A multi-tasking environment (MTE) shares the hardware resources (e.g., the display, the disk drives, and the central processor) among the active programs [Vanderheiden, 1981]. During the execution of many programs, the processor is idle; the MTEs utilize this time to run other tasks. While MTEs are currently available only for computers of moderate cost (about $1500),

emulation hardware and front end interface systems are not required.

The user interface of such a system is extremely flexible; no emulator hardware restrictions are present. The communication program, the program translating user input into application program keystrokes, is free to utilize the full resources of the computer.

The resultant configuration allows the disabled user to run virtually all currently available software packages. The menu of the disabled person's communication program can be customized to output commands (i.e., keystroke sequences) that will efficiently execute the application program. Disabled user's can even play many games on the computer that ran too quickly prior to the installation of the MTE.

The communication systems developed with this approach can be written in a standardized language to afford maximum portability to other computer systems. For example, programs written in C frequently compile and run with little alteration on dissimilar systems. The microcomputer market is rapidly improving; it is a mistake to devote much effort to developing a system that can only be used with today's machines.

Unfortunately, given the current state of MTEs, the advantages listed above cannot be fully exploited with microcomputers. While these operating systems will eventually become as capable as their minicomputer counterparts, the current limitations must be considered.

The greatest difficulty that currently exists is the transmission of keystrokes from the communication program to the application program. While many MTEs allow text transmission between jobs, it is often difficult to send "special" characters such as cursor or function keys. In systems that include multi-user capabilites, considerable effort is expended to prevent coexisting programs from corrupting each other's environment.

Control over the division of the computer's hardware resources is also restricted by current MTEs. For instance, most MTEs use the system clock to allocate time for each job. The minimum time division for the IBM PC is 1/18 of a second per job. This time slice limitation may prevent the development of some interfaces (e.g., puff/sip) without the use of interrupt based functions.

Other potential obstacles in the practical application of multi-tasking access are keyboard and screen conflicts. It is possible, though not considered good programming practice, to circumvent the techniques for software development provided by the computer's manufacturer. These "shortcuts" are sometimes used when reading the keyboard or writing to the screen. In cases where these recommended practices have been violated, special efforts may be required for the MTE to function correctly.

These problems will most certainly become less prevalent as operating systems incorporating multitasking mature. The proliferation of MTEs and RAM resident utilities are exerting pressure on software authors to adhere to standardized development practices.

IMPLEMENTATION:

Current efforts to develop working examples of communication systems using the multi-tasking approach (MTA) have been directed towards the IBM PC. This microcomputer has become the defacto standard in the business and academic arenas; many sophisticated software and hardware products are available for it. While the Apple II computer remains prevalent in the educational community, the need for access to this computer by the disabled is being adequately met with the hardware mentioned above.

A variety of MTEs have been developed for the IBM PC [Awalt, 1985]. These implementations of MTEs range in sophistication from PC-DOS™ patches to complete multitasking, multi-user operating systems (e.g., UNIX™ look-alikes). This discussion will be limited to our experience with three specific systems.

Concurrent DOS™ is an operating system designed to allow users to run both PC-DOS and CP/M™ applications in a multitasking multi-user environment. This operating system allows up to four applications to run concurrently; a windowing capacity allows users to switch between screen displays of each task.

Although Concurrent DOS is a powerful example of currently available MTEs for the IBM PC, it imposes several undesirable requirements. Concurrent DOS is large; a significant amount of memory and disk space is required to implement it. It also suffers from the undesirable qualities of multi-user oriented systems; for example, intertask communication is restricted.

In marked contrast, Multi-Job® and DoubleDOS® are PC-DOS extensions that require little additional memory overhead and operate only as a single user MTE. Each of these programs reside between the operating system and the programs that run on the PC. They allocate processor time to each task according to user specified proportions. These PC DOS enhancement programs currently are the most attractive means for implementing the multi-tasking approach.

RESULTS:

Several prototype systems have been developed that have both proven the concept and provided valuable information concerning practical implementation. Communication programs have been written for each of the three traditional input methods [Vanderheiden, 1978]. These will be discussed in terms of how hardware resources are utilized as well as difficulties peculiar to their implementation.

Direct Selection - Commercial peripherals are available that can be used to create direct selection boards of virtually any size, CAD/CAM digitizing tablets can be utilized as zero-force input devices. Less expensive tablets, having touch-sensitive surfaces, are also manufactured. The "keys" on all such peripherals can be any size, different sizes can be mixed, and some even allow non-rectangular shapes.

A communication program was written that received input from a Houston Instruments Hi-Pad™ digitizing tablet. Positional information from this tablet was converted into a sequence of

keystrokes that would have been typed on the keyboard. These keystrokes, any combination of data, text, or application program commands, are placed in an area of the computer memory (i.e., keyboard buffer) where the application looks for keyboard entries.

Encoding - The hardware for implementing this technique is less specific than in the case of direct selection systems. Any of the direct selection hardware devices could be employed; slot switches, joysticks, and arrays of discrete switches are also candidates.

The communication program handling this mode of input converts a series of actions by the disabled user into a desired keystroke sequence. This type of program is more difficult for some individuals to operate; users must stay aware of their progress in the entry of the code. The user is either required to keep track of this progress or the MTA can display a window or screen for visual monitoring of message production. The use of display feedback will potentially make encoding a more feasible approach than it has traditionally been.

Scanning - The scanning implementation requires a display to provide the visual feedback necessary for successful use. Most MTEs allow the user to specify which task the display is monitoring at any given point. This will permit the scanning array to be displayed during message composition and the application program to be displayed at other times. An alternative approach is to invest in a second display to allow for concurrent viewing of both the application and the input program.

A communication program was written that received input from a standard game adapter. The primary display adapter is used by the application program and a second display adapter and monitor are utilized to present an 8x8 scanning array. This configuration provides the highest degree of software compatibility, and allows the user to view the full scanning array at all times.

The most powerful feature shared among the above implementations is application specific menus. Since MTEs facilitate sharing of hardware resources, the input program may utilize the disk facilities to load a new set of keystroke selections when the user switches applications. For example, if a user switches from a word processing program (text intensive) to a spreadsheet (number intensive), the interface program will load the new command set and optimize digit entry. This flexibility allows the MTA to provide an efficient means of access for a large population of disabled users.

CONCLUSION:

The multi-tasking approach has great potential for providing powerful, flexible, and economical computer access for the disabled. The user can run the entire gamut of software available on the commercial market. The interface can be custom designed for each particular user/application combination. The approach, while not applicable to extremely low-priced computers, is an economical means for providing full computing resources. When compared to the cost of a traditional dedicated communication aid/computer combination or "nested computers" capable of providing the equivalent utility, a multi-tasking configuration becomes feasible. The additional cost for a system capable of multi-tasking is varies from $100 to $1000, depending on input method and desired display features.

This approach avoids costly custom hardware or modifications and does not necessarily require cooperation from software authors or computer manufacturers. The capabilities of MTEs will expand over the next few years, making this approach increasingly attractive. MTEs are available for the developing portable computer market. An MTA based system may be the ideal strategy for implementing portable communication aids that will possess a much greater range of functional capability.

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18.1

Clinical Assessment of the Case Neural Prosthetic Functional Neuromuscular Stimulation Hand System

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ABSTRACT

The Case Western Reserve University Rehabilitation Engineering Program has been developing a Functional Neuromuscular Stimulation (FNS) hand system that enables subjects with C5 and C6 complete spinal cord injuries to perform activities of daily living (ADL) tasks. Our laboratory has been in the process of assessing the clinical efficacy of the hand system. The major areas of consideration are: comparing the subjects ability to perform tasks with and without his prosthesis, evaluating the efficiency in which the subject performs the tasks, assessing the performance against the use of conventional alternative orthosis such as cuffs, splints, and modified apparatus, and assessing a hybrid approach combining FNS and surgical intervention such as tendon transfers.

INTRODUCTION

The purpose of the clinical assessment is to determine whether the FNS hand system provides a subject with enhanced function in performing ADL tasks. Qualitative assessment protocols have been used for a number of years in our upper extremity FNS program for C5 and C6 quadriplegic subjects. We are presently developing a more quantitative approach towards clinical subject assessment to investigate the following issues:

** the determination of physical profiles (i.e. sensory, motor, neurologic, and functional aspects)of our subjects prior to the dissemination of the FNS hand system and after they have used the hand FNS hand system.

** the correlation between physical profile and level of skill attained with an FNS system.

** the functionality with an FNS hand system versus the use of conventional orthoses.

** the extent of training required to attain maximum benefits in utilizing the FNS hand system

** To provide sufficient feedback to our technical staff in order that we may upgrade the hand system wherever feasible.

OVERVIEW OF SYSTEM USAGE

The subjects in our program are provided with the Case Neural Prosthetic Hand System (1) which consists of a command control device, portable patient based system , electrode cables, percutaneous electrode interface, and the intramuscular electrodes. The subjects are evaluated for control schemes to operate the FNS hand system in our laboratory utilizing the computer controlled laboratory system (2). The control scheme is typically a proportional control source (usually the transduction of motion of the opposite shoulder from the stimulated hand) or a gated-ramp activated by a switch. The stimulus parameters are derived from the linearization of the command control/stimulus output responses found during the laboratory evaluation of each electrode. The stimulus parameters and command control parameters are transferred into an EPROM which is then installed into the subjects portable patient based system.

The neural prosthetic hand system is applied in the morning (donning takes less than 5 minutes). Once this is done the subject can use his hand system at any time throughout the day. The subject is provided with a lateral prehension/release grasp and a palmar prehension/release grasp. The grasping mode is selected by a scanning technique which is activated by depressing a switch mountedto the chest next to the subject's shoulder transducer. The subject turns the system off by depressing the switch a second time. The user can also lock his hand in a set position during proportional control (which is usually accomplished by exceeding a shoulder velocity threshold on the axis orthogonal; to the proportional control axis).

CLINICAL ASSESSMENT

The clinical assessment is divided into four major sections: Sensory Evaluation, Passive Range of Motion, Motor Evaluation and Motor Neuron Mapping and Functional Application Testing.

The Sensory Evaluation section involves sensory testing of both upper extremities in areas representing spinal nerve root dermatomes. Pressure, light touch, temperature, moving two-point discrimination, proprioception and stereognosis are included (3,4,5,6,7). The Passive Range of Motion Section involves recording goniometric joint measurements according to the American Academy of Orthopedic Surgeons guidelines (8).

The Motor Evaluation and Motor Neuron Mapping is conducted with the subject sitting in his wheelchair. The scoring of the Manual Muscle testing is recorded on a 0 to 5 scale. A score of 5 indicates full range of movement against gravity and a "normal" amount of resistance at the end of the range as compared to resistance the evaluator is able to sustain with the same motion. A score of zero indicates no discernible active motion. Surface electrical stimulation is utilized for examination of individual muscle motor point excitation. Each muscle which is required for

either grasp is recorded with a "+" or a "-" indicating the presence or absence of excitability. If a muscle fails to be excitable with peripheral stimulation, it is judged to be a lower motor neuron involved muscle and therefore will not respond to electrical excitation.

The final section involves a functional application testing regime. These tests require: the subject to demonstrate proficiency in the use of the system, to evaluate the force of his grasp, to qualitatively describe the grasping patterns, to assess the level of fatigue resistance of each grasp, and to administer multiple performance tests.

The performance testing of the subject includes a Jebsen Hand Test (9), an Isolated Basic Functional Tasks test, a Coordinated Functional Tasks test and an Integration of Functional Tasks test. These tests are used to gauge the level of independence a C5 and C6 complete spinal cord injured person has attained with his FNS hand system and to compare these scores against other techniques such as using orthoses, modified apparatus, etc.

We are developing a scoring system which measures the time in which a person completes a specified task and the method in which the person completes the task. The method in which a person completes his task is classified in terms of level of independence he demonstrates during the test.

The Jebsen Hand Function Test assesses a person's unilateral ability to write, manage eating utensils, manipulate small objects, manipulate lightweight and heavy large objects (7). The test is included because of the collection of normative data and its' application to C6-C7 spinal injured individuals fit with wrist-driven flexor hinge splints. The major disadvantages of relying on Jebsen test scores alone is that the score of the individual is based on the speed of his performance and his comprehension of the tasks. There is no consideration of altered FNS prehension patterns which effect the test scores. The evaluation is strictly unilateral, thus no mechanisms for including the non-stimulated hand as an assist are measured. In many cases, use of the non-stimulated hand can greatly enhance a subject's efficiency in performing a task.

We have included a section of observational testing which is geared towards testing the subjects level of independence in various ADL situations. This section is divided into three parts: Isolated Basic Tasks, Coordinated Tasks, and Integrated Tasks. Isolated Basic Tasks test the ability of the subject to singularly pick up and place down common objects such as a telephone or a plastic computer diskette. Coordinated Tasks test the subjects ability to complete a functional task such as tissue wiping of the nose and placing a computer diskette in a disk drive and removing it. These Coordinated Tasks involve more than a basic skill involving FNS; trunk balance and the integration of shoulder and arm coordination along with forearm pronation and supination are also required. Integrated Tasks test the subject's ability to serially perform the more complicated functions. Examples of Integrated Tasks include eating and drinking from a cup followed by telephone answering, writing a message on a note pad and a napkin wipe. .

RESULTS

These are preliminary results based upon assessments fully completed in three subjects and partially completed in seven subjects.

1. Out of the three completed functional evaluations, two could not complete the Jebsen Test without the use of their FNS system. The third subject had nearly identical results with and without the use of the FNS system. Two candidates showed significant improvement of function with their FNS hand systems during the Isolated Basic Task section. The third subject did not have a significantly different score. However, this subject had difficulty with the proficiency test and scored poorly with and without the use of their FNS hand system.

2. The quality of grasp correlates well with quality of functional scores; i.e. the better the grasp pattern, the higher the test score.

3. Some of the major problems subjects encountered when completing both the Jebsen and the Basic Task portion without the use of their FNS systems include inability to actively open their hand and release the object they were holding. The subjects also experienced lack of adequate strength during grasp and inability to obtain finger tip pinch. Problems encountered when administering the functional evaluation section while using the system include inadequate training, fatigue due to lack of electrical exercise, and poor strategic approach towards accomplishing tasks.

4. No conclusions can be drawn as yet from the sensory and motor portions of the assessment

System characteristics which seemed to hinder the performance of the user of the system were:

1. Slow response time in switching from one mode of grasp to another,
2. Any control mechanism which required the subject to manipulate a slide potentiometer or similar device mounted to a lap board eliminated bilateral hand use,

3. The four channels of stimulation per grasping mode at times limited the effectiveness of the subjects hand grasp.

CONCLUSION

Observations regarding performance on the Jebsen Hand Function Test as opposed to the Isolated Basic Tasks Tests demonstrate that the tasks in Jebsen Tests alone do not necessarily apply to the various

functional requirements of C5 or C6 quadriplegic individual. For this reason we find it necessary to utilize the Functional Task Section.

Each subject performed each test to the best of his ability. Yet the need for a formal integration of the system into an occupational therapy program is becoming apparent to our research group. Our laboratory is developing a training protocol for both the therapist and the users of the FNS hand systems.

Our laboratory has also considered a hybrid approach towards the C5 and C6 quadriplegic utilizing FNS, conventional orthoses and adapted equipment, and surgical techniques such as tendon transfer. These aspects will need to be integrated into our clinical assessment in order to determine the best therapeutic modality for each subject.

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SUBJECT PREFERENCE FOR PULSE FREQUENCY WITH CUTANEOUS STIMULATION OF THE QUADRICEPS

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ABSTRACT

One factor limiting the clinical use of neuromuscular electrical stimulation (NMES) has been the discomfort associated with it. The purposes of this study were: 1) to determine whether a difference in subject comfort existed during NMES with stimulus frequencies of 30, 50 and 100 pulses per second (pps); and 2) to quantify how strongly one frequency was preferred over another. Twenty-one females between the ages of 22 and 33 received NMES to their right quadriceps femoris at amplitudes eliciting torques of 20 $\pm$ 1 ft-lb. The results indicate that high frequencies are preferred over low frequencies, especially by inexperienced subjects (100%). Experienced subjects tended to prefer high frequencies, however, they demonstrated some variability in preference (70%). Two comparison trials were made at varying levels of isometric torque, and it was found that 100 pps was preferred over a wider range of torque values than 50 pps when compared with constant torque values of 30 pps.

INTRODUCTION

Neuromuscular electrical stimulation (NMES) is a valuable treatment technique in the rehabilitation of patients with a variety of orthopedic and neurologic disabilities. Some patients receiving NMES, however, are unable to tolerate adequate stimulus amplitudes or treatment durations because of discomfort. It is therefore important to minimize the causes of discomfort to increase the number of individuals willing to use NMES in a rehabilitation program.

Previous studies at our Center have investigated the effects of waveform parameters on comfort (1). The purpose of the current study was to determine the stimulus frequency that minimizes discomfort during electrically-induced muscle contractions. The specific goals were to: 1) determine which of three stimulation frequencies, 30, 50 or 100 pps, was most preferred when compared at the same level of isometric torque; and 2) to quantify how much the amplitude could be varied before the initially preferred frequency was rejected in favor of the initially non-preferred frequency. The latter aspect would determine how strongly a frequency was preferred when compared to the non-preferred frequency.

BACKGROUND

Studies on the ability of human subjects to perceive and tolerate alternating currents date back to 1910, when Kennelly and Alexanderson found that sensory tolerance to current increased with rising frequency (2). Several authors have determined that for sinusoidal current the threshold of perception rises as frequency is increased above 300 Hz (3,4,5). Vodovnik et al studied pain responses of ten normal subjects to different tetanizing currents and concluded that sinusoidal frequencies greater than 300 Hz were more comfortable than low frequencies (6). Frequency did not seem to be a significant factor when stimulating with pulse trains, and he therefore suggested frequencies of 20-40 pps to minimize battery drain. Notermans found that stimulus frequencies of less than 10 pps caused sensations described as "tapping" or "pulsating" (7). With frequencies above 400 pps, the subjects described sensations of "vibration" or "tickling."

METHODS AND MATERIALS

Twenty-one female subjects between the ages of 22 and 33 years participated in this study. Thirteen were termed "experienced subjects" because they were either given cycled stimulation prior to test sessions or they were trained instructors in the use of electrical stimulation. The others were termed "naive subjects".

Subjects were seated in a test chair with hips flexed at 120° and knees flexed at 60°. Isometric torque was measured with a torque transducer designed and built by our Center. A microprocessor-based electrical stimulator also designed by our Center was used to deliver 300 microsecond symmetrical biphasic pulses. Stimulus amplitude and frequency were adjusted in each session as described below. Commercially available carbon impregnated silicone rubber electrodes, 5 x 10 cm in size, were used with saline-free conductive gel.

Testing was completed in three to four sessions on separate days, with test sessions at least twenty-four hours apart. All subjects completed the three sessions within two weeks.

Session I: Electrode locations were determined and marked with semi-permanent ink to ensure consistent electrode placement throughout the study. An orientation trial consisting of 12 pairs of pulse trains of either 30 & 50 pps, 30 & 100 pps, or 50 & 100 pps was conducted. Each pulse train was 1.5 seconds long, with a 0.75 second rise time and 0.75 second plateau contraction time. The two trains in each pair were 15 seconds apart, and pairs of pulse trains were separated by a 60 second rest period. The stimulus amplitude was adjusted for each frequency to produce a torque from the quadriceps muscle of 20 ft-lb, a value that is approximately 15% of voluntary quadriceps muscle contraction. The subject was asked to state which of the two pulse trains was the most comfortable. The question used each time was "Did you prefer stimulus one, stimulus two, or neither stimulus?"

Session II: The same protocol used in the Session I orientation trial was used in three trials of the following pairs of frequencies: 30 & 50 pps, 30 & 100 pps, and 50 & 100 pps. At

least 12 pairs were tested on each trial. Additional pairs were tested, if necessary, to obtain five pairs meeting the following criteria:

1. Peak torques produced by both pulse trains in a pair were equal to 20 ± 1 ft-lb.

2. Difference in torques produced by each of the pulse trains in a pair were less than 1 ft-lb.

3. The torque tracings from the strip chart recorder plateaued during peak stimulation. Traces demonstrating continued recruitment or spikes of torque were disallowed even though the torque value may have met the first two criteria.

These criteria were imposed to ensure that the amplitude and quality of the muscle contractions being compared were nearly identical.

For each of the three trials of Session II, the five pairs which met all criteria with the lowest differences in torque were considered "Selected Pairs". Only Selected Pairs were used in the determination of "Significant Preference". Significant Preference indicated that the subject favored one particular frequency in at least four of the five Selected Pairs. Any other distribution of Selected Pairs indicated a "No Significant Preference".

Session III: This session consisted of two comparison trials, at 30 & 50 pps and 30 & 100 pps. In each trial, the peak torque output of the 30 pps pulse train was held constant at 20 ft-lb. If 30 pps was the preferred frequency in Session II, the amplitude of the high frequency train was decreased until the high frequency was preferred in three successive comparison pairs. If the high frequency was preferred in Session II, then the high frequency amplitude was increased until 30 pps was preferred in three successive comparison pairs.

RESULTS

In the comparison of 30 & 50 pps, 13 of 21 subjects completed the comparison session with five Selected Pairs. Of these, 11 of 13 showed a Significant Preference for 50 pps (Table 1). Five of 8 naive subjects completed the session with five Selected Pairs and all 5 showed a Significant Preference for 50 pps. Of the 13 experienced subjects, 8 completed the comparison run with five Selected Pairs, and 6 of the 8 showed a Significant Preference for 50 pps. One experienced subject showed a Significant Preference for 30 pps and another had a No Significant Preference.

In the comparison of 30 & 100 pps, 11 of 21 subjects completed the comparison session with five Selected Pairs. Of these, 9 of 11 subjects showed a Significant Preference for 100 pps (Table 1). Five of 8 naive subjects completed the session with five Selected Pairs, and all 5 subjects showed a Significant Preference for 100 pps. Of the 13 experienced subjects, 6 completed the comparison run with five Selected Pairs. Four showed a Significant Preference for 100 pps, while two showed a Significant Preference for 30 pps.

TABLE 1. Summary of Significant Preferences for Each Comparison Trial

	Selected Pair Preference	Number of Subjects
30 & 50 pps	30 pps	1
	50 pps	11
	NSP	1
30 & 100 pps	30 pps	2
	100 pps	9
	NSP	0
50 & 100 pps	50 pps	2
	100 pps	7
	NSP	0

NSP = No Significant Preference

In the comparison of 50 & 100 pps, 9 out of the 21 subjects completed the comparison session with five Selected Pairs. Of these, 7 of the 9 subjects showed a Significant Preference for 100 pps (Table 1). Two of 8 naive subjects completed the session with five Selected Pairs and both showed a Significant Preference for 100 pps. Seven experienced subjects completed the session with five Selected Pairs. Five showed a Significant Preference for 100 pps, while 2 showed a Significant Preference for 50 pps.

Since five pairs meeting data criteria were not obtained in all trials, all of the pairs meeting criteria were examined as a group (Table 2). There was a total of 105 pairs meeting criteria for the 30 & 50 pps comparison. In all, 30 pps was preferred in 13 pairs (12%), 50 pps was preferred in 87 pairs (83%), and no preference was

TABLE 2. Summary of Preferences for All Pairs Meeting Criteria

	Selected Pair Preference	Number of Pairs	Percent
30 & 50 pps	30 pps	13	12.4
	50 pps	87	82.8
	NSP	5	4.8
30 & 100 pps	30 pps	19	20.0
	100 pps	76	80.0
	NSP	0	0
50 & 100 pps	50 pps	16	19.5
	100 pps	65	79.3
	NSP	1	1.2

NSP = No Significant Preference

selected in 5 pairs. Out of a total of 95 pairs meeting criteria for the 30 & 100 pps comparison, 30 pps was preferred in 19 pairs (20%) and 100 pps was preferred in 76 pairs (80%). In the 50 & 100 pps comparison, 82 pairs met the established criteria. In this case, 50 pps was preferred in 16 pairs (20%), 100 pps was preferred in 65 pairs (79%), and there was one No Significant Preference.

In Session III, the stimulus amplitude of the higher frequency was changed in a stepwise manner until the subjects switched frequency preference. Subjects were grouped according to those who initially preferred the high frequency (50 or 100 pps) and those who preferred 30 pps. For those who initially preferred 50 pps in the 30 & 50 pps comparison, stimulus amplitude had to be increased until, on the average, torque was 31.1 ft-lb or 55.4% above the torque at 30 pps before they switched preferences. For the one subject who initially preferred 30 pps, torque at 50 pps had to be decreased by 27%. In the comparison of 30 & 100 pps, subjects who initially preferred 100 pps (15 of 19 subjects) switched preferences when torque was increased by an average 75.7%. The four subjects who initially preferred 30 pps switched preferences when torque was decreased by an average 48.7%.

DISCUSSION

Generally speaking, subjects preferred the higher frequency in each comparison pair. Referring to Table 2, it is seen that when there was a preference, the higher frequency was selected in 228 of 276 comparison pairs (82.6%). In Session III it was shown that, in those who preferred high frequencies in constant torque comparisons, considerably more torque could be generated at the high frequency compared with 30 pps while still maintaining the same "comfort level". For example, in comparing 30 and 100 pps, torque at 100 pps could be increased by an average of 43.1% before subjects began to ever select 30 pps as most preferred. Not until torque at 100 pps was increased by an average of 75.7% did subjects consistently select 30 pps.

It is generally recommended that frequencies of about 30 pps be used in clinical applications of cutaneous stimulation because electrically-induced contractions fatigue much more rapidly as frequency is increased (8). From this study, increasing frequency seems to be a way to obtain stronger muscle contractions in patients for whom discomfort is a limiting factor, but there is necessarily a trade-off between stronger contractions and more rapid fatigue.

Very little quantitative data on muscle fatigue during cyclical stimulation is currently available to the clinician. Without these data, an intelligent decision regarding which frequency to use in specific clinical applications cannot be made. A series of studies are currently underway at our Center to provide more information on cyclical stimulation. The first of these, using cyclical stimulation at 100 pps, has been completed and is described in the following paper (9).

CONCLUSIONS

1. In comparisons of 30, 50 and 100 pps stimulation, the higher frequency was generally preferred (83%).

2. Naive subjects preferred higher frequency stimulation with remarkable consistency (100%). Experienced subjects displayed variability in their preferences but still preferred higher frequencies more often than not (70%).

3. Significantly higher torques can be obtained at the same "comfort level" with higher frequencies.

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18.3

THE EFFECT OF DUTY CYCLE ON MUSCLE FATIGUE USING A STIMULATION FREQUENCY OF 100 PULSES PER SECOND

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ABSTRACT

Very little quantitative data are available concerning the effect of duty cycle on muscle fatigue. A previous study indicated that many subjects prefer electrical stimulation at a relatively high frequency, 100 pps, when compared with stimulation at lower frequencies, i.e. 50 and 30 pps. The purpose of this study was to assess the effects of four duty cycles, 1:3, 1:5, 1:7, and 1:10, on the quality of electrically stimulated muscle torque while using a frequency of 100 pps. Ten female subjects received electrical stimulation to their right quadriceps femoris at amplitudes eliciting torques of 20 ± 2 ft-lbs for 30 minutes of cycled stimulation. The results indicate that even the least aggressive duty cycle tested, 1:10, caused significant reduction in stimulated torque 10 minutes into the stimulation session. Stimulation at the duty cycle of 1:3 caused significant fatigue after only 2 minutes. If the purpose of a treatment program is to maintain a strong stimulated contraction for more than 10 minutes, and a frequency of 100 pps is used to enhance patient comfort, a duty cycle that is less aggressive than 1:10 will be necessary to optimize the therapeutic outcome.

INTRODUCTION

Neuromuscular electrical stimulation (NMES) is a valuable tool in treating patients with a variety of orthopedic and neurologic disorders (1). In the previous paper we identified that subjects perceived a frequency of 100 pulses per second (pps) was more comfortable when compared to lower frequencies (2). The effect of duty cycle on muscle fatigue has not been assessed at this relatively high frequency of stimulation.

Many treatment programs require stimulation of 30 or more minutes to achieve adequate therapeutic effects (1). In order to maintain quality muscle contraction throughout a 30 minute treatment time, electrical stimulation duty cycles must be relatively non-aggressive. However, long periods of no stimulation between each cycle reduces the amount of actual contraction time, and potentially the therapeutic benefits of the program. Thus a balance must be struck between too much and not enough OFF time, to ensure continued quality muscle contraction while maximizing the total amount of stimulated contraction time. The purpose of this study was to determine the effects of four duty cycles on the fatigue of a stimulated muscle contraction, using a stimulus frequency of 100 pulses per second.

BACKGROUND

To date few studies have specifically investigated the effects of duty cycle upon muscle fatigue. Using a frequency of 30 pps, Yergler, as reported by Benton et al, found little or no evidence of muscle fatigue using a duty cycle of 1:5. A duty cycle of 1:5 has also been used with a stimulation frequency of 50 pps during a short treatment session and was found to be adequate to avoid fatigue (3). When a 10 ms burst of a 2,500 Hz sinusoidal signal repeated every 20 ms was used to contract the muscle, several authors have reported that a 1:5 duty cycle is adequate (4-6). No studies evaluating the effects of duty cycles with frequencies of 100 pps have been found in recent literature.

All of these studies of duty cycles have been done over relatively short sessions of 10 to 15 minutes or less (1,3-6). Studies of short duration assessments may be relevant to the high intensity, short duration strengthening programs typically used in sports medicine treatments, but cannot reliably be used to predict performance on a more traditional 30 minute or longer stimulation program. To make appropriate clinical recommendations in using various duty cycles at the higher frequency of 100 pps, a treatment time of at least 30 minutes must be assessed.

METHODS AND MATERIALS

Ten normal female subjects between the ages of 23 and 32 were evaluated in this study. Subjects were seated in a test chair with their hips flexed to 120° and their knees flexed to 60°. Isometric torque was measured with a torque transducer designed and built at our Center. Electrical stimulation was provided by a commercially available constant current unit (Ultrastim by Neuromedics). The stimulator was programmed to deliver 300 microsecond symmetrical biphasic pulses at a frequency of 100 pps. Commercially available carbon impregnated silicone rubber electrodes, 5 x 10 centimeters in size, were used with saline-free conductive gel.

Testing was completed in four sessions on separate days, with test sessions at least 24 hours apart. Each test session was used to evaluate a different duty cycle. Stimulation ON time was set to 7 seconds for all sessions, including a 2 second rise time, 4 second plateau time and a 1 second fall time. OFF times were determined in multiples of the stimulation plateau time. Four different duty cycles were evaluated including 1:3, 1:5, 1:7, and 1:10. The duty cycles were evaluated in a random sequence for each subject.

During the first session electrode locations were determined and marked with semi-permanent ink, to ensure consistent electrode placement throughout the study. In all sessions, stimulus amplitude was adjusted to produce a torque of 20 ft-lbs from the quadriceps muscle, a value that is approximately 15% of volunatry quadriceps muscle contraction. After appropriate stimulation amplitudes were determined for each subject, no further modifications were made. This was to assure that

each subject received a consistent stimulus amplitude throughout the 30 minute assessment period. Torque measures of each contraction were recorded from an electronic digital display. After every session the shape of each stimulated torque curve was inspected to ensure that a plateau of muscle tension was reached, and that the recorded torque from the digital display accurately represented that plateau. Two minute marks were made during the sequential recording of electrically stimulated muscle contractions. Recordings continued until 30 minutes of stimulation had been completed, or until the subject demonstrated a torque of less than 5 ft-lbs during five consecutive contractions.

A two-way analysis of variance with repeated measures design was used to evaluate the quality of muscle contraction of each subject at each two minute interval across the four duty cycles tested. Tukey's HSD post hoc tests were used to make specific comparisons of stimulated torque production within and between duty cycles. For all comparisons, statistical significance was set to $p < 0.05$.

RESULTS

Intra-Duty Cycle Comparisons

During electrical stimulation with the 1:10 duty cycle, a statistically significant decrement in the torque elicited by the electrically induced contraction was seen shortly after the eighth minute. Between the first and tenth minute the stimulated torque output decreased by nearly 15%. Torque continued to decrease an additional 18% during the ensuing 20 minutes of stimulation. The mean torque of the 10 subjects recorded after 30 minutes of stimulation at the duty cycle of 1:10 was 13.5 ft-lbs, or almost 68% of the initial torque.

When the 1:7 duty cycle was evaluated, stimulated torque decreased by a statistically significant amount during the first four minutes of the assessment. The decrement seen in stimulated torque produced between the first and tenth minute with the 1:7 duty cycle was nearly 28%. During the final 20 minutes of stimulation the torque decreased by another 13%. The average torque recorded after 30 minutes of stimulation was 60% of the original 20 ft-lbs.

The duty cycle of 1:5 demonstrated slightly more rapid fatigue, with significant decrements in torque first recorded between the second and fourth minute of stimulation. Stimulated torque at ten minutes was slightly less than 25% of the torque produced in the first minute of stimulation. The final 20 minutes of stimulation resulted in a further decline in torque production of about 13%. The mean torque produced at the end of the 30 minute assessment was slightly greater than 12 ft-lbs.

A significant decrement in torque was seen within the first two minutes of stimulation when the 1:3 duty cycle was tested. This decline was more than 20% in the first minute of stimulation. Torque production declined a total of 41% during the first ten minutes of stimulation. Only minor changes in torque production were noted over the next 20 minutes of stimulation. The final torque level at the end of the assessment session showed a decrease of 64%, averaging just over 9 ft-lb for the ten subjects.

Inter-Duty Cycle Comparison

The patterns of decrement in the stimulated torque production were significantly different when comparing the duty cycle of 1:10 to that of 1:3. There was a 20% difference between the two duty cycles, which occurred within the first two minutes of stimulation and continued throughout the 30 minute session. A similar degree of fatigue, as indicated by reduced torque production, was noted at 30 minutes of stimulation using the 1:10 duty cycle when compared to six minutes of stimulation with the 1:3 duty cycle. In the same manner, a similar degree of fatigue recorded after 30 minutes of stimulation with the 1:7 and 1:5 duty cycles was seen after the eighth minute of stimulation with the 1:3 ratio. The 1:7 and 1:5 duty cycles demonstrated very similar decrements in stimulated torque levels, and were not significantly different from each other, or from the other two duty cycles tested.

DISCUSSION

These data indicate that when a frequency of 100 pps is used, the least aggressive duty cycle studied, 1:10, was not adequate to avoid significant fatigue in healthy young adults. In patients who display disuse atrophy, even more rapid fatigue might be anticipated (1). The previous paper indicated that the relatively high frequency of 100 pps was preferred by subjects as being significantly more comfortable than either 50 or 30 pps (2). This might lead the clinician to use the relatively high frequency for initiation of electrical stimulation programs, potentially resulting in a decrease in the training period required to establish adequate sensory tolerance to achieve therapeutic levels of stimulated muscle contractions.

Based on the findings of this study, however, extremely low duty cycles of less than 1:10 may be necessary to provide adequate rest time between stimulus trains, in order to avoid muscle fatigue during a treatment program lasting for more than 10 minutes. Many treatment programs which use treatment time as a guideline for the adequacy of the program, have been based on more aggressive duty cycles, often 1:3 or 1:5 (1,3,5,6). In order to ensure the efficacy of a treatment that is based on a very low duty cycle, i.e. 1:10 or less, a more prolonged treatment time will undoubtedly be necessary. Because the total number of contractions will be drastically reduced in a 30 minute treatment program based on a non-aggressive duty cycle, the actual length of the treatment will have to be increased to ensure the adequacy of the treatment. Thus in programs aimed at increasing or maintaining range of motion, or managing spasticity of an antagonist muscle group, 30 minute treatment programs may have to be extended to one hour programs, or more.

An alternative to the extended treatment time may be a program that uses the relatively high frequency of stimulation only during the training period. Subsequent reduction of the frequency as the patient increases in sensory tolerance and achieves stronger muscle contractions would reduce the need for the prolonged rest periods between stimulus trains. This may allow the clinician the flexibility to rapidly train the patient and still use the traditional guidelines for the development of adequate treatment programs to ensure the efficacy of the stimulation regimen.

Finally, some treatment programs are aimed at producing muscle fatigue through electrical stimulation. These programs typically include the goal of reducing acute muscle spasm or neurogenic spasticity (1). If production of muscle fatigue is the goal of the treatment program, the use of 100 pps at a duty cycle of 1:3 will enhance fatigue while avoiding the problems associated with continuous tetanizing stimulation. Our results indicate that the same degree of fatigue can be achieved with six minutes of aggressive stimulation (1:3), when compared to 30 minutes of relatively non-aggressive stimulation (1:10). Thus the high frequency, aggressive duty cycle may allow the clinician more effective use of treatment time when muscle fatigue is the goal of the therapeutic program.

CONCLUSION

1. In comparing duty cycles of 1:3, 1:5, 1:7 and 1:10 at a frequency of 100 pps, significant reduction of muscle torque was noted during the first 2 to 10 minutes of stimulation, indicating significant fatigue of the neuromuscular system.

2. If frequencies of 100 pps are to be used in traditional NMES programs, duty cycles of less than 1:10 will be necessary to avoid significant neuromuscular fatigue.

3. A frequency of 100 pps using a duty cycle of 1:3 would be an effective means of producing a rapid neuromuscular fatigue.

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STAGES IN THE DEVELOPMENT OF USEFUL FNS-AUGMENTED WALKING

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ABSTRACT

Starting with [1] definitions of levels of useful walking, [2] a review of past efforts to restore the ability to walk in paralyzed individuals, [3] characteristics of various injuries and [4] consideration of general health; the author discusses the ongoing development of a functional neuromuscular stimulation system. Choices of muscles for implantation with percutaneous intramuscular electrodes and the quality of the FNS-augmented walking achieved with those muscles are presented.

INTRODUCTION AND DISCUSSION

Providing useful mobility to paralyzed individuals is a major goal of the rehabilitation field and in recent years much attention has been focused on walking. However, there needs to be a clear definition of what actually constitutes useful walking and how to assess the value of a walking aid. The following definition of terms was proposed by Dr. J. Perry (4). The exercise walker is a person who is capable of reaching a standing position with or without assistance and advancing with or without assistance for a few feet. The energy utilization and the type of equipment and/or assistance needed are such that it is unreasonable to walk outside the home. Still, it is useful for increasing the aerobic state, strengthening of bones and muscles and giving the individual a psychological lift. The next level is that of the household walker. This individual can walk safely for short distances around the home and perform functional tasks. Such capabilities may allow access to areas of the home that are not wheelchair accessible. The use is limited to the home because of the complexity of the assistive equipment, energy levels required, or safety concerns. The third and highest level is the community walker, a person who walks to achieve activities of daily living in the community. Almost any individual can achieve at least one of these levels with enough assistance. The real question is, "Does the gain balance the necessary effort?" Only the individual patient can answer this for himself.

Initial approaches to achieving walking in paraplegics included walking frames to increase stability through a larger base of support; bracing to stabilize uncontrollable joints; and training both to increase the aerobic state in order to allow the needed extra effort to walk, and to develop new methods of balance and conservation of energy in the actual walking process. Some patients have been able to utilize these techniques and to achieve a functional level considerably above what was possible for them prior to intervention. Others have found considerable difficulty and many have remained essentially wheelchair bound. In general, adult paraplegic individuals with neurologically complete lesions above the T12 level have eventually chosen wheelchair mobility (6) over walking.

A review of both walking and non-walking paralyzed individuals shows motivation to be extremely important. Even a normal person cannot walk if he does not try. Severe brain damage, depression or unreasonable overprotection by the family are all factors which reduce the chances of walking. Another factor is the degree of voluntary control of the available muscles. Trying to assess the residual strength of an individual who is partially paralyzed is difficult. Spasticity may cover up potentially useful function or may fool the examiner into thinking function is present which is not. Manual muscle testing in functional positions and repeated observation still are the best means of understanding the true state of function in a particular individual (1).

Our experience with stroke, head injury, spinal cord injury, peripheral nerve injury, multiple sclerosis, polio and birth defects has led to several conclusions. First a polio patient who lacks only the motor system and has an intact sensory system is able to be far more functional than the mere measurement of the muscle strength would indicate. This indicates the value of non-visual feedback. Experience with meningomyelocele children would indicate that community level walking is rarely maintained into adulthood without hip control. Generally, the paralysis in the various conditions above proceeds from proximal to distal with the worst distal.

Obviously, the general physical condition of the individual is important; including the aerobic capacity, amount of body fat, and muscle mass. The amount of energy needed to achieve a particular function has also been found important and generally any functional activity requiring over 50% of the maximum aerobic capacity has not been continued for a long period of time, no matter how seemingly beneficial to the patient (6).

An informed approach to restoring walking in paraplegics by means of FNS depends on all the above considerations. Certainly the muscles are potentially available for stimulation in most patients; although the surgery necessary to instrument them, the reproducibility of function and the avoidance of fatigue still present problems. The group of muscles that we use has

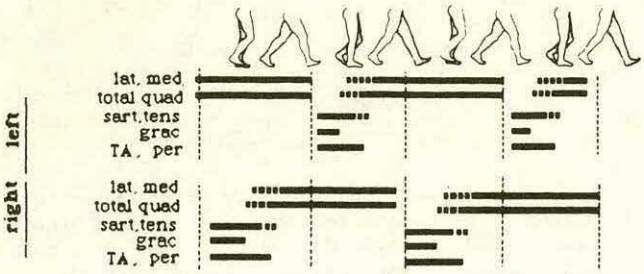

Figure 1: Left: Stimulation pattern (two repeats). lat=vastus lateralis, med=vastus medialis, total quad=four heads of quadriceps, sart=sartorius, tens=tensor fasciae latae, grac=gracilis, TA=tibialis anterior and per=peroneus longus.

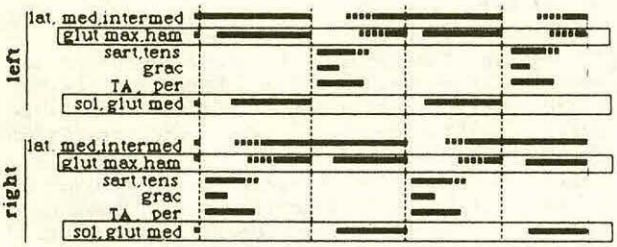

Figure 2: Left: Augmented stimulation pattern (two repeats). Additional muscles are outlined: glut max=gluteus maximus, ham=hamstring, sol=soleus, glut med=gluteus medius.

grown during the past three and half years as we have gained experience with six paraplegic subjects who use our FNS system at least three times weekly. Below we discuss the evolution of our choice of muscles.

Initially we used hip flexors (sartorius, gracilis and tensor fasciae latae), knee extensors (full quadriceps, vastus medialis and vastus lateralis) and dorsiflexors (combination of tibialis anterior and peroneus longus). Controlled by individualized stimulation patterns, these muscles allowed subjects to achieve a reciprocal gait, with support from parallel bars or a walker (2).

The problems were that this gait required much use of the arms in single leg stance. Additionally, the propulsion force came from the arms and transfer from one foot to the other was accomplished using the arms. This meant that: [1] Balance without arm support was nonexistent, precluding any functional hand use while standing or walking. [2] Most of the energy of walking was coming from the arms which are not designed for this, resulting in marked energy inefficency. [3] Walking speeds approaching normal [.7-1.5 m/s] could not be achieved (Figure 1).

Our solution to this was to add muscles, including hip extensors [hamstrings, gluteus maximus] and abductors [gluteus medius], and foot plantar flexors [soleus]. This resulted in significant improvement in walking speed

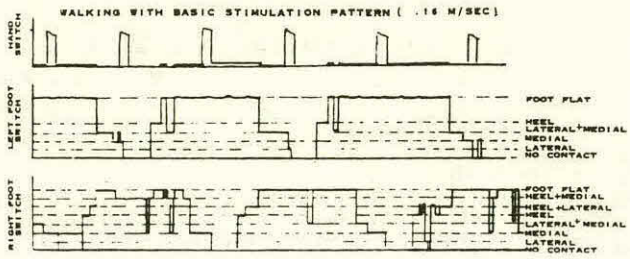

Figure 1: Right: Foot-floor contact and hand switch (operated by the paraplegic subject to initiate each step) data from a typical walking trial. Speed = .16m/sec.

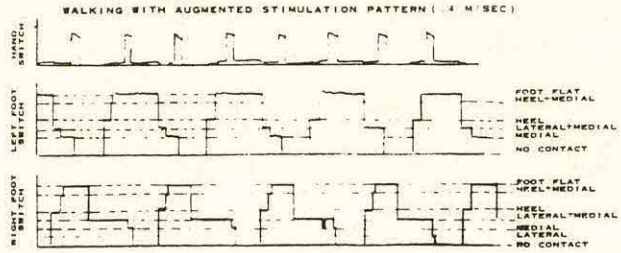

Figure 2: Right: Foot-floor contact and hand switch (operated by the paraplegic subject to initiate each step) data from a walking trial with augmented stimulation pattern. Speed = .4m/sec.

[maximum of .7m/s] (3) standing balance, and energy utilization (Figure 2). Two subjects were able to use axillary crutches and one of the two was able to use loftstrand crutches, all with difficulty. Even with these advances problems were again encountered. A situation previously unusual in the treatment of paralysis had been created: An individual with normal upper extremities and functionally near-normal lower extremities joined by a totally uncontrolled trunk segment. This was reflected in a long double support and inefficient weight transfer from one stance phase to the other. The individual was unstable during this time and required walker stabilization.

Others have proposed substituting conventional bracing for the trunk and knee and utilizing the FNS only for the reciprocal motion of the legs (5). While it is certainly possible to walk long distances with such devices, we doubt that they will be truly functional due to problems of donning and doffing, cosmetic appearance, difficulties in toileting, problems in getting up from a fall, restriction of movements and energy limitations. The combination of extensive bracing with FNS does however provide a means of exercise which will prove of value in maintaining bone strength and aerobic capacity.

We believe a part of the solution to the problem of the uncontrolled trunk segment will be the addition of the paraspinals, the quadratus lumborum, and the rectus abdominis

muscles at appropriate times during the walking cycle. The stimulation pattern for the paraspinals will be similar to that for the gluteus maximus; the pattern for the quadratus lumborum will follow that of the sartorius and tensor; and the rectus abdominis will be activated similarly to the soleus. The testing of this solution is currently underway. The general concept we have learned throughout this project is that one must closely approach natural muscle function, in terms of both the choice and number of muscles, if one is to provide truly useful walking.

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18.5

RESPONSE TO SURFACE ELECTRICAL STIMULATION OF THE QUADRICEPS IN INDIVIDUALS WITH SPINAL CORD INJURY

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ABSTRACT

Electrical stimulation of muscles paralyzed by spinal injury shows promise to be a valuable therapeutic adjunct to traditional rehabilitation therapies. Over the past two years, we have been studying a diverse group of male veterans who have paraplegia or quadriplegia. To date, 25 have participated in our program, with 20 undergoing from 3 to 8 weeks of twice-daily electrically induced exercise of the quadriceps. We have measured changes in selected physiological and psychological variables in these individuals, and have attempted to determine factors that can predict the physiological changes resulting from such exercise.
We have monitored changes in stimulated and voluntary muscle force and fatigability, spasticity, urodynamics, and psychological status brought about by participation in our protocol. We were able to increase the force and fatigue resistance of the thigh muscles of over half of our participants. Voluntary torque increased in 7 of 16 legs in individuals with incomplete quadriplegia. Increases in quadriceps spasticity and torque were most pronounced in recently injured paraplegics. Quadriceps stimulation had mixed effects on urodynamics. More than one half the patients tested before and after attempted reconditioning showed improved urodynamics; others showed no change or a worsening. Both positive and negative changes were seen in psychological status. Our preliminary findings suggest that the best results are achieved when stimulation is begun within 2 years post-injury and when initial stimulated torque exceeds 6 N-m.

INTRODUCTION

Electrical stimulation of muscles paralyzed by spinal cord injury shows promise to be a valuable therapeutic adjunct to traditional rehabilitation therapies. Over the past two years,we have been studying the application of electrostimulation to a diverse group of male veterans who have paraplegia or quadriplegia. Our goals have been to develop a set of measures that could predict how well a particular patient might respond to a 1 to 2 month period of electrically induced exercise and to measure what therapeutic benefits the patient might achieve by such exercise. To date, 25 patients have participated in our program, with 20 undergoing at least 3 to 8 weeks of twice-daily electrically induced exercise of the quadriceps. We have measured many physiological and psychological parameters in these individuals before, during and after attempting thigh muscle reconditioning. Our experiences have suggested the difficulty in quantifying "therapeutic benefits". We looked for changes in urodynamic status and in spasticity, but the results were mixed (see Results). It appeared that some positive changes were seen in the psychological status of our participants, but we need a larger sample size to make a definitive statement.

In dealing with potential predictive factors for reconditioning outcome, we have been investigating how such factors as time since injury, level of injury, extent of spasticity, residual voluntary muscle force, residual muscle mass, age, psychological status, and days of reconditioning might relate to the ultimate force and fatigability achieved in a muscle undergoing reconditioning. A trend in our data suggests that a factor like time since injury might be crucial in determining the extent of reconditioning (see Discussion). These trends might well indicate limits for selecting appropriate patients in which electrical muscle stimulation has a high probability of being effective. Intuitively, one might expect that all patients injured fairly recently (i.e., <10 yrs), and who have only upper motor neuron involvement, could undergo significant reconditioning of paralyzed muscle, deriving whatever benefits such reconditioning might produce. Unfortunately, our experience is that some patients in this category do not show appreciable improvement in muscle strength even after an extensive program of reconditioning. In fact, our preliminary findings suggest that the best results are achieved when stimulation is begun within 2 years post-injury.

No one has yet demonstated a satisfactory way to predict the outcome of reconditioning programs, or for that matter, what constitutes an optimal reconditioning protocol. A lack of reliable predictive criteria can result in unrealistic hopes (or alternatively prejudices) by the patient, therapist or physician. Finally, any use of electrical stimulation for functional purposes (such as walking or standing) requires that the appropriate muscles first be reconditioned. Knowing what factors are important for reconditioning, and how best to achieve reconditioning, thus can serve as guidelines for selection for functional electrical orthoses.

METHODS

All subjects received medical permission to participate in the study and gave informed consent to an institutionally approved protocol. As part of this study, we have measured the variability of initial baseline measures of muscle condition (torque & fatigue) and have compared this variability to lesion level, completeness and time-since-injury. Our participants to date have a bimodal age distribution (late 20's and mid 50's), and were almost equally divided into groups of 6 months or less post-injury, 6 months to 3 years, and 3 to 16 years. For the most part, these individuals had incomplete quadriplegia or complete paraplegia at T12 or above. Of the 25 subjects enrolled, 20 continued participation in the reconditioning phase of our protocol and received from 3 to 8 weeks of stimulation.

EMG and Spasticity Assessment

We used a standard clinical electromyographic workup to verify the completeness of the lesion and whether there was any evidence of lower motor neuron involvement. All patients exhibited some positive sharp waves and occasional or marked fibrillation potentials in one or more major muscle groups of the legs. Such waves and potentials normally would be taken to indicate lower motor neuron damage; however, their ubiquitousness for all patients with spinal trauma (2,9,10) indicates that a pure upper motor neuron lesion for spinal injury is probably very rare. Further, verification of lesion completeness (excluding cases with complete transections) is complicated by remnant motor pathways that can subconsciously influence lower limb reflex behavior in an otherwise "complete" individual (3,4). These influences manifest themselves during Jendrassik and other manuevers. We have in this study relied on defining "completeness" in the neurological sense (no volitional movement and no interference activity).

The pendulum drop test of Wartenburg as refined by Bajd and Bowman (1) was used in our study to measure spasticity. The subject is placed supine on a gurney with both legs free to swing at the knee without obstruction. A therapist lifts the leg manually to full extension, then drops it, while an electrogoniometer measures knee range of motion during the subsequent oscillatory motion. This test is repeated 10 times, at 30 second intervals, for each leg, the electrogoniometric output recorded on a polygraph. The normalized relaxation index was computed as:

$$R2n = \frac{\text{(first flex angle - starting angle)}}{(1.6)\text{(resting angle-starting angle)}}$$

For individuals with normal muscle tone, this value tends to be close to 1.0, while lower values indicate greater degrees of spasticity (1).

Measurement of Torque

Response to electrical stimulation of muscle was determined immediately after assessing spasticity, while the subject was still positioned on the gurney. We placed 2" by 4" carbon rubber electrodes with affixed water soaked sponge pads on the midline of the thigh over the quadriceps, with the indifferent electrode centered 3cm above the patella, and the active electrode 10 cm above the indifferent. The reference stimulus was a charge balanced pulse train of 100 mA amplitude, 20 Hz frequency and 400 µsec pulse width, with these parameters continually being monitored with a battery operated oscilloscope. During the first test session, we determined sensory and motor threshold currents, and a torque recruitment curve for each leg. Current levels of 20 to 160 mA were applied in ascending order (20 mA per step, with 3 or 4 stimuli given at each level). If an individual had any response to this preliminary test, we then fixed the stimulus amplitude at 100 mA and alternately stimulated each leg for 2.5 seconds over a 20 minute test period. During this time we measured the peak torque produced and monitored the decline of this torque in each leg. This was repeated in two subsequent test sessions, each a day or more apart. We determined three relative measures of fatigue: the duration of stimulation when the torque declined to 50% and 20% of peak, and the percentage of peak torque remaining after 20 min of stimulation.

Additional Evaluations

Urodynamics, peripheral vascular status and psychological factors were also examined.

RESULTS

Baseline Results

We found the spasticity index (R2n) initially to be centered around 0.9 (range=0.7 to 1.0) for individuals with complete paraplegia and around 0.7 (range=0.5 to 1.1) for individuals with incomplete quadriplegia.

During the first baseline assessment with 100 mA, the peak torque produced exceeded 10 N-m in each leg for 7 individuals (2 with complete paraplegia, 4 with incomplete quadriplegia, and 1 with complete quadriplegia) and in one leg for four (2 with paraplegia, 2 with quadriplegia). The majority of torques measured during this initial session were less than 10 N-m but greater than zero. In two individuals, no torque was produced in the left leg. Overall, the largest torques were produced by individuals with incomplete quadriplegia with 6 months or less time since injury.

The peak torques measured during the subsequent 1 or 2 baseline evaluations often varied widely from those initially measured, especially for individuals with less than 3 years-post-injury. The size of variations in peak torques between the 3 baselines was generally below 15 N-m, but exceeded 20 N-m for two individuals. However, in terms of percentage change, these variations could be large indeed. All 7 individuals with complete paraplegia exhibited at least one peak torque in either leg that was either twice (n=3) or half (n=4) that observed in the first baseline. Five individuals with quadriplegia (4 incomplete, 1 complete) had a doubling or halving in peak torque for at least one leg during subsequent baseline evaluations. However, most individuals who exhibited less than a factor of two change had incomplete lesions (13 legs with quadriplegia, 2 with paraplegia).

Torques at the end of a 20 minute evaluation period using 100 mA stimuli rarely exceeded 7.5 N-m for any individual. These "end torques" usually were 20 to 30% of initial peak torque for those with quadriplegia and less than 20% for those with paraplegia. In any given test session, the torques produced by the cyclic stimulation generally fell to 50% of the initial peak torque within 0.5 to 5 minutes. Three individuals with incomplete lesions could maintain greater than 50% of peak force for longer than ten minutes. Across all baseline tests, the thigh muscles of individuals with incomplete quadriplegia still were able to produce greater than 20% of initial peak torque after 20 minutes of exercise in half the legs studied. In contrast, on the majority of tests torques fell to 20% within 10 minutes in those individuals with complete paraplegia and all other lesion types tested.

While we did not search for optimum motor points in any subjects, we did use a standard placement of large surface electrodes for all subjects across all tests, which should have yielded consistent results (6). The observation that those with incomplete lesions had less variability in stimulus- elicited peak torque and fatigue was unexpected in view of their day-to-day variability in volitional control. Further, we had expected individuals with complete paraplegia to produce fairly consistent results, in keeping with isolated spinal control mechanisms. Until the physiological basis for this variability is better understood, caution is in order when assessing the status of paralyzed muscle from a single observation.(5)

Changes Brought About by Reconditioning

Changes in torque. For those eight paraplegics with complete injury who underwent reconditioning, six had bilateral increases in peak torque; and two had bilateral decreases (Figure 1). One of the two paraplegics with incomplete injury was extremely spastic(R2n= 0.22) and exhibited a slight mixed increase and decrease in peak torque; the other had a slight decrease in one leg and no change (as expected) in the leg with lower motor neuron involvement. For the nine quadraplegics with incomplete injury who underwent reconditioning, six showed bilateral increases in stimulated peak torque, two had bilateral decreases, and one was mixed (Figure 2). Eight had changes in voluntary force production brought about by reconditioning. Four had bilater- al increases in voluntary knee torque, three had bilateral decreases and one showed a mixed change. One quadraplegic with a complete injury (16 years post injury and with marked flexor spasticity) had no change from the initial trace stimulated torques even after eight weeks of stimulation.

The torque that a muscle was able to produce after 20 min of cyclic exercise was also measured. We termed this torque the "end torque", and it represented one measure of the fatiguability of the muscle. Except for two individuals, all subjects had higher end torques after their reconditioning, indicating that the endurance of the muscle was being built up. As a further measure of fatigability, we determined the time when the torque declined to 50% of peak torque. It too was almost universally elevated, further indicating increased endurance.

Spasticity changes. The spasticity index R2n was also affected by the reconditioning protocol, especially for paraplegics who showed the largest increases in peak torque (Figure 3). These individuals also reported that their knee spasms were stronger, but occurred less often. Such an increase in spasticity was confounded by the fact that these individuals all were less than 6 months post-injury, and might have normally had increasing spasticity. Yet, individuals with incomplete quadriplegia who did restrengthen did not necessarily have increases in R2n (Figure 4). There was no apparent correlation so far between changes in spasticity and days of exercise.

Other findings. Seven patients that underwent 4 or more weeks of reconditioning had pre- and post-reconditioning CT scans taken at exactly the same distances between the greater trochanter and patella, thus permitting analysis of cross sectional changes (Figures 5,6). Some patients had increases of 5 or more sq cm, while others exhibited little change. There appeared to be little correlation so far between changes in quadriceps cross section and changes in force, although muscle being converted from fast to slow twitch can show a size decrease and have less fatty inclusions.

The ability of peripheral electrical stimulation to affect urodynamics was mixed. Slightly more than one half the patients tested before and after attempted reconditioning showed improved urodynamics; others showed no change or a worsening. Peripheral vascular status was for all practical purposes unchanged by our reconditioning intervention. The most consistent changes were seen in psychological status (see Discussion).

Predictive Factors

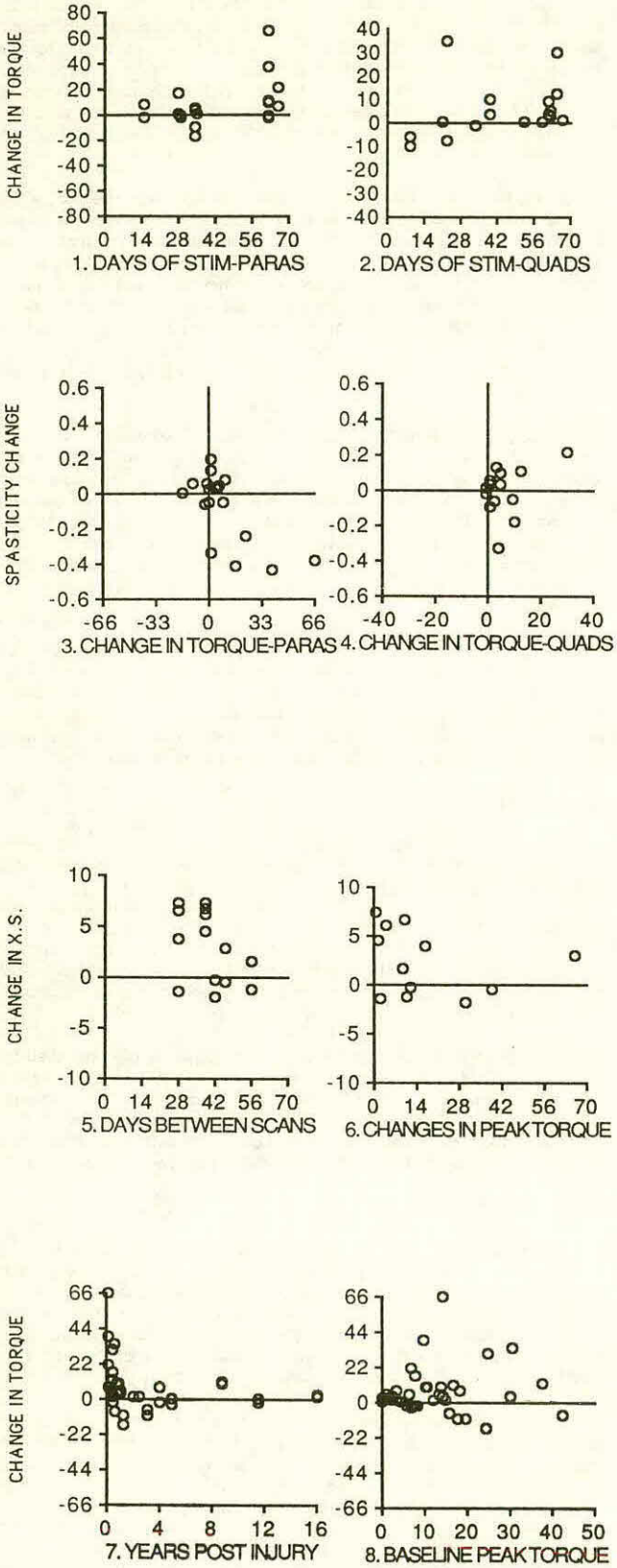
Perhaps our most striking preliminary finding to date is that all the patients in which we saw marked changes in peak stimulated torque were within a year or two post injury and had initial peak torques (to a 100 mA test stimulus) above 6 N-m (Figures 7,8). This finding still appears to hold even if we adjust for the days of stimulation or for the initial baseline peak. In fact, those with paraplegia of less than a year's duration often had initial baseline torques below that shown by patients whose injury was of longer duration. Legs with initial peak torques below 6 N-m showed no increase in peak torque, even if they were stimulated for from 21 to 66 days.

We had originally thought that the amount of spasticity (as quantified by R2n) might serve as a predictor. This has not so far been the case, as almost all paraplegics had R2n's between 0.8 and 1.0, and the quadriplegics, between 0.65 and 0.9. Changes in peak torque did, however, correlate with changes in spasticity (see above). The relationship between initial muscle mass and ultimate force production is now being analyzed.

Psychological Factors

Psychological factors have generally been overlooked when assessing the benefits (or risks) achieved by use of a prescribed regimen of therapeutic electrical stimulation of muscles paralyzed by spinal cord injury. Such factors include the patient's self-image, moods and his perception of the forces that control his life or health. Previous studies have demonstrated that individuals with spinal cord injury frequently have diminished self-concepts and dysphoric moods subsequent to their injury (11).

Some interesting trends were observed in these measures following participation in our protocol (5,7). Completion of the protocol did not influence subject's levels of self concept nor alter their locus of health control. However, participation in the protocol positively influenced subjects' response set on the Tenessee Self Concept Scale by decreasing conflict measures between various aspects of the self concept. The measures of mood states reflected feelings during the last week of our protocol. The group as a whole reported increases in tension, anger and fatigue. Since protocol termination frequently coincided with pending discharge from the hospital, we are currently attempting to differentiate between the influence of protocol termination and/or hospital discharge. These results suggest both state measures of emotion and attitudinal/trait



measures like stability of self concept can be influenced by rehabilitation efforts employing rehabilitation technology to impact upon the behavioral consequences of SCI. Perhaps the greatest difference in psychological status was uncovered during the clinical interview held after program participation. Almost all patients reported satisfaction with participation in the study, even though at least half had no measurable increase in stimulated knee torque production. Many wished to continue participation past the prescribed time limit and expressed some frustration at being unable to continue. Many of the positive benefits perceived by the patients were attributed to *their* exercise via stimulation rather than to the use of a stimulator box and a set of electrodes.

DISCUSSION

Electrical stimulation of paralyzed muscle has the potential to cause significant changes in the way that an individual with spinal cord injury perceives himself and his world, for better or for worse. It is incumbent upon clinicians and researchers using electrical stimulation to be aware of these considerations. We feel that the measures that we have used, given a larger sample size, should yield important insights into these perceptions.

Our preliminary findings point out some very important factors that need to be considered when electrical stimulation of paralyzed muscle is proposed. First and foremost, early reconditioning may be essential. And, secondly, muscles may weaken past a point where it is no longer viable for them to be reconditioned.

The present study is evaluating a task of major importance to the patient yet does not promise unrealistic hopes, has provided data for evaluating basic physiological processes and adaptation in the spinal cord patient, and has offered the potential for designing more efficient and effective therapeutic programs for patients with spinal injuries.

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THE ULTIMATE WHEELCHAIR-A MOBILE PLATFORM THAT PERMITS THE PERSON TO FULLY ADAPT TO HIS ENVIRONMENT

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ABSTRACT

The standard wheelchair in use today lacks the ability to function as more than just as a chair with wheels. For any disabled person who is bound to a wheelchair, they have a hard time interacting with their environment. It is a struggle to adapt the environment to fit them and their wheelchair.

The design presented in this paper reapproaches the concept of what a wheelchair should do. This concept provides a mobile platform design that serves as a base on which accessories and options can be added. These accerssories help adapt the person to the environment instead of adapting the environment to the person.

BACKGROUND

Many "new" designs have been proposed or developed for the wheelchair. Most have approached the redesign of the common wheelchair by adding lightweight materails or special wheels. Little thought has been given to the concept of how a wheelchair should function. For the sever ely disabled person, the wheelchair is their environment and home. They don't just use it to move around, the severely disabled person eats, works and plays out of this chair. Even the not-so-severely disabled are still restricted in their movements. The standard wheelchair lacks the ability to function properly in an "able body" environment.

Thus, the need to re-think the concept of a wheelchair is apparent. Disabled people are becoming more important and functional in our environment. They need to have the ability to participate effectively in our society. Since they are bound to this chair, the chair needs to become more integrated to the environment.The concept needs to be of adapting a wheelchair to the environment and not adapting the environment to a wheelchair. No longer should a wheelchair be thought of as just a chair on wheels.

DESCRIPTION

The basic mobile platform would have standard sides while the rest of the components such as cross bracing and the seat vary in size to meet the different physiques of the occupants. Using the standard sides, accessories could be designed to meet the disabled person's needs and wants. Thus, the new mobile platform has customized the main structure as well as the options.

One version of the concept for a mobile platform is shown in diagram 1. The basic platform consists of two major parts. Two identical side panels, each being composed of two mirror image pieces, are the fondation of the concept. They are the structure on which the cross bracing, seating and wheels are secured. With the basic structure established, the side panels have molded in slots and recess areas that serve as snap-in locations for accessories. (See diagram 3) These snap-in locations have connection plugs for power and control of accessories. A wiring harness is sandwiched between the two mirror image pieces of each side panel that runs between the side panels as well. With a wiring bus port provided, the mobile platform can easily be tailored to meet many different needs. A person could have several accessories,a desk, storage or a light, and simply snap-in the one they need for the day. Other examples will be covered in the results section.

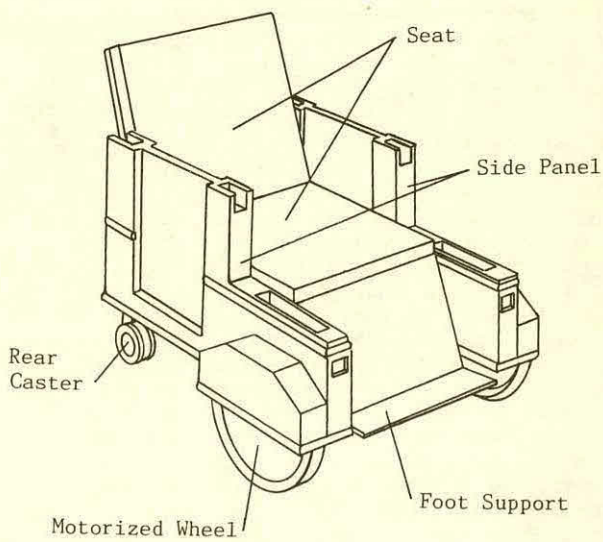

Diagram 1 - Mobile Platform

There are several other important aspects of the side panels. Both side panels have lights located in the front for night travel. There are two rubber strips around the ends that run the length of each side panel. They serve as bumpers to protect the mobile platform and allow for easy access through doors. A typical problem for standard wheelchairs is access through doors. With the mobile platform, the door

will be less of a problem since there are less objects for it to get hung on. Another advantage of the rubber strips is that they cover up the bolts and screws that hold the mobile platform together. This gives the mobile platform a cleaner look.

In this version of the mobile platform, there are two cross braces that hold the two side panels apart. This allows for the seating, one unit consisting of the seat back, bottom, leg rest and foot rest, to slide forward on tracks. The occupant then has the ability to exit the mobile platfom to the front or to either side by way of a built-in transfer board located in the seat bottom. The side exits are easily accessable since the front drive wheels are smaller in diameter compared to the standard wheelchair. They are 12 to 14 inches in diameter with the motor being the hub of the wheel. Locating the motors in the wheels saves considerable amount of space. Caster style wheels are located in the rear of the chair under each panel.

Power for the wheel motor, lights and any accessories is provided by nickel cadium battery packs. These are located in the side panels directly below the seat. The advantage to using this type of battery is that it takes up less space and weighs considerably less. If any additional power is required, extra batteries can be added under the seat. The battery pack will have a recharging cord built in. Thus, the battery pack can be plugged in for recharging at the office or at home during the night.

MATERIALS AND METHODS

The side panels and most of the other major parts of the mobile platform are made from plastic using the double sided vacuum forming process (See diagram 2). This process is much like standard vacuum forming where the plastic is heated, pulled over a mold and vacuum applied to pull the plastic into the recesses and corners of the mold. The only difference is this process is done to two pieces simultaneously and then the molds are pressed together under tremendous pressure. This fuses the two pieces together into one strong hollow piece. Ribs, foam, metal or wood reinforcements can be added to give the piece extra strength and stability. The single unit can have textured surfaces and be water-tight. Also, the areas that serve as snap-in units can have locking tabs molded into the pieces. This eliminates the need for any additional hardware.

Besides all physical advantages that double sided vacuum forming has, an added advantage is the low cost of molds and set-up. Depending on the number, finish and quality of the pieces, the cost is relatively cheap compared to other manufacturing processes for plastic.

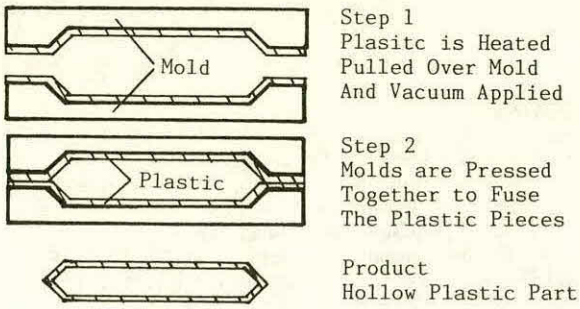

Diagram 2 - Double Sided Vacuum Forming

FUNCTION

The basic mobile platform provides a base on which accessory packages can be added. This can be considered the great advantage the mobile platform has over other wheelchairs. In this section, some of the many variations on the mobile platform will be examined.

The side panel can be considered the only standard part of the mobile platform. When a person receives their mobile platform, it will come with two side panels and a varied assortment of items such as wheels, cross bracing, seating and accessories. These items fall into two catagories. One catergory is the snap-in accessories. These are the units that a disabled person can easily "snap-in" while still being seated in the mobile platform. They would not be able to do this by themselves. The second catergory is the accessories that deal with the set-up of the mobile platform. They are the items like cross bracing, seating and wheels that are installed in the initial assembly of the mobile platform.

The first category of accessories is the snap-in variety. The different areas that the accessories snap into have standard dimensions(see diagram 3). This allows for one snap in area to have several accessories that can be changed out. In addition, an accessory could be made to fit into sereval different areas as well. An example would be a storage system that has several variations. A disabled person would go through a catalog an choose which storage system preferred: the unit that fit into the recess area or the forward slot area of the side panel.

A possibile accessory includes a rain protection system. This accessory snaps onto the back of the wheelchair and can be pulled down over the occupant. Other accessory packages include two types of desk units. One unit is storable in a pocket located in the side panel's recess

area. It is simply pulled out and layed across the side panels. The second unit would have a large work area that is supported by an articulating arm. The hole unit snaps into the side panel's forward slots.

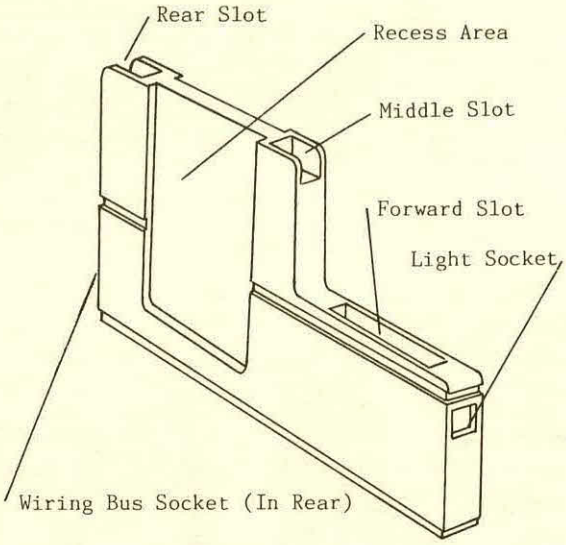

Diagram 3 - Side Panel's Accessory Locations

Two other accessories include a reading lamp and a mechanical arm. The lamp simply plugs into the side panel's rear slot area. The mechanical arm is snaped into the side panel's forward slot area and is used for picking up briefcases or other similar items.

The second catagory of accessories deals with the physique of a person and how the mobile platform functions. The cross bracing would be the first item to consider. Either a fixed or folding cross brace could be attached to the side panels. The folding cross brace would allow the mobile platform to collapse to only a foot in width. This would allow for it to be stored behind a car seat or in a trunk. The fixed cross brace version of a mobile platform would be for more of the home or hospital use.

Once the cross bracing is established, the seating for the mobile platform could be picked. With the folding version of the mobile platform, flexible or flip up seating would be used. Also, the different parts of the seat such as the back, bottom, leg and foot rest, can be made as one unit or as individual items. Each part would have its own adjustments. Thus, a seat back could easily fold down to form a flat surface for a person to lay on.

Another possibility for seating is to have extra supports that simply snap-in. These supports would give a person who has a hard time holding up his body the support where he needed it. With power in the mobile platform, bean filled cushions could be shaped to fit a person's body and then have the air pumped out by built in vacuum pumps. This would allow for the cushions to hold their shape yet be chageable at a later date.

Another area that deals with the function of the mobile platform is the wheels. They are designed to simply bolt to the side panels as one unit with the fenders. With a little help a disabled person could attach or replace the standard wheels. Possible options include a wider wheel for use in the garden or on the beach. It is even possible to have skis bolted on for going down the slope!

The ability to have manual drive is eaily accomplished. Panels with rachet drive mechanisms are snapped into the recess areas of the side panels. A snap out panel in the fender is taken out allowing for a drive wheel to rub against the front wheel. The occupant simply moves the rachet arm back and forth to provide movement.

DISCUSSION

The parts for the mobile platfrom can be made inexpensively compared to parts for a standard wheelchair. Assembly of the mobile platform is straightforward and simple. Assembly of the mobile platform would not be labor and machinery intensive. Another possibility is to offer the mobile platform in a kit form. A person could go into a health care store, pick up the appropriate pieces and take the unit home for assembly. This would only require a couple of wrenches and a screw driver. If a piece needs replacing, the chair can be taken back to the store for a new piece, eliminating the need to send it back to the factory.

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AN ANTI-SHIMMY DAMPER FOR WHEELCHAIR CASTERS

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ABSTRACT

Caster shimmy is a common and sometimes dangerous characteristic of wheelchair casters. A simple device has been developed to provide mechanical friction about the stem of the caster, thus preventing shimmy for normal wheelchair speeds.

INTRODUCTION

Caster flutter or shimmy is a common problem with wheelchairs and other vehicles using casters such as shopping carts, hand trucks and the like. Considerable work has been done in overcoming shimmy in the tail and nose wheels of aircraft. Kauzlarich[1] has developed a comprehensive theory of flutter that is applicable to wheelchairs and introduced the grooved tire as one means for overcoming the problem. Mochel[2] has studied hydraulic and mechanical friction dampers in reducing flutter and determined that a small amount of mechanical friction can effectively eliminate flutter for typical wheelchair speeds. Wave washers and O rings have been used to pre-load the caster stem bearings to provide the necessary friction. These solutions require frequent adjustment and are less than satisfactory for general use. A device has been designed specifically to provide the correct amount of friction, independent of the bearings and with the expectation of lasting the life of the wheelchair without adjustment.

Description (See Figure 1)

The device consists of a set of conical wedges that fit between the caster stem and the housing. The inner wedges fit around the stem and mate with the outer wedges that fit the inside of the housing. The mating surface between the wedges forms an angle of about ten degrees. A compression spring co-axial with the stem forces the inner wedges into the outer wedges, which in turn forces the outer wedges against the housing and the inner wedges against the stem. The resulting friction against the stem produces a moment of about 7 in lbs. The moment induced between the outer wedges and the housing is considerably larger, because of the larger radius thus preventing rotation between the outer wedges and the housing.

The prototype wedges were made from black acetal rod. A steel spring with a constant of about 16 lbs per inch was compressed on assembly to provide about 10 lbs of force. The wedges were machined but not polished.

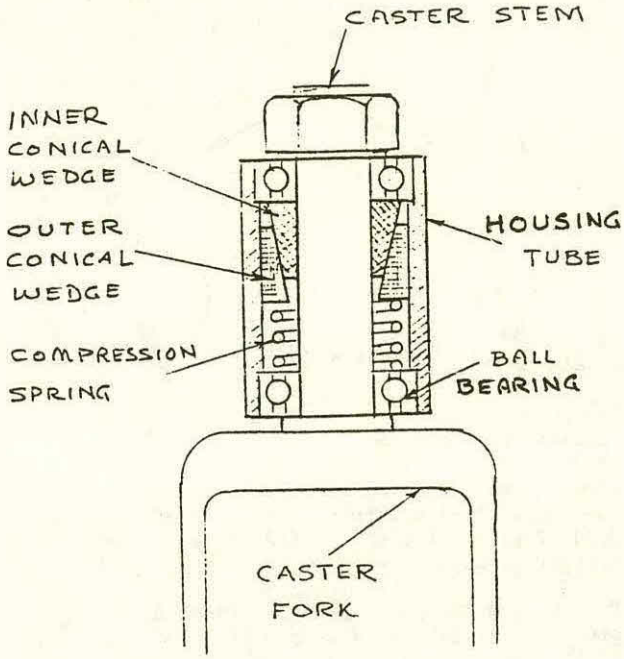

FIGURE 1

Testing

The design of the device is such that any slack in system due to wear would be taken up automatically by the compression spring. To test this, the device was mounted to a caster assembly which was in turn clamped to a bench. A gear motor, with a crank and connecting rod was linked to the caster wheel in a manner that would swivel the caster forth and back through an angle of about 45° at a rate of 68 cycles per minute. The rate was chosen to simulate turning in normal use. At the beginning of the test the torque was 7 in lbs. After about 200,000 cycles this value fell to about 4 in lbs and remained at the value through a total of 1,000,000 cycles. The system was then disassembled and the parts examined visually. There was no apparent wear on any of the parts. The drop in friction after 200,000 cycles was attributed to polishing of the caster stem.

The device was again mounted for testing in the same manner, except that several squirts of motor oil were added to simulate what would happen if someone lubricated the bearings with an oil can. Since sealed bearings are used, this oiling is not recommended but could happen. The results showed an actual increase in the friction around the caster stem. This is thought to be due to the wedges being driven further into engagement.

A second test was designed to determine the effectiveness of the device in preventing caster flutter. The device was mounted in a wheelchair on one front caster. The other front caster possessed inherent frictions, but was otherwise unrestricted. The wheelchair was placed on a moving belt, tethered with a nylon tow rope and loaded with 100 lbs of marble chips. The belt was accelerated to its maximum speed ... 11 1/2 kph (7.2 mph) and the caster wheel struck with a rubber mallet to initiate flutter. The blow caused the other caster wheel to go into unstable oscillation (flutter) but had no effect on the caster equipped with the damping device.

CONCLUSIONS

The conical wedges forced together with a compression spring provide an inexpensive and reliable way for providing friction about the caster stem. The friction so provided is adequate in controlling caster flutter for typical wheelchair speeds. In production the conical wedges could be produced by injection moulding. The amount of friction produced is a function of the spring force, which could be chosen for different applications. A torque of about 7 in lbs appears to be adequate for general use and the additional effort required for turning the wheelchair is not apparent to the user.

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ANALYSIS OF MANUAL WHEELCHAIR BRAKES

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ABSTRACT

The typical manual wheelchair manufactured in the United States is equipped with wheel locks that are effective only as parking brakes. Dynamic braking of a wheelchair is provided by manually applying a retarding force to the wheel or handrims. A mechanical dynamic brake would allow a wheelchair user to control the wheelchair's speed, direction, and to bring the wheelchair to a complete stop safely and easily by absorbing the wheelchair's energy of motion. Dynamic brakes could enhance control, safety, and convenience for many wheelchair users.

NATIONAL SURVEY

In order to determine the feasibility and practicality of dynamic brakes for wheelchairs from the user's perspective, a survey by the University of Virginia Rehabilitation Engineering Center was published in the March 1985 issue of PARAPLEGIA NEWS, a publication of the Paralyzed Veteran's of America. The publication reaches a large number of disabled individuals and, therefore, presented an effective means of distribution of the survey. The primary goal of the survey was to obtain wheelchair users' impressions of existing braking systems and their interest in dynamic brakes. The secondary goal of the survey was to obtain a profile of the wheelchair user for comparison purposes. The brake survey consisted of an introductory section and two questionnaire sections. The introductory section discussed dynamic brakes and stated the purpose of the survey. The first questionnaire section consisted of seven questions concerning present braking systems and dynamic brakes for wheelchairs. The second questionnaire section consisted of fourteen questions relating to the wheelchair user.

One hundred sixty eight (168) disabled individuals responded to the survey. Seventy-two percent (72%) of the respondents felt that the wheel locks currently available on manual wheelchairs are inadequate. Approximately 81% of the respondents felt that there was a need for dynamic brakes. The percentage of respondents who felt a need for dynamic brakes was significantly higher than those who felt that there was no need for dynamic brakes (12.1%). Seventeen percent (17%) of the respondents were not interested in obtaining dynamic brakes. The majority of the respondents (51.5%) stated that they would be willing to spend between $25 and $50 in order to obtain dynamic brakes. Twenty-three percent (23%) were willing to spend between $50 and $100 while 8.5% were willing to spend over $100.

Questions concerning speed limiting brakes were also included in the survey. Speed limiting brakes were described in the introductory section as: "a mechanism that would automatically be applied when the wheelchair reached a certain speed. The purpose of this design would be to maintain a safe speed for a wheelchair when travelling on hills and in other dangerous situations."

A majority of the respondents felt that speed limiting brakes would be helpful (64.8%) while 19.1% did not feel that they would be helpful. The remaining respondents had no opinion. Over 34% of the survey respondents stated that they were not interested in obtaining speed limiting brakes, 41.9% were willing to spend between $25 and $50 and 18.8% were willing to spend between $50 and $100. The remaining 5% were willing to spend over $100.

Chi-square tests were performed to determine whether the response to a specific survey question was independent of the type of wheelchair or user. The results of the chi-square tests indicated that there was no evidence that the response to the questions concerning dynamic brakes or speed limiting brakes was dependent on the type of wheelchair or user. The majority of the survey respondents felt a need for dynamic brakes regardless of sex, age, level of disability, type of wheelchair, geographic area of residence, purchaser of equipment, or whether they are independent in daily activities.

EXPERIMENTAL TESTS

Commercially available bicycle caliper and hub brakes were acquired to test for suitability as wheelchair brakes. Bicycle brakes were used in this project because they were inexpensive, lightweight, and manually actuated. Dia-Compe #810 rear side-pull caliper brakes weigh 0.33 lb_f and cost about $10.00 including the cable with casing. These brakes are readily available through any reputable bicycle repair shop. The Sturmey-Archer Elite VT P3473 model hub brakes were modified for testing. The axle of this brake was redesigned to accept

attachment only to one side. These brakes weigh 1.6 lb_f, cost $45.00 each and are only available through Bauer Cycle Supply, Inc., Minneapolis, MN.

Two types of tests were performed on each brake to determine their performance characteristics. The first were bench tests using a Prony brake to measure brake capacity, while monitoring brake temperature. The second test evaluated the brakes as they would be mounted on a wheelchair using a wheelchair treadmill.

The initial tests were conducted to determine the fade characteristics of the brakes. Brake fade occurs when the temperature of the brake lining material reaches a high temperature resulting in a reduction of the coefficient of friction. Consequently, the braking torque decreases. The brakes were bench mounted on a small engine lathe in a Prony brake configuration. A moment arm and an oil damped spring scale were used to measure the braking torque. Brake pad temperatures were monitored with embedded thermocouples at various test speeds and actuating forces during the brake fade tests.

Identical models of the hub brakes were modified for use on an Everest and Jennings Premier wheelchair. The modification consisted of altering the axle shaft so that the brake could be mounted in a cantilevered configuration on the side frame of a wheelchair instead of between the forks of a bicycle. An actuating mechanism was also designed to be used with both types of brakes. The modified brakes were tested to determine whether the brakes would be adequate during actual use by conducting experiments using a treadmill designed to accommodate wheelchairs. Similar tests were conducted with the caliper brakes mounted on the same wheelchair.

RESULTS

The caliper and drum brakes were bench tested at speeds of 48, 84, 124 and 156 rpm. Typical wheelchair wheels have a diameter of 0.62 meters resulting in wheelchair speeds of 5.5, 9.1 14.3 and 17.9 km/hr, respectively. The actuating force varied from 17.64 N to 49.49 N for the drum brake and from 58.8 N to 88.7 N for the caliper brake. The time duration of each test was 15 minutes unless either the brake faded or the temperature of the brake lining approached the manufacturer's recommended safe temperature which was 450 F.

Figure 1 illustrates the braking action of the caliper brakes. The actuating force, 88.7 N, was kept constant while the running speed varied from 48 to 156 rpm. There was little variation in the braking torque between each test. The highest braking torque, 15.2 N-m, occurred at the highest speed while the lowest braking torque, 14.2 N-m, occurred at the lowest speed.

Figure 2 shows the variation of the hub brake torque versus time for a low actuating force (17.64 N). The braking torque was not constant throughout the test, but increased with time. One possible cause for the increase in braking torque during each hub brake test was the increasing temperature of the friction lining material as the test progressed. Once the temperature of the friction lining had increased to a temperature where the lining is more pliable, the lining comes into better contact with the brake drum, thereby, increasing the brake friction.

Brake fade occurred in 8 out of 13 tests conducted on the hub brake. The results of the tests indicated that brake fade is dependent on the temperature of the brake lining. The average temperature at the time of brake fade was 285 F $\pm$21.76 F.

The braking capacity was calculated using the braking torque measured during each brake fade test. Figure 3 shows the relationship between the experimentally determined braking capacities of two Dia-Compe caliper brakes and the minimum and maximum braking capacities determined for a 75 kg load on a 25 kg wheelchair on a 1:12 slope (4.76 degrees). The caliper brakes could supply sufficient braking capacity to maintain the rider's speed on a 1:12 slope. However, the rear wheels could not be locked using the actuating forces used during the brake fade tests. A comparison between the braking capacity of two Sturmey Archer hub brakes and the maximum and minimum braking capacities required for a 75 kg rider on a 1:12 slope can be seen in Figure 4. The only tests that yielded braking capacities large enough to maintain a rider's speed on a 1:12 slope were those conducted at the highest actuating force of 49.5 N.

Both types of brakes had the capacity to lock the rear wheels during the treadmill testing. The braking force, measured by a load cell during the treadmill testing, was not constant throughout the five minute tests. Four tests were run on each type of brake and only in one set of tests (high actuating force, high treadmill speed) did the fluctuations in the hub braking force exceed those measured for the caliper brakes. The caliper brakes tended to "grab", whereas the hub brakes provide a smoother braking action.

The results of the various tests indicated that the Sturmey-Archer hub brakes have more advantages than disadvantages when compared to the Dia-Compe caliper brakes. The force required

by a wheelchair user to actuate the brake is less for the drum, or hub brake than for the caliper brake. This is beneficial because many wheelchair users have limited strength and range of motion. The hub brake does not require frequent adjustment or a special mount for the wheelchair, unlike the caliper brake. In addition, the hub brake is not as easily affected by wet weather, does not have a tendency to lock-up during use, and provides quiet braking action. Unlike the caliper brake, the hub brake could fade during periods of extended use. However, it is unlikely a wheelchair user would ever experience brake fade. In a typical test, brake fade occurred only after 4.5 minutes at a speed equal to 124 rpm with an actuating force of 49.49 N. To reach this point of initial brake fade, the wheelchair user would have to travel 1.07 km, at 14.25 km/h with the brakes fully applied.

RECOMMENDATIONS

General recommendations for a braking system for a manual wheelchair include:

1. The brakes should be lightweight and durable.
2. An ideal braking system would allow the user to use quick-releasing hubs, if so desired.
3. The brakes should not require frequent adjustment
4. The braking system should not hinder the wheelchair's ability to fold.

The interface between the brake and the user should be investigated further. The actuator should be both lightweight and easy to use. In addition, the position of the actuator should allow for easy transfers in and out of the wheelchair. Future investigations should look into the pros and cons of providing differential braking through a single-sided actuator. For future evaluation of the brakes considered in this project, it is recommended that additional testing of the braking system on the treadmill and evaluation of the brakes by handicapped users be used to obtain their opinion of the various types of dynamic brakes.

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ACKNOWLEDGEMENT

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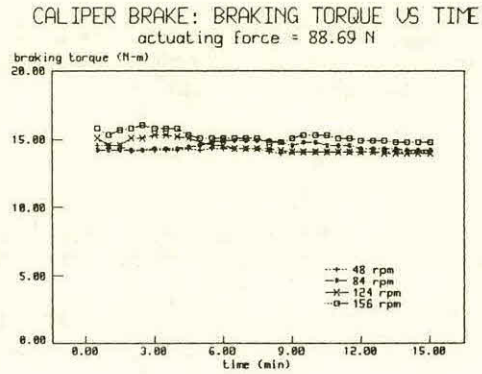

Figure 1

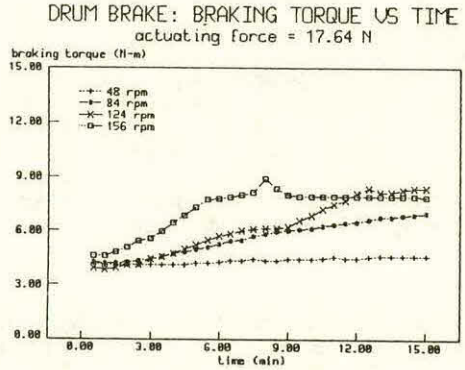

Figure 2

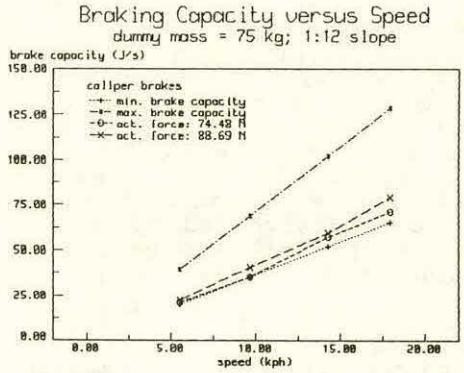

Figure 3

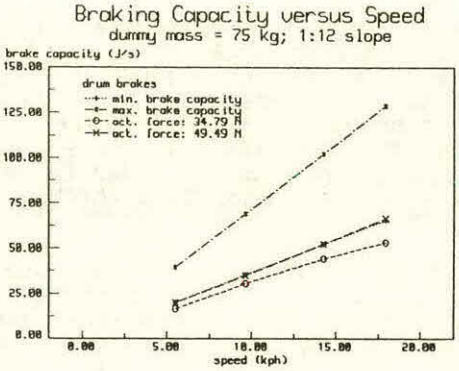

Figure 4

FATIGUE ANALYSIS OF WHEELCHAIR FRAMES

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ABSTRACT

From the results of previous work, low-cycle fatigue has been proposed to be a major factor in wheelchair frame failure. Experimentation at the Veterans Administration Prosthetics Center showed wheelchair frames failing before 8000 stress cycles which resulted from the chair rolling over uneven surfaces. In studies at the University of Virginia REC, strain levels were measured in a wheelchair frame as it traveled over several different configurations of rods and slats. This experimental work provided evidence that low-cycle fatigue is a significant factor in wheelchair frame failure. Some suggestions for design changes in a wheelchair frame are included in the paper.

INTRODUCTION

The purpose of this research was to determine if fatigue is a significant mechanism in the failure of wheelchair frames. Experimental studies of wheelchair frames at the University of Virginia-REC (UVA-REC) and Veterans Administration Prosthetics Center (VAPC) have lead to this work.

At the UVA-REC, wheelchair frames have been analyzed structurally in both static and dynamic modes (1,2). In the static analysis, experimental and computational methods were used to determine the magnitudes and locations of the four largest stresses in an occupied, stationary wheelchair (1). Results from the study showed that maximum stresses occur in the cross-tubing. This is a reinforced area on the wheelchair frame where failure is unlikely. These results lead to the conclusion that where failure is unlikely. These results lead to the conclusion that wheelchair frames do not fail due to static loading. Therefore, some type of dynamic mechanism must lead to wheelchair frame failure.

In the dynamic study, several types of caster forks were analyzed on a wheelchair as it traversed a bump (2). Although major attention was given to the caster forks, stresses were measured in the tubing behind one of the caster forks on the chair. Wheelchair frame failure occurs at this location (3,4). When the caster contacted the bump, stresses measured at this location were approximately four times greater than those from the static analysis. These results led to the consideration of the forces caused by rolling over small obstacles as a cause of wheelchair frame failure.

In addition to the studies at UVA-REC, tests were undertaken at the VAPC to determine the effect of rolling over small obstacles on a wheelchair frame (3,4). A wheelchair attached to a carousel was towed around a circular track over a series of obstacles. The configuration is shown in Figure 1. Measurements were made of the dimensional changes in the chair after a specific number of revolutions over the obstacles. The scheduled test series could not be completed due to wheelchair frame failure. The frame failed after less than 8000 stress cycles resulting from traversing the ramps and slats. One of the failure locations was in the horizontal tube just behind the front caster fork.

Both of the wheelchairs tested at the VAPC failed before 8000 collisions with obstacles. High stresses measured during the UVA-REC caster evaluation tests occurred when the wheelchair rolled over an obstacle. From these two results, it was proposed that wheelchair frames fail due to low-cycle fatigue. Low-cycle fatigue is a mechanism that leads to metal failure after a few thousand cycles of plastic strain. This cyclic straining occurs at stress concentrations in a structure as a result of large loads (5). There is a stress concentration on the frame behind the front caster fork due to a small drain hole in the tubing. Therefore, this area is susceptible to low-cycle fatigue.

The empirically determined Manson-Coffin equation, equation (1), is used to describe low-cycle fatigue.

$$\frac{\Delta\varepsilon_T}{2} = \frac{\sigma_f'}{E}(2N_f)^b + \varepsilon_f'(2N_f)^c \qquad (1)$$

where $\Delta\varepsilon_T/2$ is total strain amplitude,
$2N_f$ is the number of reversals to failure and
b, c, E, σ_f' and ε_f' are material properties.

Values were determined for the low-cycle fatigue parameters, b, c, σ_f' and ε_f', for stainless and low carbon steels.

The Manson-Coffin equation was then used to compare theoretical and experimental strain amplitude.

METHOD

Experimental methods were used to determine the significance of low-cycle fatigue as a mechanism of wheelchair frame failure. Two types of tests were performed. In the first test, samples of wheelchair tubing instrumented with strain gages were placed in four-point bending to determine the low-cycle fatigue parameters of wheelchair materials. In the second test, strains were measured as a wheelchair traversed several sets of obstacles on a treadmill.

RESULTS

The experimental results were used to determine if low-cycle fatigue was indeed taking place. Strains measured as the wheelchair frame rolled over obstacles were placed on the strain-fatigue life curve for the corresponding material. From the values of the strains, an estimated wheelchair frame life could be determined.

Experimentally determined low-cycle fatigue parameters were compared with those found in the literature for similar materials. Using the resulting parameters in the Manson-Coffin equation for low-cycle fatigue, the strain-fatigue life curves for UNS G10050-10090 steel and UNS S30400 stainless steel were plotted as shown in Figures 2 and 3. UNS G10050-10090 steel was the material in the Everest and Jennings wheelchairs used in the VAPC tests. UNS S30400 stainless steel was the material in the Stainless wheelchair used in the UVA-REC tests. Test results are shown on the corresponding curves in Figures 2 and 3.

Most of the strains in the wheelchairs made from UNS G10050-10090 steel were in the low-cycle fatigue region as shown in Figure 2. The largest dimensional changes occurred in a measurement that included the horizontal tube just behind the front caster fork. Thus, it appears that the wheelchairs tested at the VAPC failed due to low-cycle fatigue. However, the strains in the wheelchair made from UNS S30400 stainless steel were in the high-cycle fatigue region as shown in Figure 3. Therefore, it appears that the wheelchair tested at UVA-REC did not enter the low-cycle region.

While a contradiction appears to exist between VAPC and UVA-REC test results, closer analysis reveals an important difference between the two types of tests. In the VAPC test, large sections of the wheelchair frame were measured to determine strain. These sections included areas of stress concentration where the plastic strain of low-cycle fatigue would occur. In the UVA-REC tests, strains were measured at specific points on the frame that were located away from areas of stress concentration. Therefore, although plastic strain may have occurred at areas of stress concentration, it could not have been measured with strain gage technology (5). It is reasonable to conclude that the probable cause of failure during the VAPC tests was low-cycle fatigue.

CONCLUSION

There are two conclusions that can be drawn from this work. First of all, there is more evidence that low-cycle fatigue plays a significant role in the failure of wheelchair frames at locations of stress concentration. Secondly, since the horizontal tube just behind the front caster fork is highly susceptible to failure, redesign of this area may extend wheelchair frame life.

Although the results may initially appear contradictory, the wheelchairs tested at the VAPC appear to have failed due to low-cycle fatigue. The plastic strain associated with low-cycle fatigue could not be measured with the strain gage instrumentation used during the UVA-REC tests. Plastic strain may have occurred during the tests at UVA-REC and remained undetected.

Results of the VAPC tests suggest that the horizontal tube just behind the front caster fork is a critical location on wheelchair frames. This location can be strengthened by design changes in the wheelchair frame. Two possible changes to the frame are dimensional changes and removal of the curved section of tubing as in the British wheelchair design. With the current wheelchair design, the life of the frame could be increased significantly by changing the horizontal tube just behind the front caster fork from the 2.56 cm O.D. X .13 cm wall typically used to a 3.8 cm O.D. X .4 cm wall. This change would reduce the magnitude of the strain experienced by the tubing at this location resulting in a more acceptable high-cycle fatigue mode.

Thus, the number of cycles to failure would be expected to increase by an approximate factor of 100. Another possible design change would be the removal of the curved section of tubing. It would be replaced with a straight section as shown in Figure 4. This design is currently most prevalent in wheelchairs in Great Britain.

Results from this work provide evidence that low-cycle fatigue is a significant factor in the failure of wheelchair frames at stress concentration locations. However, more work needs to be done in this area before final conclusions can be drawn.

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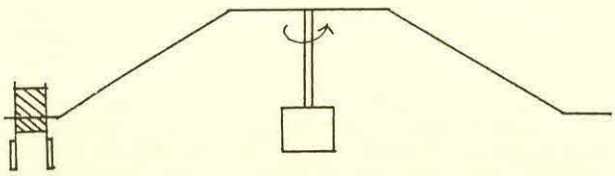

Figure 1. Sketch of VAPC Test Carousel

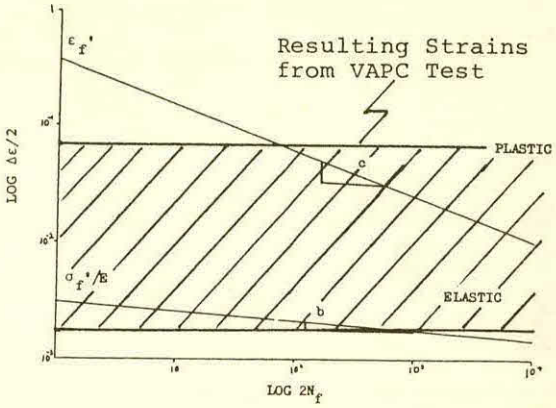

Figure 2. Strain-Fatigue Life Curve for UNS G10100 Steel

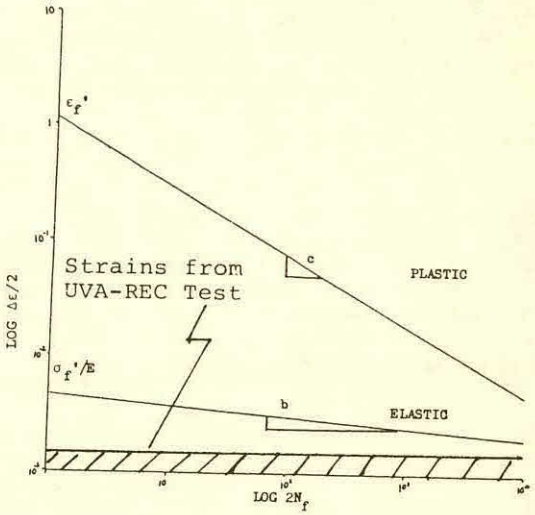

Figure 3. Strain-Fatigue Life Curve for UNS S30400 Stainless Steel

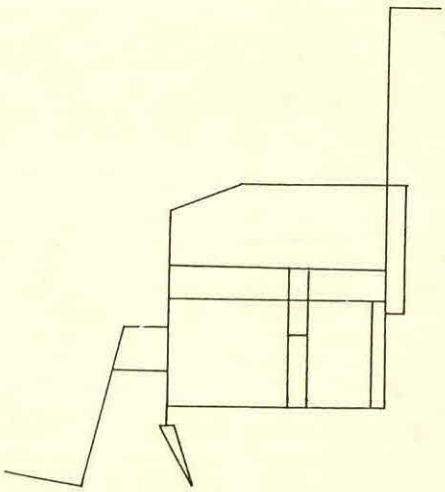

Figure 4. Wheelchair Frame with Curved Section Removed

THE POSITIONER DESK: A DYNAMIC STANDING DEVICE FOR YOUNG CHILDREN

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ABSTRACT

The evolution of the design for the positioner desk, an alternative positioning device provided for 25 young physically disabled children, is described with a discussion of clinical results.

INTRODUCTION

Adaptive equipment for positioning and mobility is playing an increasingly larger role in the lives of young disabled children as both early intervention and rehabilitation engineering programs abound, the availability of commercial equipment increases, and the awareness of therapists and parents grows. The introduction of equipment at early ages is regarded to have significantly positive effects in preventing deformity and enhancing motor development (1), as well as in facilitating cognitive and social growth (2). And it is becoming more widely recognized that one piece of equipment cannot meet all the needs that a child may have in the course of his/her day, neither functionally nor therapeutically (3). A primary piece of equipment may provide the type of postural support needed for a child to sit for longer periods of time during activities, but alternative devices are also needed to allow changes in position and types of movement, e.g. for sidelying, prone, standing, or kneeling. Two years ago we reported on the positioner desk, a device designed to assist children in developing gross motor skills while providing the opportunity for controlled lower extremity weightbearing (4). As of this writing we've built 25 desks and, while the therapeutic principles and the basic design have remained the same, our choice of materials, the design of specific components, and our understanding of appropriate clinical use have changed.

BACKGROUND

The first positioner desk was built four years ago for a child who was unable to stand without support but whose hip and knee flexion contractures precluded the use of a prone stander. His cognitive and social skills were so low that his teacher had concerns about placing him in equipment which would further isolate him from the other children. His muscle tone and abnormal postural reflexes resulted in a pelvic obliquity such that, at the age of 3½ years, one hip was subluxed and the other appeared at risk. His orthopedist and his physical therapist were both recommending lower extremity weightbearing to try to prevent the development of fixed skeletal deformities, and the therapist had been experimenting with the Norwegian Balans chair (5) as the support surface for the child. The Balans chair, which recently has become very popular and widely copied in this country, is based on the work of Keegan (6) and Mandel (7) and has been aimed primarily at providing better seated posture for the growing segment of the population who work all day sitting down. The basic concept is to allow the individual to sit with a more open hip angle than

Figure 1. The positioner desk in use by a 4 year old child playing with his twin brother.

the typical chair with its 90° seat to back angle permits, thus relieving tension on the hamstring muscles and discouraging the lumbar kyphosis produced by a posteriorly tilted pelvis. In the Balans chair this is achieved with a forward tilting seat and knee supports which keep one from sliding forward. We incorporated this general principle into a box-framed device which contained a tractor seat, knee block, chest support, and foot platform and which was attached to the end of a sand table in the classroom so the child could stand and play with the other children. We left enough space between components for him to move within a prescribed range, coming to stand and sitting back down on the seat independently. In addition to accomplishing the therapeutic objectives initially identified, this first prototype allowed us to explore the concept of a dynamic device which would allow a child to practice the gross motor skills being worked on in therapy independent of the guidance and assistance of the therapist's or parent's hands. As the design of the desk evolved and we gained experience with more children, we began to understand how the various components could relate to each other to inhibit abnormal patterns of movement and facilitate more normal movement. This resulted in continual changes in the design and selection of the individual components, the hardware, and the frames themselves with the goals of increasing their effectiveness and ease of adjustment and decreasing production time.

DESCRIPTION OF THE PROJECT

Population

The 25 positioner desks have been built for children ranging in age from 2 years 0 months to 11 years 10 months. The mean age at initiation of use was 4 years 4 months, the median age 3 years 6 months. Twenty-four of the children had diagnoses of cerebral palsy, 19 with secondary diagnoses of cognitive

delays or mental retardation, and one child was diagnosed with Down's syndrome. These children were referred by their primary therapists who worked with them at home and/or in schools across western Massachusetts.

Frame Design

The first prototype mounted on the sand table proved to be quite bulky and difficult to adjust, and it lacked the versatility necessary for it to function effectively within an ever-changing program of developmental therapy. All of the desks since then, therefore, have been built as freestanding units, on a scale which fits well into a living room or a preschool classroom. The frame is essentially a three-sided box with two casters on the front, intended to be used as a stationary device but moveable within an indoor setting. The first 14 desks were made from half-inch particle board but this was changed to quarter-inch phenolic board to save time in construction. The phenolic requires no additional finishing after slots and edges have been rounded over and sanded, in contrast to the three coats of finish required to properly seal the edges of the particle board. The phenolic board's durable laminate finish is also easy to clean, and the appearance of the lighter "butcher block" surface has proven more popular with the families of the children who use the desks. The frame we are using for initial fittings has one side made from clear three-eighths inch polycarbonate, allowing us to observe the child's position, take measurements, and photograph the process for use during assembly of the final product.

The configuration of the frame has changed very little since the second desk. We have tried building larger frames for four older children but found in three of the four cases that they were inappropriate, both clinically and socially. The desks looked too big and out of place in the environments of these children, they were difficult to transfer heavier children in and out of, and there were serious concerns about the safety and therapeutic soundness of having children with higher levels of spasticity use this type of device.

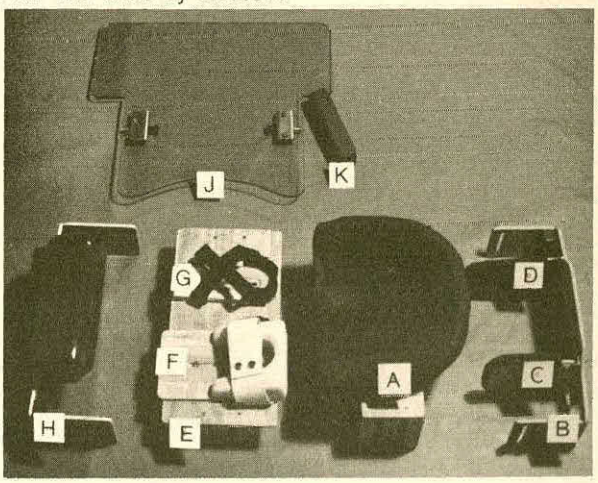
Another change which occurred fairly early in the development of the desk was in the manner of attaching components to the frame and allowing adjustment. This was originally done with a series of holes in the sides of the frames, with bolts through the frame into the components. We soon found that using slots in the frame permitted easier adjustment while the child was in the positioner, allowing us to assess immediate responses of muscle tone and balance to subtle changes of angle or position of one or more components. By standardizing the system of slots we are now able to fabricate the desk frames before the initial fitting, as well as to recycle them when they are returned to us after the children have outgrown their use. This is in keeping with the overall design plan to standardize as many components as possible, looking towards production on a wider scale by a commercial manufacturer.

Component Design

The individual components determine the effectiveness of the desk for each child. In the production of the first group of positioners, new components were continually added to the design package to customize the unit for each child we tried to fit. As we worked with more children, however, we found that certain components were consistently appropriate, some were needed only in certain cases, and others were so difficult to work with that they actually discouraged use of the desk by parents and staff. Our most recent efforts have concentrated on standardizing the components, reducing the number of different parts and making them interchangeable and more easily adjustable. Using slotted phenolic endplates on all the cross members, for example, has made a finer range of adjustment possible and allows us to use the same hardware to mount each component in the frame.

The key component in all of the desks has been the seat because of the crucial role of the seat angle in positioning the pelvis and directly influencing muscle tone throughout the body. Twenty-three of the desks were made with tractor seats of various sizes [A in Fig. 2], fabricated from sculpted foam on a plywood panel and mounted on a crosspiece in the frame. The contour of the seat cradles the child's buttocks and encourages hip abduction; it also provides some guidance to the right position when the child sits down from standing. A stretch terry cloth cover has been most effective in keeping the child in place on the seat, particularly with steeper angles. Three more severely involved children were given a more confining seat, to be used in a more static way, with the child held in a fixed position of partial weightbearing and not coming to stand from the seat.

Figure 2. The components currently in use, referred to in the text by letter.



Seatbelts were provided on about half of the desks, with the instruction to use them when the child was to sit for longer periods and not have the close supervision we consistently recommend when the desk is to be used dynamically.

Trays were provided for all 25 of the desks [J]. The design of the trays has not changed since the second desk; most have been of clear quarter-inch polycarbonate to allow the child to see more of his/her body and reduce the isolation from his/her surroundings. The brackets into which the tray slides [K] are fastened to the frame with knobs so that the family or staff can easily adjust the

height and angle of the tray for different activities. The material and design of the bracket have been changed due to problems of durability with constant use and adjustment. We have also been using commercial tray locks (8) to allow two positions of the tray without moving the bracket, and easy removal of the tray when getting the child in and out of the desk. Many children have learned to use the tray as a support surface when coming to stand, shifting their weight forward and pushing down with their hands.

Trunk support has been provided in a variety of ways, depending on the child's need for support and and our own level of technology at the time the desk was provided. We are now using the same crosspiece for chest or back support that we use to hold the tractor seat and the knee block [B]. This piece is generally used in front of the floppy or flexor-dominated child, and behind the child with extensor-dominated patterns, though we usually determine placement of the piece after the child is seen sitting on the seat with other components in place. The cross member is also slotted to allow mounting of lateral trunk supports. Straight [C] or angled [D] lateral pads are chosen according to the height of the support on the child and whether or not they interfere with upper extremity function. Nine of the children have used their desks with chest pads, four of these with a strap around the back as a safety precaution, and 17 children have used back pads of various styles. Lateral trunk supports have been placed on 14 desks.

Only four children have been able to use the desks standing directly on the floor of the frame with no additional support for foot placement. The rest have had an adjustable angle footplate [E] with some type of shoeholder. The original positioners had a shaped ethafoam heel cup mounted on a plywood plate, with a two-inch wide elastic strap across the ankle, looping through a D-ring and snapping [F]. While this design has proven most effective for holding those children who tended to pull their feet out, it was awkward to fasten and the uncoated foam was difficult to keep clean. We also experimented with rubber webbing, velcro, and formed plastic shoeholders, but the most adjustable and simplest holders have been the wooden "sandals" with one-inch webbing and velcro straps [G] which we purchase from a commercial manufacturer (8). We are now requesting that the child be provided with proper shoes for standing in the desk, a higher top laced or velcro-closed shoe which will stay on the foot better. Children who use AFO's or inhibitive casts generally wear them when using the desk.

The component which underwent the greatest changes, both in our philosophy of how and when to use it and in overall design, is the knee block. Our earlier attempts at providing a knee block were too difficult to operate and keep adjusted in the right position. It was only after we devised a piece using our interchangeable crosspiece with a thick ethafoam pad which slips into and out of place without changing the position of the cross member [H] that we began using the knee support more consistently. This can now be fitted very closely, making the desk much safer and allowing children with less muscle control or strength to achieve standing independently.

RESULTS

Of the 25 desks provided, we were able to document 14 that are still in use. Nine have been discontinued for reasons including achievement of goals (three children began walking), medical problems, physical growth, and changes in programs or therapists. Those who have discontinued their use of the desks had used them for periods of time ranging from 7 to 22 months. Two of the desks which are still in use have been used for over three years, and these children have been seen for re-evaluation and adjustment or modification of the desks several times. In addition to accomodating physical growth in these followup visits, we have removed support for children who have made progress in their trunk control and ability to sit and stand, and added or modified supports to block the abnormal postural reflexes and movements which develop with increasing muscle tone in more severely involved children. We are finding that the desk is less appropriate for children with severe spasticity as expressed in high levels of extensor tone and frequest thrusting, requiring much more customization of each component and more frequest monitoring by our staff. We discontinued the use of the desk by one child whose apparent behavioral use of his abnormal reflexes posed a threat to his safety in the desk.

CONCLUSION

While we are continuing to provide positioner desks to children with a consistent waiting list of 5 to 10 referrals, we are now looking into other ways to incorporate this type of positioning and support into children's gross motor programs.

ACKNOWLEDGEMENTS

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ABSTRACT

Patients requiring a tracheostomy for ventilator support sometimes use special tracheostomy tubes that enable them to speak. These tracheostomy tubes require an air control port to be covered to direct air across the vocal cords. Control of the air flow can be done by others or by the patient unless neuromuscular impairment prevents this. For these patients, equipment has been developed to allow independent activation of the air flow. Both freestanding and wheelchair based systems are described.

INTRODUCTION

Providing means of communication for ventilator dependent patients is an important part of their health care. The standard tracheostomy tube used with these patients are designed with an inflatable cuff to ensure air flow down into the lungs. As a result, air does not pass over the vocal cords for phonation. In some cases nonverbal forms of communication such as writing or keypad can be used. Patients with quadriparesis due to spinal cord injury or neuromuscular disease require other alternatives to allow communication. Two methods often used at this institution are 1) To partially deflate the cuff allowing air to flow up across the vocal cords during ventilation and 2) to use a talking tracheostomy tube instead of the standard type. It is this second method of vocalization which utilizes the independent actuation technology.

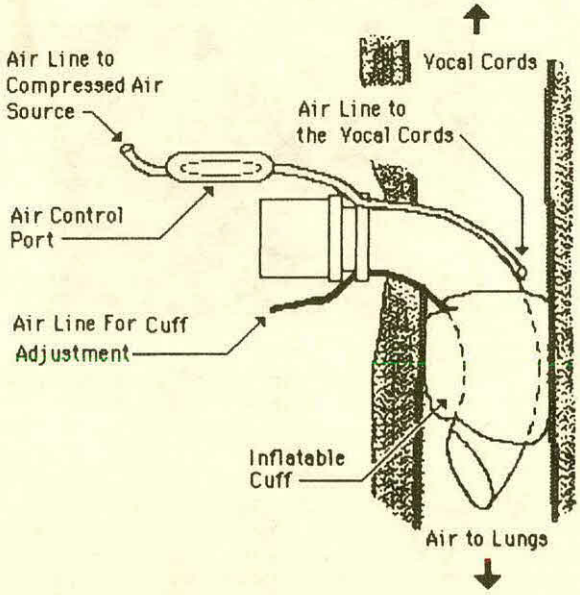

Figure 1: Schematic of the Portex tracheostomy tube.

The Portex model (Portex Inc., Wilmington, Mass.) is one such talking tracheostomy that is often used. A schematic of the Portex is shown in Figure 1. It is a single-cuffed tracheostomy tube with an additional small gauge air line running along its curved surface and ending just above the cuff. The external end of this air line is connected to a source of compressed air that is regulated to aproximately 5 liters/minute, usually enough to produce audible speech. A "tee" in the air line provides a port to control the air flow to the vocal cords. Air escapes out of the control port until covered by a finger to direct air through the air line to the vocal cords. Patients who are unable to cover the air port are usually dependent on assistance from another person. Systems have been developed enabling these patients to control their talking tracheostomies. Four different variations of these systems, including their portabilization, are presented.

PATIENT EXPERIENCE

Case 1:

The first system was developed in the fall of 1981 for a 71 year old woman who had A.L.S. complicated by pneumonia. During her hospitalization a refferal was received for an assistive device to allow her independent control of her talking tracheostomy. The original prototype used a normally open (low pressure) solenoid valve placed in the air line as a replacement to the air control port described above. In the open state air escaped through the solenoid valve. When activated the valve closes directing air up and across the vocal cords. Resistance in this normally open solenoid was such that a small amount of air continuously flowed across the vocal cords causing dryness and irritation. To alleviate this problem a new solenoid valve was chosen that could handle the high pressure air flow from the wall outlet in the hospital room and was placed between the wall outlet and a flow meter. The patient activated the solenoid valve with her hand using a custom paddle switch.

Eventually the patient recovered from the pneumonia and was released from the hospital. The system was made transportable by using a compressed air tank placed on a standard tank cart. A solenoid valve and a humidifier bottle were also mounted on the tank cart. The solenoid valve required an external 110 VAC outlet. A significant drawback of this system was the refilling of the tanks that was necessary.

Case 2:

This system was developed in the spring of 1983 for a 35 year old man with rapidly progressive A.L.S. During his hospitalization the patient became ventilator dependent. Due to his lack of motor control an automated talking tracheostomy system was desired. A bedside system was used by the patient who valued the independence it provided. As a result, a portable talking tracheostomy system was fabricated to give him the same function when in the wheelchair.

A platform mounted to the bottom of the patient's manual chair held the ventilator and a battery. Pressurized tanks were chosen as the source of compressed air. The tanks were mounted to the back of his chair so that they did not interfere with the ventilator when the chair was fully reclined. The problem of refilling the air tanks was alleviated by the generosity of a local fire station that agreed to refill the tanks as needed.

A new solenoid valve was chosen for this system that could be powered by the 12 volt ventilator battery. A momentary switch, connected to an electronic latching relay, controlled the solenoid valve. The user interface was a Zygo lever switch (Zygo Industries, Portland, OR.) which was activated by the patient's head. Upon activation of the momentary switch, the solenoid valve would latch open allowing air to flow through a humidifier bottle and to the talking tracheostomy. A second activation of the switch turned the air flow off.

Case 3:

This system was designed in the summer of 1983 for a 17 year old male with a C1-2 complete spinal cord injury. In the design of a portable system for this patient the problem of refilling spent air tanks was a major concern of the patient's family. A design was therefore implemented which utilized a portable air compressor, powered by the ventilator battery. The system shown in Figure 2, as well as the patient's ventilator and battery were successfully mounted to his electric chair.

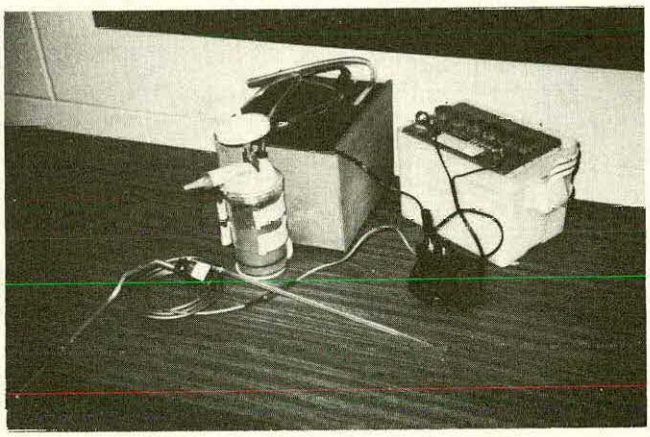

Figure 2: Components of the compressor based talking tracheostomy system.

A wooden box, built to house the compressor, was monuted underneath the wheelchair seat and in front of the structural cross bars. Great care was taken to minimize the noise of the compressor. Two air filters were placed on the inlet port and high density polyurethane foam insulated the compressor inside the box. Air leaving the compressor passed through a humidifier bottle which was held in a plastic bracket and mounted to the back of the wheelchair's legrest.

The control switch was the only real visible part of this system. A momentary trigger latching switch was again used to activate the solenoid valve. A simple microswitch was fabricated and operated by jaw movement.

Case 4:

This system was developed in fall of 1984 for a 49 year old man with C5-6 quadriplegia from an episode of polio at age 15. This patient utilized the automated talking trache system at his bedside using an environmental control unit. He became a proficient user of the system so that when he acquired an electric wheelchair the next logical step was to provide him with a portable system.

Figure 3: Mounting for the oxygen tank based talking tracheostomy system.

This patient required oxygen assist at a rate of approximately 2 liters/minute on his ventilator. For this reason, an oxygen tank provided the compressed air for his talking tracheostomy. A ventilator tray was mounted to the back of his power reclining electric wheelchair. Figure 3 shows how the wheelchair controls and the talking tracheostomy system were mounted on the back of the chair. A modified tank cart was attached to

the chair back so the tank could be easily replaced. The hardware was mounted such that the chair could be fully reclined without interference.

A Y bracket was attached to the oxygen tank regulator to direct flow to both the ventilator and the talking trache system. Flow meters were attached to each branch allowing each to be adjusted independently. Because the patient was in the chair for only a few hours a day, the ventilator and the solenoid valve were powered by the wheelchair batteries. An automatic switching relay was built and added to this electrical circuit to alternate the 12 volt drain of this system evenly between the two batteries.

The patient had enough upper extremity function to drive the electric chair, operate the recline switch, and a microswitch to activate the talking tracheostomy system independently with his left hand. The microswitch was used to trigger an electronic latching switch as has been done in previous systems.

DISCUSSION

These systems have been useful in allowing the patients to independently control their talking tracheostomy systems thus eliminating their dependency on other persons. Feedback from therapists working with these patients indicate that the independent voice control has enhanced communication as well as other aspects of these patients' treatments. For patients who have returned to the community it has expanded their ability to communicate with family and friends.

To assure that these systems are used properly and safely they were well documented on the patient's chart. In addition, a brief set of instructions was placed on or near the equipment indicating the proper set-up and maintenance required. Because these devices are customized for each patient it is our responsibility to repair anything that malfunctions. This can be time consuming, but is unavoidable.

When deciding between compressed air or an air compressor as a source, several considerations must be examined. Using the storage tanks requires that they be refilled when empty. If the patient is going to be in the hospital or a nursing home this can usually be handled by the respiratory therapist. If, however, the patient is going to be at home, arrangements have to be made for refilling. Although the portable air compressor alleviates this problem one must still compensate for the noise produced by the running compressor. For the case described here the noise did not pose a major problem.

The four systems described have all been extremely successful both from the patients' and the therapists' point of view. It is expected that there are a significant number of patients with tracheostomies who could benefit from similar systems. We continue in our attempts to improve the system design and implementation in order to make them simpler to set up and use.

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DIAGNOSTIC INTERVENTION TO DETERMINE SKILLS AND SYSTEM REQUIREMENTS FOR A NON-ORAL, BLIND, PHYSICALLY DISABLED ADULT

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ABSTRACT

An alternate form of communication was successfully developed for a non-oral, blind, physically disabled woman after twenty years as a passive respondent. Active and graded manipulation of message sender and receiver variables was undertaken to develop successful communicative interchanges. The issue of how to assess and develop functional capabilities prior to recommending a sophisticated electronic device are discussed.

INTRODUCTION

The reaction of a client to an alternate communication system and his/her successful use of it is influenced by several factors. The client's personality characteristic of having been a "talker" or "listener" prior to becoming non-oral must be considered in relation to the time spent as a passive respondent (2). The rigidity of the client's routine, his/her adaptability to change, his/her recognition of self-capabilities and willingness to use new strategies must be balanced against the known demands of available technology. Equally important are environmental considerations. These include the opportunities available to communicate, the expectations and demands of communication partners and their attitude towards an alternate form of communication (1).

The following case study demonstrates the importance of going beyond the initial step of matching client needs to an appropriate system. We must consider actively intervening to alter both message sender and receiver roles using a personalized, graded approach to ensure the success of a better, alternate form of communication.

BACKGROUND

Violet, a 40 year old native woman, suffered a cardiac arrest during an appendectomy at the age of twenty resulting in spastic quadriplegia and uncontrollable dystonic movements of limbs, face and trunk; blindness; and loss of speech. She has resided in an extended care facility since that time.

An initial communication assessment in 1965 revealed functional language comprehension in the presence of a severe dysarthria. Five months of speech treatment and follow-up evaluations over the next fifteen years indicated minimal improvement in functional speech. Violet was initially introduced to a temporary, alternate form of communication in May 1983. However, the prerecorded spoken "need" messages developed for her and accessed by head activation of a pillow switch were used decreasingly over the next few months. In September 1985 a request was received to reassess the situation.

EVALUATION

Physical/Sensory Abilities

Throughout the day, Violet was seated in a manual wheelchair with chest, thigh and ankle straps to offset strong extensor spasms. Unexpected stimuli resulted in a total body startle reaction. Attempts at voluntary movement resulted in increased spasticity and uncontolled ballistic movements. It was observed that Violet oriented her head fairly accurately to voices and meaningful sounds. Consequently, head rotation and forward flexion were systematically examined by having Violet activate a lever switch and a buzzer by using first her left cheekbone and then her chin. The right side of Violet's face was reportedly hypersensitive to touch and therefore not examined for potential switch access. Head rotation was demonstrably the better movement. Violet activated the buzzer on demand in an average of 5.6 seconds (range of 4 to 9 seconds), maintained pressure on the switch an average of 6 seconds (range of 2.2 to 10 seconds) and released the switch on demand in an average of .5 seconds (range of .2 to .9 seconds) over 10 trials. Adequate coordination of these movements was evidenced when she accurately imitated three part Morse code signals, a hearing screening could be completed, and an accurate response system was initiated with two buzzes for "yes" and one for "no" to supplement oral output.

Cognitive/Language Abilities

Although previously unrecorded or unknown, Violet reported having completed Nurses Aide training after graduating from high school and then working for two years in her profession. Her current cognitive/language skills were assessed informally using a series of graded auditory scanning tasks and switch activation to select the requested target.

Speech discrimination skills were assessed first using bisyllabic words. When given the task of discriminating categories of words, Violet successfully indicated both the category and the vocabulary item. Spelling skills were assessed using Violet's first name and then increasingly less personal vocabulary. Anticipatory errors on the preceding letter were found twenty five percent of the time in otherwise correct spellings. At these times Violet indicated an error by saying "no". Appropriate thought formulation, topic maintenance, sentence structure and grammar were evident in general conversation with Violet using the spelling tape.

Communication Environment

Communicative interactions with Violet using the twenty question approach occured primarily during one-on-one activities of daily living and therapy sessions. She spent the rest of her time in her room and had few visitors. This personalized,

socially isolated routine was reportedly due in part to Violet's preference for select contacts and activities and in part to unquestioned staff acceptance of her dependence.

INTERVENTION

To achieve a successful transition from a somewhat rigid, passive communicative lifestyle to a self-initiated interactive one, graded steps of change were required.

1. The first step was to maintain the typical content of Violet's interactions, expressions of need, while varying the method in which these were communicated. The auditory scanning system developed for her involved the attachment of a positionally adjustable lever switch to her wheelchair with a connection to a portable audio cassette system. Violet scanned prerecorded cued messages on channel A heard at a soft volume through a pillow speaker mounted near her left ear. She held the switch closed using her left cheekbone to access prerecorded expanded messages on channel B projected through a second speaker at conversational volume.

2. The next step involved helping Violet to recognize her potential range of communicative topics. Various message samples were introduced in role playing practice and in prearranged conversations with staff and residents. After one month of training Violet began editing message content spontaneously to create such messages as "I'm bored, what could we do differently today?" Content added by the therapist included conversation starters such as "What's new with you these days?" and "Heard any good jokes lately?" Violet's interest in clarifying and creating messages, increased the use of a spelling tape. She had to this point resisted its use, disliking her spelling inaccuracies. Continued training helped her realize that message interpretation was more important and that in fact, telegraphic spelling and sentence formation facilitated faster information exchange.

3. Finally, variables in Violet's interactive environment were manipulated to create the need and reinforcement for Violet to communicate using her new system. A staff inservice was provided using active dramatization and handouts. They were informed that the success of using this system was as much their responsibility as Violet's and had bearing on whether a more sophisticated, costly system would be recommended for purchase. Violet's preferred times for interaction were agreed upon by staff and added to systematically over time. To extend Violet's range of communicative partners a letter dictated by Violet containing autobiographical information and an invitation to come and chat, was distributed to all staff and residents. The next step will be to train a volunteer to communicate with Violet and become involved in expanding her interactions through available social/recreational programs.

SUMMARY

Violet's alternate system of communication was quickly accepted and used as a result of the system's adaptability, the graded assessment and realization of her capabilities, and the aggressive role taken in manipulating her environment. The flexibility of using an audio tape system allowed active manipulation of rate, style, quality and acceptability of voice output as well as ease of content editing, creation and sequencing. The limitation of a restricted message set proved an asset by enhancing the use of a spelling system. The opportunity for Violet to provide input to her system's development created the motivation to extend herself beyond passive learning and recognize her own communicative potential. Active listener involvement in manipulating equipment during conversation helped staff to recognize and change past communication patterns and realize Violet's potential. Violet now feels a strong need to initiate communication independently. Plans for acquiring a portable system with auditory scanning, switch input, synthesized voice output and print capabilities are now underway.

ACKNOWLEDGMENTS

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ENHANCED COMMUNICATION FOR A SEVERELY DISABLED DYSARTHRIC INDIVIDUAL USING VOICE RECOGNITION AND SPEECH SYNTHESIS

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ABSTRACT

A method is described to assist a speech impaired person to communicate. By utilizing voice recognition technology, a low cost system can interpret difficult utterances and represent the user's output in a variety of modes. Other features include paginated vocabulary to minimize misinterpretation, and the ability to control the user's environment. An example, in the form of a case study, is given of a severely physically disabled 18 year old girl who is also visually impaired but possesses the ability to produce semi intelligible utterances consistently.

INTRODUCTION

Speech technology is one of today's most active technological research areas.(1) In spite of all the effort channeled into this area however, change in its implementation, especially speech recognition systems, is slow. Until very recently, most of the recognition systems on the market used identical filtering functions, and most systems under many thousand dollars are speaker dependent (must be trained specially by each person who will use them). Speech synthesis systems are making more rapid improvement, though the better systems such as the DECtalk are still quite costly.

Both of these branches of speech technology will be widely applied, and are even now having broad impact, especially in industrial settings. The volume of demand justifies an active, competitive market with hardware and software offerings that will vie with each other in features and price.

However, the market for special applications of technology for disabled people is, and probably will remain, small. As a rule, technology for the disabled is first developed in the broader arena of large scale industry, then adopted whole or modified for disabled individuals. Since this is the general case, professionals who work with the disabled, and disabled individuals themselves, watch the non-disabled marketplace to see what technology can be used to satisfy the needs of their clients and themselves.

Speech technology, and in particular speech recognition, will allow greatly enhanced interaction with the computers of the future for individuals with manual dexterity and motor control disabilities who have intact speech mechanisms.

A less obvious application for speech recognition is with the population of individuals who attempt to use speech as their primary method of communication but who are dysarthric (have difficulty producing speech due to impairment of the organs of speech or their innervation)(2) and whose speech may be partially intelligible to persons who converse with them regularly, but who are virtually unintelligible to the untrained interlocutor.

Present developments in speech recognition follow two lines: speaker dependent and speaker independent (systems which can recognize words spoken by anyone). Since the dysarthric voice will deviate grossly and unpredictably from the norm, these speakers will most likely use speaker dependent systems. The primary requirement for use of such a system is consistency of sound production. Even for someone who might only be able to produce ten discrete utterances, if he or she can produce those sounds consistently, the functional size of that individual's vocabulary could be increased by the use of *word selection schemes.*

For instance, in a two dimensional array of vocabulary elements, items could be selected by row and column. Ten utterances could be used to select one hundred words by having both rows and columns indexed by number. A third dimension could be added by the use of "pages" of vocabulary, each one carrying one hundred elements. In a technically unlimited proliferation, the limiting factor quickly becomes the memory capacity and cognitive ability of the user.

In another scheme, the voice commands would drive a cursor around on a screen of vocabulary elements. Here the utterances needed would be reduced to five (up, down, left, right, and select) or even three, with wraparound (up, left, and select) but would have to be used in conjunction with each other for effect. There is a tradeoff between size of the utterance set and number of utterances needed to access any element. The intended user's voice consistency is the most significant criterion in the scheme selection. The higher the number of reliable, distinct utterances, the simpler the selection scheme may be.

The research effort we have undertaken is to assess the feasibility of using a speaker dependent recognition device, essentially as an interpreter. It "listens" to dysarthric speech and outputs the interpreted word to a speech synthesis module or other output device, and to an environmental control module.(3)

This system is intended to allow disabled individuals to access computer systems at home and over the phone. It is one of a family of aids for the disabled called COMEC (COMmunication and Environmental Control) based on low cost personal computers. Several versions of COMEC have been designed(4) using popular brands of personal computers and their peripherals.(5) COMECs provide independence in communication and appliance control.

This paper will describe the research subject with whom we are initially experimenting, the hardware/software system we have devised, and some preliminary results of the subject's interactions with the system.

THE CLIENT

Laura is a severely physically disabled 18-year-old, visually-impaired individual with the ability to produce semi-intelligible utterances consistently. She attends a public school and receives special services there through a local rehabilitation center. Her academic functioning is below age level, though she does have a good memory. It took her just a few sessions to memorize the civil alphabet which is the vocabulary we are using for individual letter key replacement strings. Each word associates with the letter it begins with. This is necessary, even with a "normal" speaker, because using the names of letters would cause many recognition errors (b, d, v, p).

Her severe physical limitations make it impossible for her to use any direct selection or multiple switch input system; her effective use of even a single switch is questionable.

Her visual impairment makes it impossible to have a monitor as an integral part of any system she would use for communication. Her method of communication currently is by speaking, though she frequently must repeat herself, even when speaking to someone who is with her daily. For these two reasons, a speech synthesizer was included in her prototype system.

THE VOICE INPUT DEVICE

The IntroVoice II for the Apple Computer, sold by The Voice Connection (formerly Voice Machine Communications, Inc) is the system we began with, because of several factors. An Apple II is the computer accessible to students in her school, and the IntroVoice II is reasonably priced. Other reasons are technical. They include:

1. Transparency: This refers to the fact that the computer's monitor program does not detect any difference between input from the keyboard and input from the voice recognition device. This characteristic is central to the ability of the user to have access to all standard software and software features. To the extent that a device is not transparent, various finessing adaptations will be necessary. The IntroVoice II carries all necessary memory on its own board, does not interfere with monitor routines, and is therefore fully transparent.

2. Pagination: This refers to the compartmentalization of vocabulary so that not all words are active at one time. This factor speeds recognition and minimizes incorrect recognition, in effect "widening the spaces" between voice patterns. IntroVoice II allows pagination, with a common page feature that permits some words to be always active. [Figure 1].
3. Parameter Variation: The IntroVoice II permits adjustments to the accuracy of fit necessary to exceed the recognition threshold.(6) This may prove useful in adjusting the device for use by a person whose ability to exactly reproduce his or her own utterances is below par.

Recognition errors can occur in two critically different ways. The first is a recognition error where the voice input system recognizes a wrong word. This can be a significant problem, especially in our environmental control function, where an incorrect appliance might be inadvertently activated. The second type of error occurs when the recognition device does not match the spoken word to any of its templates in memory. This results in a "rejected" error message and no action at all is performed. This causes less of a problem, and changing the recognition parameters should decrease the frequency of this type of error. It should be noted that it also a "deletion" can also occur. This happens when the recognition unit does not respond when an utterance is made. These were not considered in this paper.
4. Vocabulary Construction Software: This should be relatively easy to learn and to use. IntroVoice II permits both construction or editing of vocabularies and training or retraining of voice patterns in a simple nested menu framework described below. It presents three concepts central to its use. *Prompt strings* are simply the words or phrases that appear on the screen while a vocabulary is being trained. They have no other use than as prompts for the speaker. *Key replacement strings* are the actual keyboard characters which will be sent into the computer whenever the voice recognition unit has identified a close enough fit with one of its pretrained words. These strings must be designed to match the application software. And *voice patterns* are representations of the actual utterance, based on their spectrum analysis, and stored as a template. One of the strengths of the system is that when specifying the replacement string, it can be any string, including the hexadecimal representation of control characters, or a sequence of characters including a return, etc.

The hardware of the IntroVoice includes a 68B03 microprocessor, audio spectrum analyzer chip, 16K of RAM (Random Access Memory) and 8K of EPROM (Erasable Programmable Read Only Memory) firmware. The IntroVoice II uses the signal processing chip from Interstate Electronics to implement a 16 channel filter bank. The filter bank covers frequencies from 200 to 7000 Hz. Response time for word interpretation is 50 + N msec., where N is the size of the active vocabulary.(7) The minimum between word pause is 160 msec. The length of each word/phrase can be from .08 to 1.25 seconds.

Installation of the IntroVoice is not difficult. The main circuit board is placed into any slot of the Apple. A second small circuit board is placed on the motherboard, between the IntroVoice main board and the keyboard interface chips. In this way, both voice input and keyboard input are simultaneously active. A microphone and treadle switch (optional) to activate input interpretation are included with the packaged system.

The IntroVoice is a template matching system. Several discrete steps for training each vocabulary are necessary. When the software that is included with the system is loaded, a menu appears. There are several other menus nested below this main menu, for vocabulary construction and editing, voice training, testing of recognition, actual recognition and modifying some parameters of recognition. Within the vocabulary construction option, one types in prompt strings and the desired replacement strings for up to 160 vocabulary words, and up to twenty pages. Using multiple pages helps decrease recognition errors because a smaller number of templates are compared to the voice input.

After the vocabulary is constructed, the operator saves it on a floppy disk and then moves to training mode. Each of the words in the specified vocabulary is printed on the screen and the trainer is asked to say each word three times. These voice templates also are saved on floppy disk. When recognition mode is specified from the menu, a file name is requested and loaded into the IntroVoice memory. Control is then returned to the Apple monitor and any application program can be loaded and run.

THE VOICE OUTPUT DEVICE

The speech synthesis module currently being used is the Echo II+. While the quality of the speech it produces is not the best that is commercially available, this card is made for the Apple and is compact, very reasonably priced and many institutions already have them. Laura has adapted to listening to this device easily. Since Laura cannot read the screen, we have included the use of two synthesis modules, one for her as a private preview channel, and a second, for speaking out loud. In the laboratory we are using a DECtalk as the second speech synthesis device. The cost of the DECtalk may be prohibitive for many users. In this case, two Echo II's would be used. The DECtalk is a much more sophisticated system(8) than the Echo II, providing the option of several different voices, and providing more human-like speech.

THE ENVIRONMENTAL CONTROL SYSTEM

The hardware and software that provide the ability to turn on lights, radios, answer the phone, etc., is the Bi-Comm PC-1. This device uses radio frequency signals sent through the home AC power lines to send signals to commercially available BSR modules, one of which could even be in a neighbor's home to signal the need for assistance(9) (provided there is no transformer between the two homes).

THE SOFTWARE

The software that ties together the hardware and software already mentioned is written in assembler and BASIC and is used as the application program to the voice input system. It ensures that the recognized strings are passed to the voice output device, and remembers the length of the last string, for purposes of error correction. If more than one string must be erased, the system erases one character at a time after the first.

Subroutines exist for sending the message to the printer, erasing the message, monitoring the length of the message (the maximum length for an individual string is 256 characters), speaking the whole message out loud, operating the environmental control unit, and invoking a help command. The help command can be called at any time and will speak and print to the monitor the name of the page you are currently on, and every vocabulary word that is available on that page and the common page (all the words that are always active). In addition, every time the vocabulary page is changed, an announcement informs the user which page they have moved to.

RESEARCH RESULTS

Laura worked with her Occupational Therapist in choosing her own vocabulary, labeling the pages, and memorizing the modified civil alphabet mentioned earlier. The chosen vocabulary was incorporated into the application program and the IntroVoice II. Work with the research subject directly was then begun by the researchers.

The initial results seen with the first subject were as follows:
The first time we attempted training and recognition with the client, we trained 10 trial words, and obtained a recognition rate of 45% with three training passes. Laura was anxious and unfamiliar with the system, the trainer, and the method of using the system. It took 45 minutes to train the 10 words, due in part to a habit of Laura's of confirming everything that is said to her, and due to the stress associated with the situation (causing her to have increased difficulty producing speech).

The second time we did training with Laura, we trained one page of her vocabulary, consisting of the "people" page [Figure 1]. This training took approximately 30 minutes, and when we entered into recognition mode, we obtained a recognition rate of 50%. Most of the errors occurred with similar sounding words such as Sharon and Karen, and Cathy and Artie, which sound very much alike when spoken by Laura.

The third training session was of the "feelings" page and the common vocabulary (a total of 16 words) [Figure 1]. In a test session immediately following the training session, the recognition rate for the "feelings" page was about 60%. One word on the common page ("correction") was very difficult to train (about 30 repetitions were required for training), and the recognition rate for this word was close to 0. It seems that Laura simply does not say this word consistently. A different vocabulary word was chosen for this function.

In one particularly encouraging sequence during this test session 10 words in a row were identified correctly by the Intro-Voice.

In the third training session above, of the 30 recognition attempts 11 were errors. Of these 11, 10 were rejection errors, and 1 was a misinterpretation. This was also encouraging because it is preferred that the recognition device reject an utterance rather than guessing an incorrect word.

Once training was completed, the next several work sessions with Laura were spent doing recognition testing drills. These statistics are still being gathered and interpreted, and will be used to determine changes needed in training words and recognition thresholds. Preliminary results of almost 600 repetitions of the vocabulary words in the "people" and "alphabet" nodes were as follows:
Correct recognition was 54%, rejected utterances were 28% and misinterpreted words were 18%. While the rate of correct recognition is not as high as anticipated, these two pages are the two largest in Laura's vocabulary. Recognition on the pages with fewer words will almost certainly be better.

Training itself is a tedious process, especially with Laura. The training word must be spoken to her, then the switch must be activated, then Laura must speak the word. The number of times Laura had to repeat several of the words was closer to ten than the three the IntroVoice usually requires. It should also be noted that many recognition errors and training inconsistencies occur because Laura has a tendency to begin speaking the word before the switch has been depressed. For this reason, as well as to allow Laura to work totally independently, the switch is either going to be eliminated, or a dependable switch activation site will be identified.

An unanticipated benefit of Laura's using the speech recognition system is that her speech intelligibility is greatly improved when she is using the system because she does not try to produce several words with an inadequate breath. Coordination of breathing is a problem for Laura, and with this system she is forced to be sure she has adequate air before she begins speaking. Since she is speaking these words in isolation, her enunciation itself is also better than she typically produces. Whether this improved articulation will have long term carryover, and be noted in other research subjects is not known. One clinician did report such findings(10).

One disadvantage when Laura uses this system is that her actual rate of speech production is reduced. The major implication of this is that the client will have to distinguish when to use this system (especially with unfamiliar interlocutors and for printed output) and when to rely on her own speech. This process should increase the sophistication of her communication strategies.

Several ways to improve Laura's recognition rate are possible. The first is to create a larger number of pages with smaller vocabularies on each. Another would be to choose vocabulary words so they sound very different. A variation of this would be to make nicknames or code words for one of two words that sound too similar. Still another way to improve recognition with this system is to change the reject threshold, making the recognition unit more or less demanding in its guessing when the spoken word does not exactly match any of the active word templates. In addition, Laura currently has limited understanding of the need to speak in her "typical" manner. She frustrates easily, and allows her emotions to alter her pronunciation significantly. She will be loaned a system to work with over the summer, both independently and with a therapist's assistance. By using the system and understanding the need for staying calm, Laura's rate of recognition will hopefully improve.

CONCLUSION and RECOMMENDATIONS

This client's form of communication and environmental control depends totally on familiar interlocutors and aides. She has no way of preparing messages in advance or storing them, speaking with strangers or operating appliances on her own. The prototype system described here attempts to offer her those options. From these initial sessions, the prospect for success of this system looks promising for this initial user.

Further investigation into the use of voice recognition systems with a larger sample population is indicated. Further statistical study of the recognition rate, vocabulary and the determination of appropriate vocabulary modification with this initial user is planned, as well as changing the reject threshold to determine its optimal setting for this particular user. Patterns of speech which are more and less consistently reproduced should be identified, with the possible long term goal of identifying consistent patterns of speech in the cerebral palsied and head injured population. It is proposed that this be done, in part, by recording the speech of several speech impaired individuals, doing spectral analyses of these recordings and comparing them. Of additional interest would be the comparisons of these voice samples to other 'normal' speakers.

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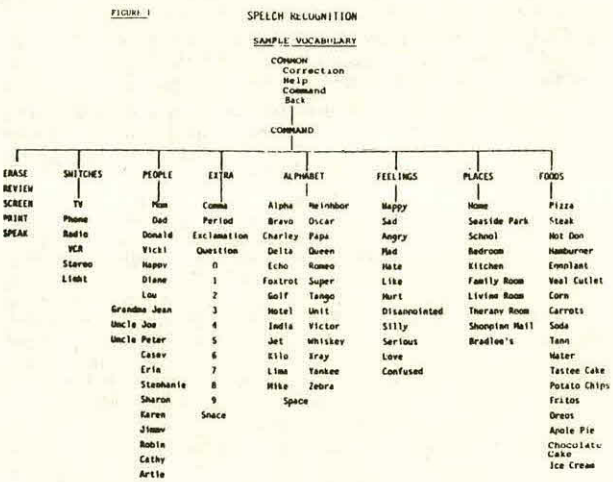

INTEGRATION OF MULTIPLE COMMUNICATION DEVICES: A CASE STUDY AND INCEPTION OF A MODEL

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ABSTRACT

A case study is presented of a nonspeaking, physically disabled individual whose communicative needs and capabilities indicated the need for continual access to a manual communication board, a portable voice-output device, and a mobility control for an elecric wheelchair. This paper examines the issues involved in the physical and functional integration of these assistive devices and identifies the need for a decision model for the selection and integration of multiple communication "systems" for some nonspeaking individuals.

INTRODUCTION

Technological advances in the field of augmentative communication have led to the availability of computerized communication devices, adding to the ever-increasing array of aided and unaided systems or options available for augmenting the communication of nonspeaking individuals. In spite of the potential of those computerized communication devices with a large storage capacity for language and multiple output modes, there may still be a need for several systems to meet the complex needs of a nonspeaking individual, given that individual's capabilities. When multiple aids are required to effect the best communicative interaction possible, the clinician is faced with the challenge of integrating the augmentative communication systems as well as integrating these systems with mobility and/or environmental control systems, in terms of their design and use in the natural setting. This challenge requires the co-ordinated efforts of a rehabilitation team.

The following case report illustrates such a challenge, the issues involved, and the solutions provided through the co-ordinated services of a multidisciplinary team.

CASE REPORT

The client was a 19 year old male, with severe motor impairments resulting from a head injury sustained at 5 years of age. He was dependent upon others for all activities of daily living but operated an electric wheelchair with a hand-controlled joystick. He had been followed by the Augmentative Communication Service (ACS) since age 13.

His latest assessment indicated that a recent change in residential setting from children's residence to adult group home, an anticipated change from an educational to a vocational setting, and the concomitant needs of his partners necessitated greater independence, efficiency and effectiveness in communication than was afforded by his existing means of communication. More specifically, the review of his current systems of communication revealed that he had no residual speech or vocalizations, but used facial expressions, eye-contact, hand and arm gestures, and a graphic display of over 400 Blissymbols to communicate. The latter he accessed directly, accurately and quickly by pointing with his right middle finger. His spelling and reading abilities were poor: he had a small sight vocabulary and was able to identify the initial consonants of many words.

Apart from the communication of propositionally simple messages (e.g., yes/no responses) which could be conveyed effectively and efficiently through gestures and/or facial expression, he was dependent for linguistic communication upon the proximity, ability and willingness of his partners to co-construct his message by repeating, noting and interpreting each item indicated by him. Many of his regular partners could not accommodate this dependency: they were nonspeaking individuals, nonreaders, were unfamiliar with the Blissymbol system, or had severe time constraints, and were therefore unable to co-construct his message. Thus his current means of communication were no longer adequate for all of his needs.

A computerized device with voice-output and a large programmable memory could potentially meet most of his needs for greater independence and efficiency in communication. The client's capabilities could not accommodate an abrupt transition to a technical device, thus the required communication systems included (i) a portable voice-output device to afford him greater independence in communicating with nonspeaking and nonreading

partners, greater efficiency in communicating with busy staff, and greater effectiveness in communicating at a distance; (ii) a manual communication board which he used effectively to contribute novel information on a wide range of topics with partners who were able to co-construct his message; and (iii) the unaided system of eye-gaze, gestures and facial expressions, which was an efficient and effective means of conveying non-propositional messages.

The selection of the communication systems guided their integration and the integration of the communication systems with the client's mobility system. Two major issues were considered simultaneously during the integration process: namely, design and use of the systems in the natural setting. For clarity these issues will be discussed separately.

Integration through design

The principle design requirements were specified by: (i) the client's motor abilities; (ii) his need, while in his primary position (i.e., a wheelchair), for independent and continual access to the mobility control, voice-output device, and large vocabulary on a manual communication board; (iii) his need to switch independently between communication and wheelchair driving; (iv) the relative importance, frequency and duration of communication, mobility and feeding relative to the available physical space; (v) the need for any "system" to be light, simple and quick to remove and reposition by staff who were busy and/or unfamiliar with his system; and (vi) the need for durability and stability to withstand frequent handling by many individuals and being transported from place to place. The main design features which met these requirements are illustrated in Figure 1: the labelled features are described briefly below.

1. A box housed both communication aids with their display surfaces permanently tilted at an angle which afforded greatest access and unrestricted view for wheelchair driving.
2. Cut-out for access to recharging unit.
3. Velcro fastener permitted easy access to display.
4. Auxiliary display permanently affixed to box: the vocabulary items were mounted on pages inserted into thick, transparent, plastic envelopes for protection. Rigid plastic sheet inserted into each envelope between two pages allowed the client to turn the pages independently and efficiently.
5. Metal bar extending from box took the weight of the open pages off the joystick.
6. Remaining space on tray was sufficient for cup, small plate etc. and for supporting his right arm when not being used.
7. Waterproof seal around voice-output device and a lexan sheet covering the graphic display on box.
8. Box mounted on tray by inserting pegs on base of box into holes in tray: this mounting plus the weight of the unit was sufficient for stability, but afforded its quick and easy removal, and accurate repositioning, by one person.
9. Programming keys protected from accidental activation by rigid plastic sheet under keyguard: sheet unscrews and swings away for programming.
10. Control box of joystick was mounted on undersurface of tray to avoid recurring problem of damage caused by improper positioning of tray.
11. Enlarged knob and elongated shaft of joystick maximized driving control.
12. Pages closed over box for driving: core vocabulary replicated on last page for convenience.
13. Cut-out for speaker to minimize loss of volume as a result of embedding device. Inaccessible volume controls were relocated to keys on display.

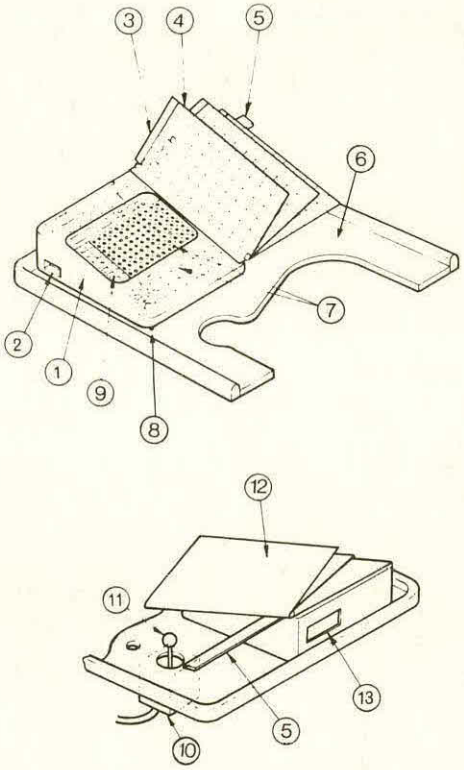

Figure 1. Major design features of integrated communication systems.

Integration through system use

Appropriate, effective and efficient use of the communication and mobility systems required: (i) the client to have technical competence in their operations; (ii) the client to have pragmatic competence in use of the systems to attain his communicative goals with a range of partners in a variety of contexts; (iii) the caregivers in the client's environment to be familiar with the systems and with strategies that best facilitate interaction with the client; and (iv) one individual to assume responsibility for functions the client could not do independently, such as programming the voice-output device and the maintenance of the entire system.

These requirements were addressed not only by instructing the client in the mechanical operation of the systems, but also by training effective use of the systems in interaction. In addition, in order to re-establish the client in his natural environment, with his newly acquired communication skills and systems, an individual training programme was provided for the person who was responsible for system maintenance and co-ordination of his ongoing communication programme. A training seminar designed to instruct caregivers in interaction strategies which maximized the client's use of the communication systems was provided.

DISCUSSION

Traditionally the goal of communication assessments of nonspeaking individuals has been to select a "system" of communication which best matches the abilities and needs of the individual. Various algorithms have been proposed for the election/selection of an appropriate augmentative communication system (1). The decision process has typically utilized an either/or rationale in which the advantages and disadvantages of one system are weighed against those of another (i.e., aided vs unaided, picture symbols vs Blissymbols). The availability of computerized communication devices have increased the options available to the clinician and have been viewed merely as additional steps in the decision process: electronic vs nonelectronic display; Device A vs Device B. Such decision procedures have typically culminated in the selection of a single augmentative system or technical device. Thus the provision of a high technology device such as a portable voice-output device generally heralds the removal of the manual communication board: rarely do "high-tech" communication devices co-exist with manual communication systems (2). Furthermore, the transition from a manual communication board to a computerized device is often viewed simply as a technical one: little consideration is given to potential problems faced by the nonspeaking individual arising from the increased communication possibilities afforded by the computerized device (2). This "single system" model has been challenged by recent research findings which have identified the multi-modal nature of communication. Nonspeaking individuals use a variety of systems or modes to communicate: residual speech, vocalizations, eyegaze, facial expression, and gestures, as well as communication boards and technical devices (3,4). Furthermore, there is growing evidence that these individuals shift among their available modes and devices as different communicative situations arise with different partners and contexts (3,4).

Our experience with this client and others requiring multiple communication systems, together with the research cited above, indicates the need to extend clinical intervention from a primary focus on training technical competency with a single system to include training technical and functional competency in multi-modal communicative systems. A "multi-system" decision model is required for the selection and integration of the various components: selection and integration through design and use are highly interrelated and must therefore be considered simultaneously.

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A MODEL FOR TRAINING SCHOOL PROFESSIONALS FACILITATING COMMUNICATION FOR STUDENTS REQUIRING AUGMENTATIVE COMMUNICATION DEVICES FOR WRITTEN COMMUNICATION

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ABSTRACT

The availability of aids for written communication, as well as the implementation of the Ontario Ministry of Education's Bill 82 (a policy allowing many augmentative communication users new educational opportunities) has created the need for a larger audience to understand the adaptations and use of high technology for children with special communication needs. To increase the understanding and skills of school personnel, a two-day workshop format has been included as part of the Augmentative Communication Service prescription and recommendation process. This paper will briefly describe the background, the content and future directions for this training session.

BACKGROUND

The Augmentative Communication Service (ACS) of the Hugh Macmillan Medical Centre, in Toronto, Ontario serves a large clientele of nonspeaking individuals in the province of Ontario. The mandate of ACS is to provide assessment and support services to nonspeaking clients requiring communication augmentation for face-to-face and/or written communication.

ACS intervention has always involved the client's facilitators - the family, school personnel, and care providers - who interact with the child on a daily basis. Since the inception of ACS, appropriate training for facilitators has been included as an essential part of the service delivery. A variety of training sessions are offered regularly to instruct facilitators in the development of face-to-face interaction strategies; aid in specific system application; and to facilitate the use of the microcomputer as a written communication aid.

Augmentative communication aids for face-to-face and/or written communication can be prescribed by ACS and funded through the Ontario Assistive Devices Programme. Stationary written communication aids are provided for the child's home use only. A two-day computer workshop is provided for the child's family to facilitate the use of the computer in the home.

ACS recognized that the provision of the microcomputer in and of itself was not sufficient training for the development of functional written communication skills. For some children, basic written communication skills would have to be developed in the educational setting before an appropriate prescription for home could be made. In recommending that a school board provide equipment for school use, the need for a new training format emerged. School facilitators would now require training in the use of the microcomputer as a written communication aid. Initially it was thought that general computer literacy could be provided by dealers or community colleges with some input from ACS. Most facilitators found that the amount of general information was overwhelming and largely irrelevant to their specific needs. ACS has since developed a two-day workshop in an attempt to meet this particular need of school personnel involved with the ongoing support and education of students using stationary augmentative communication devices for written communicaton.

THE WORKSHOP

The specific goals of the workshop are as follows:

1. To help facilitators feel comfortable using their student's recommended equipment.

2. To help facilitators gain some knowledge of hardware and software that support the development of the student's written communication skills.

3. To teach facilitators basic customizations of software (i.e., Talking Blissapple (1)) and other peripherals (i.e., Adaptive Firmware Card (2), Elementary Mod Keyboard (3)) that will allow them to exploit the potential of the system recommended for their student.

4. To have facilitators become aware of resources related to their system.

5. To have facilitators view the computer as a tool that facilitates

written communication for the nonspeaking child.

6. To have facilitators gain a perspective that will allow them to creatively use the computer as a tool to provide opportunities to develop the child's interaction skills.

7. To provide a forum for facilitators to share similar concerns and ideas for children with a range of special communication needs.

Participants

All workshop participants required prior involvement with the ACS client who would be using the recommended equipment. The school board was required to have the recommended equipment in the child's class before the facilitators could be referred to the workshop. These pre-requisites were stipulated because the hands-on workshop format lent itself to immediate application with specific children. ACS has encouraged school boards to send the child's aide, teacher, the board's computer consultant and speech pathologist. In most cases a combination of two of these individuals attended. The participants came to the workshop with a wide range of skills in computer literacy, augmentative communication, and curriculum development.

Equipment and Materials

The equipment typically recommended for a child's written communication system might include an Apple 1le system (4) with duodisk drive, printer, Votrax Personal Speech System (5), an Adaptive Firmware Card (AFC) with an expanded keyboard or single switch access, or an Elementary Mod Keyboard (EMK) System. All hardware, software, peripherals, manuals and customized materials needed by the child were available for the facilitators' use at the workshop. Manuals had been highlighted and marked ahead of time so that facilitators would not have to deal with an unnecessary volume of written material. Some custom set-ups for the students were also prepared ahead of time so that facilitators had the opportunity to experience the range of adaptations available to their particular student. To ensure that all facilitators had sufficient hands-on experience, one complete system was provided for every two facilitators.

Staff

This type of workshop requires a very high staff to participant ratio. Space and equipment constraints have typically limited the number of participants to twelve. Throughout most of the session one instructor introduced new material while another monitored the participants' pace. The planning and extensive coordination placed heavy demands on both clinical and clerical staff. Additional technical staff was required on an on-call basis to deal quickly with any equipment malfunctions.

Content

The content and organization of this workshop has been adapted over the past year to meet the very specific needs of the clients and facilitators. Approximately 40% of the time spent at the workshop dealt with the technical "how to's" of customization of software through peripherals such as the AFC. Another 30% of the time was spent examining communication software such as the Talking Blissapple, Mapwriter (6), Story Machine (7), Kidwriter (8),and Printshop (9). Handouts and highlighted manuals allowed participants to study and review additional material on their own.

Because the recommendation for such equipment is based on the need for an augmentative communication aid, the emphasis throughout all sessions was placed on the on-going development and facilitation of both face-to-face communication with the child's traditional system, and written communication skills with the stationary system. The enthusiasm for the potential that the microcomputer holds, must not override the aim that the equipment will increase the child's communication opportunities, and not contribute to further social isolation. Participants were asked to share their ideas in planning for the role of the computer in their classroom. They examined activities around the computer that could encourage independence, peer interaction, stronger social skills and the development of a positive self-image. Unique applications of software were encouraged to help the child achieve significant communicative and educational goals. A sample agenda follows:

Day One

9:30 - 10:45 Introduction to the Apple 2e

- role of technology in augmentative communication
- why an Apple
- parts/manuals/terminology
- special peripherals

- general care
- trouble-shooting

11:00 - 12:00 Introduction to the Blissapple
- use of the Blissapple with the regular keyboard and sample display, level 1 commands

1:00 - 2:30 Introduction to the AFC
- hands-on with Mapwriter
- intro. to AFC
- use of Mapwriter with sample custom AFC set-up

2:45 - 3:45 Creating Custom Set-ups
- participants create a custom set-up with AFC for Printshop

3:45 - 4:30 Small Groups
- review of any topics

Day 2

9:00 - 9:30 Individual Review

9:30 - 10:45 Adding and Changing Blissapple Vocabulary
- customizing Blissapple vocabulary for a particular student

11:00 - 12:00 Software Demonstrations
- Kidwriter
- Storymachine
- samples of custom adaptations

1:00 - 2:30 Computers and Communication Issues
- purpose of this system
- physical set-up
- programme development
- developing a communicative environment

2:45 - 4:30 Wrap-up and Review

Evaluations

All sessions to date have been evaluated through a questionnaire distributed at the end of the workshop. Responses to these questionnaires indicate that the workshop has been a valuable experience for facilitators at all levels of expertise. ACS will continue to offer this workshop while developing new formats to address the needs of a growing number of professionals involved with augmentative communication users.

CONSIDERATION FOR THE FUTURE

Although this format does address the goals outlined earlier on a short-term basis, there exists a need for augmentative communication technology to become more useable on a day-to-day basis. Students who have new facilitators each year may not be able to grow with the technological possiblities of their systems if the facilitators cannot readily integrate the microcomputer as an effective communicative element in the student's educational setting.

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DYNAMIC EFFECTS OF FUNCTIONAL ELECTRICAL STIMULATION ON SEATING INTERFACE PRESSURE AND TISSUE SHAPE

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ABSTRACT

Functional electrical stimulation may have application as a method for preventing pressure sores. This paper presents a pilot study investigating dynamic effects of functional electrical stimulation for this purpose. Pressure measurements are used to show that the seating interface pressure is altered during bilateral electrical stimulation of the gluteus maximus muscle. Ultrasonic imaging techniques display the effect FES has on the tissue shape.

INTRODUCTION

Pressure sores (decubitus ulcers, ischemic ulcers, etc.) represent a severe and costly problem for many disabled individuals. This is particularly true for those who are wheelchair dependent and have sensory loss. A research program has been implemented to determine whether functional electrical stimulation (FES) can be used to prevent the formation of pressure sores. This report is an extension of work previously presented (1).

It is generally accepted that the primary cause for the development of pressure sores is the local restriction of blood vessels due to external forces exerted on the tissue. There are a number of mechanisms through which FES can possibly help prevent pressure sores (see below). However, to develop stimulation parameters which are effective in preventing pressure sores, these "positive" effects must be weighed against possible "negative" effects which include: increased intramuscular pressure, muscle oxygen requirements, metabolite production, fatigue, heat, and sweat.

The use of FES may prevent the development of pressure sores by several mechanisms:

1) Tissue undulation and variations of seating interface pressures permitting increased blood flow.

2) Increased local blood flow to muscle and surrounding tissue in response to muscle contraction.

3) Long term effects of chronic stimulation including: increased muscle bulk and strength; increased vascularization; changes in tissue properties such as metabolic and muscle twitch characteristics.

Pilot Studies have been designed to study the tissue undulation and seating interface pressure variations produced as immediate/dynamic effects of FES. Interface pressure measurements were made using two different pressure measurement pads. Tissue shape changes were measured using ultrasonic imaging.

EXPERIMENT I - PRESSURE MEASUREMENTS

Materials and Methods

Bilateral stimulation of the gluteus maximus was performed on able-bodied subjects while seated in a wheelchair. A standard wheelchair seating position was defined as having the following conditions:

1) Backrest to seat angle of at least 80 degrees.
2) A minimum 2 inch clearance from the popliteal fossa to the forward edge of the wheelchair.
3) Footrest adjusted to keep the thighs parallel to the seat and floor.

Stimulation was provided through surface electrodes using a dual channel neuromuscular stimulator (Respond II - Medtronics, Inc.). Controllable parameters of the stimulator include amplitude, pulse rate, and signal "on" and "off" times. Cathodes were positioned bilaterally on the gluteus maximus, each approximately one-third of the distance from the sacrum to the greater trochanter, superior to the sitting surface. Specific location was determined through monopolar stimulation and visual inspection for maximum contraction. A common anode was placed on the sacrum.

Stimulation intensity was set with the subject in the standard seating position. With a frequency of 3 Hz., the intensity was adjusted to elicit approximately a one-inch medial-lateral movement of the knee. This level of stimulation was easily tolerated by all sensate subjects.

Initial pressure studies used a Scimedics pressure evaluator pad. This system measures pressure based on the pneumatic pressure inside a three inch diameter pad. When the external pressure exceeds the internal pad pressure a light sensor is activated. The interface pressure is taken as the pressure reading at sensor activation. The pad pressure can be adjusted between 0 and 250 mm Hg.

A series of trials was conducted on two able-bodied subjects to determine maximum pressure variations produced with FES. For these trials the Scimedics pad was secured beneath one ischial tuberosity. The interface pressure at that point was measured at rest and then during stimulation. The difference between the two pressures was recorded as the interface pressure change induced by FES. These trials were performed at each of five different stimulation frequencies: 3, 5, 10, 30 and 50 Hz; and on four different seating surfaces.

This pressure study was extended with the implementation of the Texas Interface Pressure Evaluator (TIPE) pad. The TIPE pad is 18 inches square and consists of a 12 X 12 matrix of activation switches. The entire pad is contiguous and each activation section is at the same pressure.

The internal pressure can be adjusted between 0 - 100 mm Hg. Each of the 144 switches can be closed or opened independently. A closed switch indicates that the external normal pressure at that point is greater than the internal pad pressure; an open switch indicates external pressure less than the pad pressure. The TIPE pad output is a 2-dimensional array of these closed and open switches.

A computerized interface was developed to facilitate real-time data collection of the TIPE pad information (2). Utilizing this interface, the TIPE pad output can be sampled at a rate of 30 times per second and stored on floppy disk for analysis. This system was used to dynamically study the effects of stimulation on the seating interface pressure.

Two able-bodied subjects were tested using the TIPE pad. The subjects were seated in the standard position on a 1 inch medium density polyurethane foam pad over a hard seat. Stimulation frequency was 50 Hz. TIPE pad data was collected for four different interface pressures, both at rest and during stimulation.

Results

Table I summarizes the results of measuring the interface pressure changes under the ischial tuberosities using the Scimedics pad. Only maximum interface pressure changes are recorded. All values are for a stimulation frequency of 50 Hz.

Table I. Maximum Pressure Variations with FES using the Scimedics pad

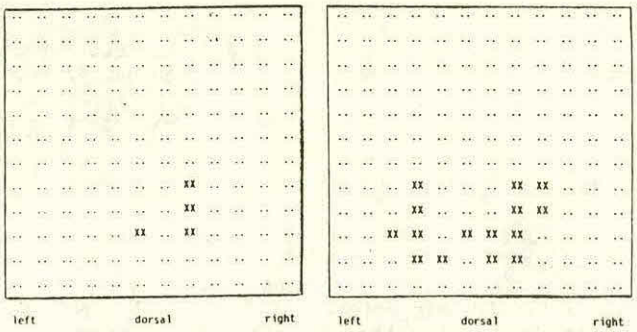
Surface	Subject A mm Hg. (mean + sd)	Subject B mm Hg. (mean + sd)
slung seat	9.44 + 1.42	10.11 + 1.69
1" gel pad	5.00 + 1.10	15.83 + 4.90
temper foam	2.67 + 1.15	10.67 + 2.31
ROHO	4.17 + 0.98	11.67 + 6.22

Extended stimulations were carried out on 5 subjects. Deterioration of this pressure variation was noted, indicating a possible fatigue phenomena. This fatigue effect varied greatly between subjects.

A sample of the results using the TIPE are shown in figures 1 and 2. In figure 1a, with no stimulation, only 4 of the squares on the TIPE pad are activated; In figure 1b, during stimulation, an additional 11 squares are activated. These are summarized in Table II.

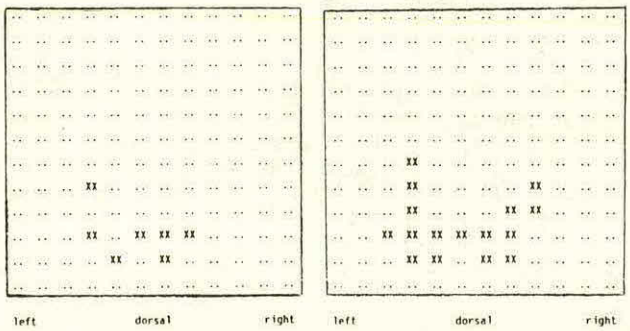
Table II. TIPE pad data summary

Figure no.	stim freq.	pressure	number of squares
1a	off	50 mm Hg.	4
2b	50 Hz.	50 mm Hg.	15
3a	off	45 mm Hg.	6
4b	50 Hz.	45 mm Hg.	16



a. no stimulation b. with stimulation

Pad pressure = 50 mm Hg.
Figure 1



a. no stimulation b. with stimulation

Pad pressure = 45 mm Hg.
Figure 2

EXPERIMENT II - ULTRASONIC IMAGING

Materials and Methods

An ultrasonic pulse echo image acquisition system (U.I. Octoson - Ausonics Corp.) was used to image the buttocks of five able bodied subjects under various seating conditions. The Octoson is capable of rapidly producing a repeatable high quality echogram. It has eight large aperture transducers immersed in a water bath. The transducers are placed around a curved arm that has five degrees of freedom of movement and is capable of producing echograms in transverse, longitudinal, or oblique planes. A video imager (Matrix Instruments) was used to produce photographs of the Octoson display screen.

Ultrasonic images were generated under three conditions:

A) Buttocks suspended in water with no external load and no stimulation. To obtain this view, the subject supported himself by his arms with hips flexed at approximately 90 degrees and buttocks suspended in the water bath of the Octoson.

B) Subject seated on plastic membrane with no electrical stimulation. This simulates the standard seating interface condition and produces an approximation of buttock distortion caused by sitting.

C) Subject seated on a membrane with electrical stimulation applied. This shows the effect of FES on the buttock tissue configuration during sitting. FES electrodes were placed and intensities set as described in Experiment 1. An image was generated while a bilateral stimulation of 50 Hz. was being applied.

Results

Figures 3, 4, and 5 show the images of the buttocks obtained from one subject in trials A, B, and C respectively. Noticeable distortion of the buttocks can be observed in trial B compared to trial A as would be expected. In trial C, the effects of FES evoked buttock contraction are clearly visible. The distortion produced by sitting is noticeably reduced and a shape much nearer that of the suspended buttocks (no load) is achieved. Similar results were obtained for all five subjects tested.

DISCUSSION

The pressure measurements utilized in Experiment I have a great deal of inherent error. One major problem is that the transducer alters the shape and properties of the seat interface. The Scimedics pad is particularly difficult to use because slight shifts in the subject's seating position produced large changes in the pressure measurement. Additionally, it only provides information about one point on the seating surface. The TIPE pad is much improved in that it can provide information about the entire seating surface. Shifts in seating position are recognized and accounted for. The TIPE pad's flexible design allows for minimal alteration of the seating interface shape.

Even with these inherent errors, the results show that FES to the gluteus maximus is able to alter the seating surface pressure distribution. The maximum pressure reading under the ischial tuberosities was always reduced during stimulation using the Scimedics pad. The results using the TIPE pad are even more revealing. For example, at 50 mm Hg., 11 additional switches were activated during stimulation. This implies that the forces at the seating interface were being redistributed. By using different air pressures in the pad, an accurate representation of the change in pressure distribution due to FES can be obtained.

Ultrasonic imaging showed how the body tissue shape was changed during stimulation. The shape during stimulation more nearly resembles the shape of the suspended buttocks. This may imply a reduction in tissue distortion as compared to the tissue shape while seated. This change in shape also helps to demonstrate the redistribution of forces exerted on the tissue during stimulation.

These pilot studies demonstrate that FES can vary seating interface pressures, produce tissue undulation., and reconfigure the shape of the buttocks under load. It is hypothesized that these effects can contribute to the prevention of pressure sores. Future experiments will include dynamic blood flow studies, investigation of long term chronic effects of FES, and clinical trials with disabled subjects.

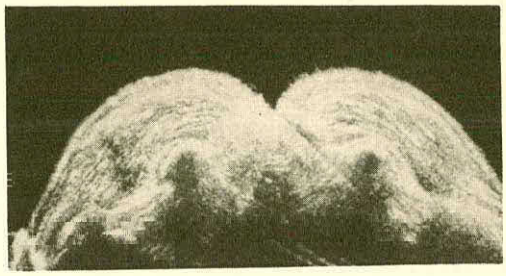


Figure 3. Buttocks suspended in water

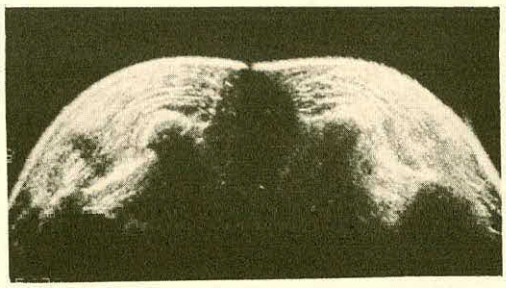


Figure 4. Subject seated on membrane

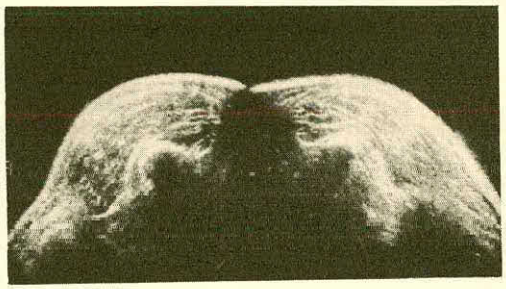


Figure 5. Subject seated, FES applied

ACKNOWLEDGEMENT

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FES REHABILITATIVE APPROACH IN INCOMPLETE SCI PATIENTS

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ABSTRACT

Electrical stimulation strengthening of paralyzed knee extensor muscles in incomplete spinal cord injured patients is described. Three different groups of patients were identified: (i) patients where an improvement of both voluntary and stimulated muscle force was observed, (ii) patients with an increase of stimulation response only, and (iii) patients with whom no effect of electrical stimulation training was recorded. Possibilities of orthotic use of functional electrical stimulators for correction of ambulation in incomplete spinal cord injured patients are also presented.

INTRODUCTION

In the last decades different traffic preventive measures, improved vehicle engineering, more efficient first aid, improved transport to the hospital and advanced treatment in an emergency center result in reduction of number of complete paraplegic and tetraplegic patients. As a consequence, many more incomplete cases are arriving in spinal units (1). According to statistical observations there is greater number of incomplete patients among tetraplegic than paraplegic patients (2). On the basis of these findings it can be concluded that incomplete tetraplegic patients represent an interesting group of paralyzed patients suitable for FES treatment.

METHODS

Because of the complex neurophysiological state of incomplete SCI patients it is impossible to predict the outcome of FES rehabilitation process when the patients are soon after the accident admitted to the spinal unit. Similarly, it is not possible to decide what rehabilitation aid the patient will need after recovering of the spinal cord injury. The first step in FES program is therefore application of therapeutic electrical stimulation.

The therapeutic electrical stimulation consisted of cyclic stimulation of partially paralyzed knee extensor muscles where stimulation trains of 4 seconds and pauses of equally 4 seconds alternately followed one another. The electrical stimulation was applied through large (6x4 cm) sheet-metal electrodes covered with water-soaked gauze. The electrical pulses used were rectangular and monophasic. A stimulation frequency of 20 Hz, a pulse duration of 0.3 ms, and a stimulation amplitude of sufficient intensity to bring the legs into full extension were used. During the training, the patients were positioned supine with both lower extremities semiflexed to approximately 30 degrees by a pillow under the knees. The FES session lasted half-hour a day. After the therapeutic electrical stimulation program in incomplete SCI patients an increase of the voluntary movement, strengthening of stimulated atrophied muscles, reduction of contractures, increased range of motion and lesser spasticity were noticed.

The effects of muscle strengthening program were tested and assessed through isometric knee-joint torque measurement. The patients were seated during the measurement on a special chair to which the electronic force transducer was attached. The isometric knee joint torque was assessed once every week in each incomplete SCI patient. The training program lasted for about two months.

Both voluntary and electrically provoked knee joint torques were assessed in a group of seven incomplete spinal cord injury patients. After the training program three different groups of incomplete patients were encountered: (i) In the first group there were patients where both voluntary and electrically stimulated muscle force were improved. (ii) In the patients from the second group the stimulated muscle force only was increased. (iii) In the third group of patients neither voluntary or stimulated response were augmented.

In general, the fourth group is also possible where voluntary muscle force would be increased with the stimulated muscle force remaining at the same initial level. Such an outcome is not logical and was not found during the investigation.

RESULTS

Seven case studies belonging to three different groups of patients are described in further text. The results of the strengthening of the right knee extensors in incomplete tetraplegic patient B.C. are presented in Fig.1. The 21 years old patient had incomplete C-7 spinal cord lesion after a fall. When the electrical stimulation training program started he was two months after injury. During 85 days of muscle strenghtening program, his isometric knee joint torque was measured once every week at 80 V and 140 V of m.quadriceps stimulation. The isometric torque achieved during the maximal voluntary knee extension was also assessed. The muscle exercise resulted in a rather linear increase of the torque measured. An increase of about 50 Nm was observed when comparing the initial and final results obtained both during electrically stimulated and voluntary muscle contraction. The patient, therefore, belongs to the first group of SCI patients as defined in Methods. The voluntary knee joint torque increased from almost zero to about 50 Nm what was found sufficient for unassisted walking with the help of two crutches. Patient was not using wheel-chair after leaving rehabilitation center.

There was considerable difference encountered between the final stimulated and voluntary muscle force in the patient B.C. In the following three patients almost the same maximal voluntary and stimulated knee joint torques were found. Nevertheless, the patients belong to the first group as an improvement of stimulated and voluntary movement was observed. The patient L.V., 30 years old, had an incomplete C-5,6 lesion after stab wound. She came from another country three years after

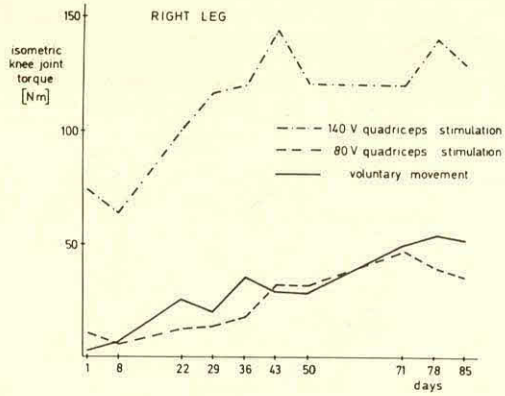

Fig.1

the injury. Because of rather strong pain sensation the stimulation amplitude was not increased over 80 V. The time course of maximal stimulated and voluntary torque are almost identical (Fig.2). The final result of the strengthening program was 50 Nm of voluntary knee joint torque. This was sufficient for unassisted standing. The patient was able to walk for short distances by the help of two crutches and two peroneal stimulators triggering right and left steps. She was occasionally using wheel-chair.

Another patient (M.K., 17 years) suffered C-5,6 incomplete spinal cord lesion from vehicle accident eight months before being admitted to the rehabilitation center from neurological clinic. Both stimulated and voluntary joint torque had quite low values at the beginning of the program and increased to about 30 Nm in the middle of the training process (Fig.3). The voluntary torque achieved was not enough to provide solid support. The patient was using right m.quadriceps stimulation during stance phase of walking while swinging of the right leg had to be triggered by peroneal stimulation. The patient was using wheel-chair to a considerable extent.

Similar results were obtained also in patient S.D., 46 years old, and suffering from a tumor at C-2 spinal cord level. She came to our unit six years after surgical intervention. Inspite of an increase in both voluntary and stimulated muscle force (Fig.4) she was unable to stand unassisted.

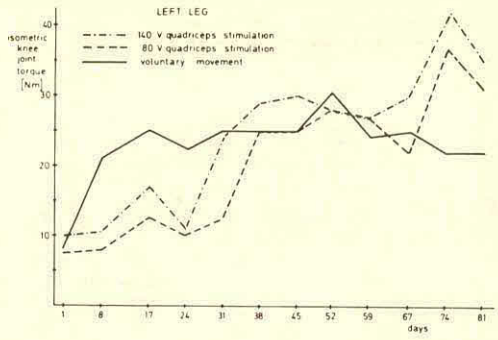

Fig.3

Standing and short distance walking was provided by unilateral m.quadriceps and n.peroneus stimulation.

Patient R.A. (42 years) suffered C-3,4 incomplete spinal cord lesion from car accident. Electrical stimulation training program was started eight months after the injury. The success of training program is shown in Fig.5. Both stimulated and voluntary joint torque had quite low values at the beginning of the program. The voluntary isometric torque remained at this initial value for the rest of the program. In contrary, the stimulated isometric knee joint torque was constantly increasing. Patient R.A. belongs to the second group. After the rehabilitation program the patient was able to walk by the help of a walker and bilateral m.quadriceps and n.peroneus stimulation on short distances only.

The patient Z.J. (26 years) also belongs to the second group. The incomplete T-11 spinal cord lesion resulted from motorbyke accident. The stimulation was delivered to him eight months after the injury. Here again, no improvement was observed in voluntary movement. Interesting is the knee joint torque trace belonging to 140 V stimulation amplitude. The noticeable drop in the joint torque coincided with a serious urinary infection (Fig. 6).

The last patient belongs to the third group of incomplete SCI patients. The C-6,7 incomplete patient D.A. (43 years) had a car accident. He came to

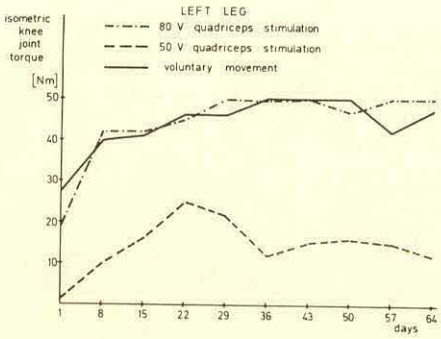

Fig.2

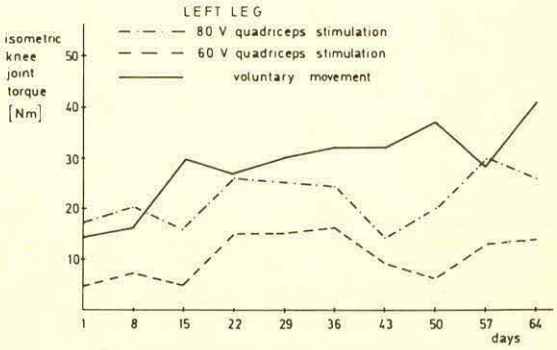

Fig.4

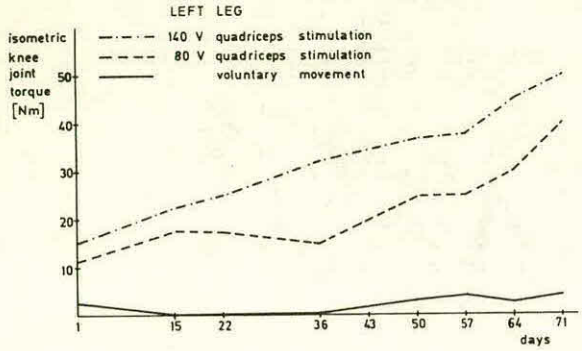

Fig.5

the spinal unit for the electrical stimulation purposes two years after the accident. It is evident from Fig.7 that no effect was achieved by daily stimulation of his knee extensors. The patient remained confined to the wheel-chair.

DISCUSSION

The final goal of electrical stimulation training program is restoration of ambulation in incomplete SCI patients by the help of functional electrical stimulation (3). In many incomplete paraplegic and tetraplegic patients exaggerated extensor tone can be observed in their lower extremities providing more or less safe standing to some of those patients. Many patients are unable to brake this exaggerated extensor tone during standing and hence they are unable to achieve adequate flexion for gait. It has been shown that in thoracic and cervical complete and incomplete patients electrical stimulation of an afferent nerve augments dorsiflexion, knee and hip flexion in a total lower limb flexion reflex pattern. In this way peroneal stimulation can be efficiently used to initiate a step in incomplete SCI patients.

Four different groups of patients were encountered with respect to their needs for application of different orthoses based on FES: (i) Incomplete SCI patients who were able to stand but required unilateral peroneal stimulation to elicit the flexion response and initiate a step.

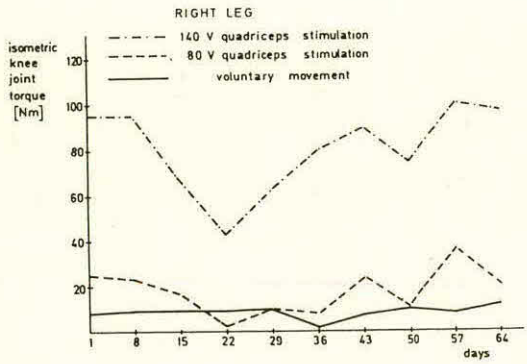

Fig.6

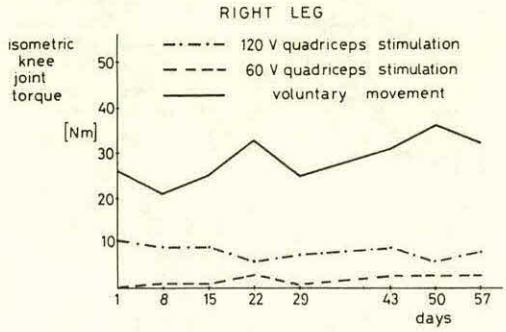

Fig.7

(ii) Patients where bilateral peroneal stimulation was found helpful. (iii) It was observed that in a great number of incomplete tetraplegic patients one leg was almost completely paralyzed while the other leg was under voluntary control and sufficiently strong to provide safe standing for short periods with only crutches. Unilateral simulation of knee extensors and an afferent nerve was helpful in these patients. (iv) In the last group there are patients where a minimum of four channels of FES are required for synthesis of a simple reciprocal gait pattern such as in complete thoracic patients.

ACKNOWLEDGEMENTS

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TRIGGER SWITCHES FOR IMPLANTABLE GAIT STIMULATION

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ABSTRACT

A microprocessor controlled, implantable gait assist system is being developed at our Center. In the proposed system, preprogrammed control sequences will be initiated and terminated by sensors. Initial systems will be radio frequency coupled, but the ultimate goal is to implant all hardware inside the body. We are looking for consistent events during the gait cycle which can be detected by either unobtrusive external or implantable sensors. Bilateral EMGs of four upper extremity and trunk muscles, crutch forces, foot contact patterns, and vertical acceleration of the thigh have been studied as possible trigger sources in seven incomplete spinal cord injured (SCI) persons. The timing and phasing of the triggers varied between subjects but were consistent within each subject. Foot contact patterns, vertical acceleration, and crutch forces were consistent in all subjects. The EMGs were consistent in most, but not all subjects. From the initial results it seems feasible to control the stimulating sequences with selected triggers.

INTRODUCTION

At the current stage of development, almost all functional electrical stimulation (FES) gait assist systems are used inside laboratories. To make FES practical outside of the laboratory the size of the instrumentation and ease of application must be greatly reduced and simplified. The goal of the FES gait assist program at the Rancho Rehabilitation Engineering Center is twofold: to improve the ambulation of stroke and incomplete spinal cord injured (SCI) persons to an efficient level; and to allow practical implementation of FES gait assist systems outside the laboratory. Ambulation is considered efficient when the energy expenditure is comparable to that of a wheelchair user (1).

Presently an RF coupled system is being developed (2,3). This RF coupled system can utilize both internal and external trigger sources. The external trigger sources studied were footswitches and force sensors in the crutches. The implantable trigger sources studied were the electromyographs (EMG) of four bilateral upper extremity and trunk muscles, and the vertical acceleration of the legs. These trigger sources were selected based on previous studies of normal gait (3,4,5) and analysis of pathological gait (6). An additional consideration in selecting external triggers was implementation with minimal interference to the user, an often overlooked area of concern.

This project is designed to develop reliable, unobtrusive external and implantable triggers to control stimulating sequences with minimal conscious effort by the user. In addition, these triggers can act as system safeguards to ensure the person is walking properly. If a trigger is not detected then the system senses this and can act appropriately. The footswitches and force crutches were selected for use with the initial systems, but hopefully will be replaced with completely implanted triggers in later systems.

METHODS

The subjects were incomplete SCI persons with walking speeds greater than 30 m/min. This criterion was established because recent studies at our Center have shown this is a minimum walking speed for community walkers. All subjects walked with a four point reciprocating gait.

Incomplete SCI persons were tested during level walking, ascending a 5° ramp, and descending a 5° ramp. Trigger sources tested were:

1. Footswitches; heel and metatarsal.
2. Forces on the crutches.
3. Bilateral EMGs of the latissimus dorsi, erector spinae, posterior deltoid, and lateral head of the triceps.
4. Vertical accelerometers mounted over the greater trochanter.

EMG signals were recorded with percutaneous 50 micron wire electrodes. All EMG signals were differentially amplified through a Biosentry EMG transmitter and recorded on a Honeywell Visicorder and magnetic tape. The signals were digitized, and then normalized to a maximum muscle contraction. The onset and offset times were then calculated using 10% of maximum as a cutoff level. The forces on the crutches were measured by strain gauges mounted in the crutch tip. Crutch contact was used to calculate onsett and offset times. The vertical accelerometers were fabricated in our lab. At heel contact a large oscillation in the acceleration was observed. This oscillation was used to determine the onset and offset times of the vertical accelerometers.

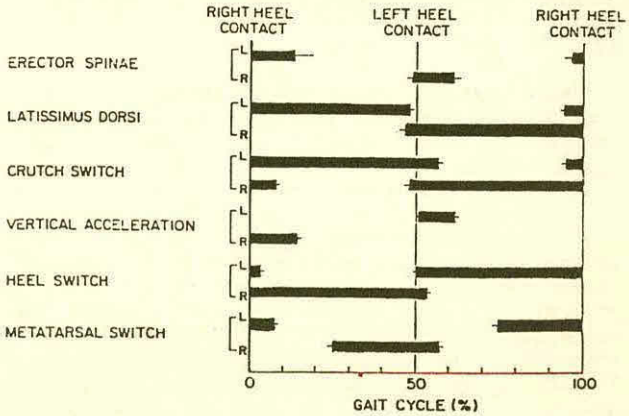

FIGURE 1

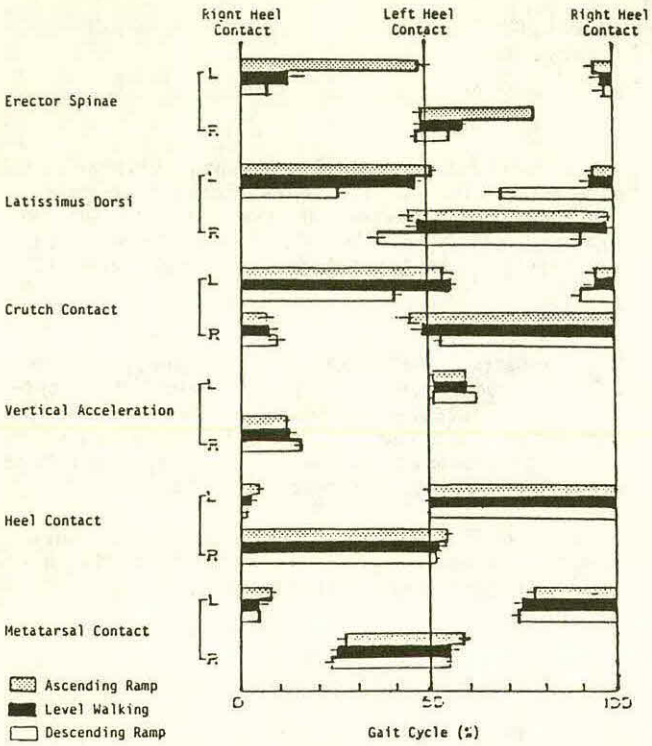

FIGURE 2

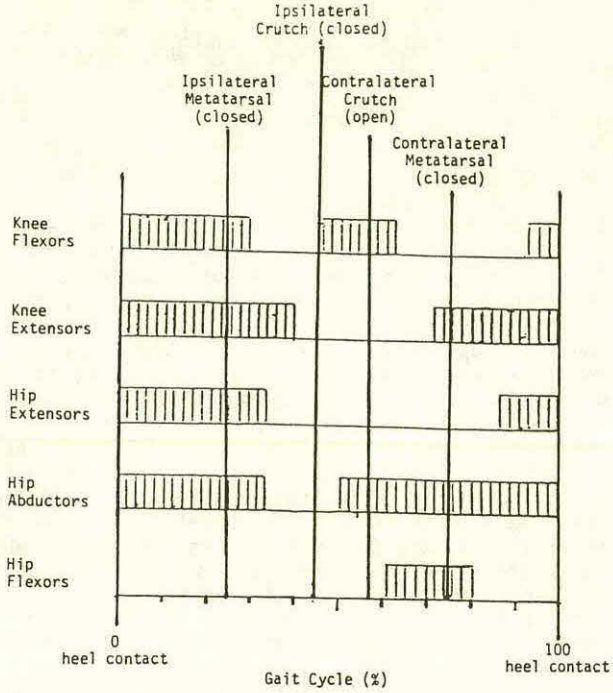

FIGURE 3

RESULTS

At this time, seven incomplete SCI individuals with walking speeds ranging from 31.3 m/min to 65.4 m/min have been tested. The data shown in the following figures are from a representive subject. This subject is a 31 year old male with an incomplete T12 lesion. He walks with forearm crutches and an ankle-foot orthosis. His level walking speed is 57.2 m/min.

Averaged activity of the triggers sources over a normalized step cycle (N=12) during level walking are shown in Figure 1. Standard error of the means (SEM) of initiation and termination of each trigger are indicated by horizontal bars. Note the SEM are less than 3% of the gait cycle for all triggers except the erector spinae. The posterior deltoid and triceps results are not shown because they were inconsistent.

A comparison of trigger timing during level and ramp walking is shown in Figure 2. The erector spinae and latissimus dorsi timing shifted dramatically, while the timing shifts for the other triggers were relatively small.

In Figure 3 the timing of the metatarsal and crutch switches are superimposed on a proposed stimulating sequence(8). The closing of the ipsilateral metatarsal switch with timing delays can terminate the stimulation to the knee flexors, knee extensors, hip extensors and hip abductors. The loading of the ipsilateral crutch can initiate stimulation to the knee flexors and hip abductors. Contralateral crutch opening can terminate knee flexion stimulation and initiate the hip flexors and knee extensors. The closing of the contralateral metatarsal switch can terminate the hip flexors stimulation and initiate the knee flexors and hip extensors stimulation.

Similiar results were seen in the other six subjects, although the timing of the triggers shifted, their consistency was very good. For the two subjects whose posterior deltoid and triceps activity was consistent, the timing was coincident with the ipsilateral crutch activity, implying that these muscles are used to stabilze the shoulder and elbow joints respectively.

DISCUSSION

The timing of the triggers can be understood if the gait cycle is analyzed. The subjects walk with a four point reciprocating gait in which the contralateral arm swings in phase with the leg. The contralateral crutch generally strikes the ground just prior to heel contact and lifts off just after toe off. As the crutches are loaded the latissimus dorsi becomes active to stablize the shoulder, which explains the coincident activity of the crutches and latissimus dorsi. The vertical accelerometers detect the impact at heel contact and therefore their timing is coincident with ipsilateral heel contact. As in normal gait, the erector spinae are active just prior to heel contact to stabilize the trunk for heel contact. The coincident timing of the implantable triggers with the external triggers indicates that the implantable triggers could replace the external triggers in later systems.

Certainly the timing and phase of each trigger will vary between subjects, but the microprocessor controlled FES gait stimulator will allow one to tailor, by means of software programming, the triggers to that person's gait.

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RELATIVE ENERGY COSTS OF LONG-LEG-BRACE AND FNS AMBULATION

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ABSTRACT

Two complete paraplegic subjects (T4 and T8), implanted with intramuscular electrodes, walked with functional neuromuscular stimulation (FNSW) using a rolling walker and a reciprocal gait. Subjects also ambulated with long leg braces (LLBA), using a rolling walker, and a drag/swing-to gait. Physiological data were collected during FNSW and LLBA. Energy costs (kcal/kg/min) were 59% to 76% of maximal aerobic power [(MAP) arm ergometry + Leg FNS] during FNSW; and 50% to 60% of the MAP obtained from arm ergometry during LLBA. As speed increased there was no increase in the energy cost of FNSW(Spearman Rank coefficient rho = -.40); energy efficiency (kcal/kg/meter) equalled that of LLBA. Recovery time, O_2 debt, and perceived exertion also decreased as FNSW speed increased. At speeds approaching 0.4 m/sec FNSW energy costs were similar to LLBA and energy cost relative to working muscle mass during FNSW was lower than observed during LLBA in the literature.

INTRODUCTION

Since functional neuromuscular stimulation (FNS) has the potential to be used as an orthosis for support and movement of paralyzed limbs, it is desirable to compare the energy costs of FNS walking (FNSW) to that of long leg brace ambulation (LLBA). The limited function and high energy costs of LLBA have been an obstacle to LLB acceptance by paraplegic individuals for daily use (11,12) (standard knee ankle foot orthosis, Craig-Scott orthosis, and reciprocal gait orthosis). The acceptance of FNS as an orthosis will largely be determined by the balance between the amount of energy used and the degree of function gained.

The purpose of this study was to compare the energy costs of FNSW to LLBA in terms of; 1. Energy cost: kcal/kg/minute; 2. Percentage of maximal aerobic power (MAP = maximal aerobic O_2 consumption) obtainable by the muscle mass used during activity; and 3. Energy "efficiency": (kcal/kg/meter). Oxygen debt was considered in conjunction with recovery time. Emphasis was placed on MAP since one can work for an indefinite period at or below 50% MAP (1).

METHODOLOGY

Two neurologically complete paraplegic subjects (1: T4, age 22 and 2: T8, age 25),were examined physically. No pulmonary or cardiovascular disease was found and the subjects had no secondary disabilities. They were implanted with percutaneous intramuscular electrodes in the quadriceps (vasti lateralis, medialis and intermedius), tensor fasciae latae, sartorius, gracilis, semimembranosus, posterior portion of the adductor magnus, gluteus maximus and medius, soleus and tibialis anterior.

Initial training consisted of the following FNS activities; stimulation of implanted muscle for an hour per day; standing between parallel bars; and isokinetic quadriceps extension exercises. Cardiovascular conditioning (arm ergometry) and LLB and FNS gait training were concurrent with FNS implantation. Prior to data collection, subjects were able to do LLBA for indefinite periods of time and distances at speeds between 0.1 m/sec to 0.3 m/sec. Both were able to do FNSW for distances greater than 200 meters at a rate of 0.3 m/sec with brief standing rests. MAP for arms was assessed using a standardized arm ergometry protocol (3). MAP for combined arms and legs was determined from the same arm ergometry protocol with the addition of FNS stimulation of lower extremities simulating gait with 20 lb weights on each leg. MAP was assessed in a sitting position.

Both subjects wore custom-made, rigid-ankle LLB with a 5-10 degree dorsi-flexed foot. Subject 1 had a pelvic band. For FNSW, a portable microprocessor-controlled stimulator (9.5 cm x 5 cm x 14 cm, weight 774 gm) was used to deliver programmed electrical stimuli to the implanted electrodes. A switch on the stimulator enabled the subjects to take each step at will. The stimulation parameters were; 0-150 microseconds pulse width, 25 Hz frequency, and 20 mA amplitude (9). Data was collected in the afternoon, three to four hours after eating, during a two week period. Subjects stretched and did range of motion exercises before LLBA and FNSW to reduce spasms. Basal metabolic rate in METs (basal/metabolic support requiring 3.5 ml O_2/kg/min) and heart rate were monitored until they became stable. The Beckman MMC Horizon with the advanced exercise system (SensorMedics, Anaheim, CA) was calibrated using standard gases, volume checks and temperature checks. The subject was instructed to sit quietly in his wheelchair for at least 15 minutes. He was fitted with a Hans-Rudolf face mask with a two-way valve, connected to the MMC (mobile metabolic cart), and data were collected for 5-10 minutes. The subject was told to walk down a 60 meter walk-way for 30 or 60 meters while the MMC was rolled alongside. The following parameters were recorded by the MMC; volume of O_2 consumed ($\dot{V}O_2$), volume of CO_2 expired ($\dot{V}CO_2$), respiratory exchange ratio (RER= $\dot{V}CO_2$ /$\dot{V}O_2$), $\dot{V}O_2$/kg (bodyweight), and METs Each subject walked as fast as he comfortably could with his current stimulation program or

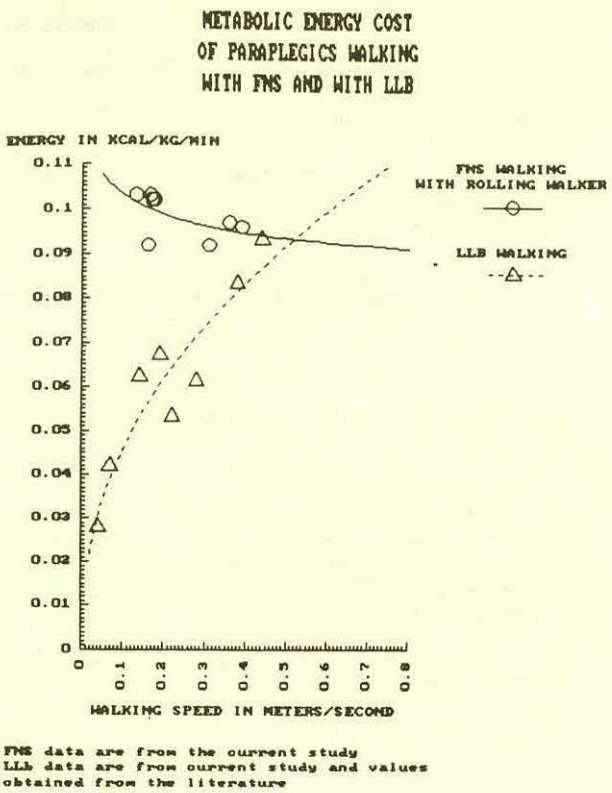

Figure 1

LLB. Data was collected continuously during activity, and following activity until the subject's MET level returned to resting values. Heart rate (HR) and blood pressure (BP) were taken immediately after activity. Data were transferred from the MMC to a MINC 11/23 computer (Digital Equipment Corporation, Maynard, MA), on which the following were calculated: O_2 debt, kcal/kg/min, velocity, kcal/kg/meter and recovery time.

RESULTS

The mean energy cost during FNSW was 0.095 kcal/kg/min (range from 0.087 to 0.103). Energy cost tended to decrease as speed was increased (Figure 1) (Spearman rank coefficient rho = -.40). Mean energy efficiency was 8.9 x 10^{-3} kcal/kg/m (range from 4.0 x 10^{-3} to 13.4 x 10^{-3}). MAP for arm ergometry was 6 METs for both subjects. For combined arm and leg work, the MAP was 9 METs (subject 1) and 8 METs (subject 2). During FNSW, percent MAP ranged from 59% to 76%. Mean velocity was 0.21 m/sec. Oxygen debt (range from 1.33-2.30 l O_2) was calculated as the difference between the total O_2 consumed during the recovery period minus the amount of O_2 that would have been consumed during an equal rest period. Recovery time was between 5.0 and 14.5 min. Mean HR and BP ranged from 72 beats/min and 102/66 mmHg at rest to 178 beats/min and 122/79 mmHg at the end of activity for subjects 1 and 2 respectively. RER ranged from 1.04 to 1.20 and was well above utility post walk.

Data for LLBA were collected in two trials. Results are reported for subjects 1 and 2 in order. LLBA required 50%/60% of MAP obtained from arm ergometry for velocities of 0.19 /0.22 m/sec. HR increased to 150 / 116 beats/min and BP increased to (112/80)/(114/80) mmHg after activity. The subjects incurred O_2 debts of 0.31/ 0.96 l and required recovery times of 2.0/5.3 minutes. The energy efficiencies were calculated to be 3.93 x 10^{-3}/5.63 x 10^{-3}kcal/kg/m (Figure 1).

DISCUSSION AND CONCLUSION

The 6 MET level obtained by both subjects during arm ergometry was less than one standard deviation away from the mean of wheelchair sedentary persons (those with injuries ranging from T7 to L2) (14). The increased MET level (33% to 50 %) observed when arm ergometry and FNS leg exercise were combined greatly emphasizes the importance of muscle mass. Since only two trials were conducted with LLB in this study and there was no variation in speed; the LLBA data is presented in Figures 1 and 2 along with data from the literature (2,3,5,6,8,11). When looking at previously published studies, it is obvious that the energy cost of LLBA increases as speed increases. In the present study, the energy cost of FNSW remains unchanged or decreases slightly as speed is increased from 0.13 m/sec to 0.39 m/sec (Figure 2). As the speed of LLBA approaches 0.4 m/sec, the energy cost increases and approaches that of FNS walking.

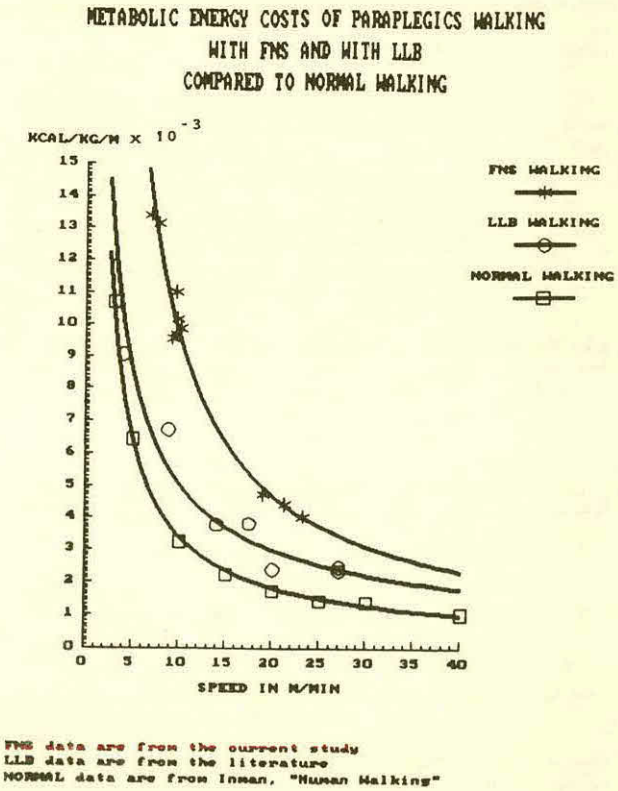

Figure 2

1. FNS uses more energy than LLB at speeds below .4 m/sec..
2. We cannot sustain FNS walking (59-76% of MAP) for long periods
3. As speed of LLB increases so does energy cost.
4. During FNS walking, as speed increases no significant variation of energy cost is seen.
5. FNS has the potential to use less energy than LLB at speeds approaching those of normal walking.
6. We can decrease FNS energy by decreasing stimulus parameters, increasing pelvic control and by providing trunk stability.

ACKNOWLEDGMENTS

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SENSORY FEEDBACK OF MACHINE STATE AND SHOULDER POSITION COMMAND INFORMATION FOR USE WITH FNS HAND ORTHOSES

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ABSTRACT

Coding algorithms are described that are used in an electrocutaneous communication system that provides cognitive feedback of machine state information and the output command signal from a shoulder position controlled FNS orthosis system. A linear array of five electrodes placed on the skin of the upper back serves to display the feedback information to the user. Specific machine state feedback information includes selection of the GRASP MODE; selection of the shoulder position that is the starting point for the command range; entry into a LOCK GRASP state; and a REALIGNMENT ERROR SIGNAL to facilitate the user in returning to the state of ACTIVE VOLUNTARY CONTROL of the grasp.

INTRODUCTION

The FNS hand orthosis systems being developed at Case Western Reserve University (1) incorporate several user selectable options that increase their versatility. Unfortunately, as more functions are added to the systems, the issues of operator control become increasingly demanding. Generally, users obtain voluntary control by means of a chest mounted push button switch which serves mainly to turn the system ON and OFF, and a two-axis proportional joystick transducer that monitors the user's shoulder movements. Protraction-retraction of the contralateral shoulder is employed by the user as a proportional control to grade the contraction of the forearm muscles and hence regulate the force of the grasp. Logical command signals are derived from rapid vertical shoulder movements. User selectable functions include a choice of two hand grasp modes; the capability to specify any arbitrary position of shoulder rotation to be the "ZERO" or start point for the shoulder controller command movements, and the ability to latch or "LOCK" the grasp at any desired level of force.

A scheme that employs auditory cues presently provides feedback for the user to verify the acceptance or execution of commands. The auditory feedback scheme has several deficiencies: The codes can be difficult for the user to learn and to remember; they are sometimes impossible to attend to in noisy environments, and in certain social settings such as dining, they can be a source of embarrassment. For these reasons, we have replaced the auditory machine state feedback with a new design that is based on electrocutaneous communications techniques. With this scheme electrical stimuli are impressed on the skin in an area of the spinal injured person's skin that retains normal sensation. The electrical stimulation of the skin elicits tapping or buzzing sensations which are coded to input information. The electrocutaneous codes to achieve this are input using an electrode display that also provides a shoulder position feedback signal to facilitate more precise command production (2) and that will eventually also support the feedback of grasp force information. A discussion of how a linear five electrode display is used for machine state and shoulder position feedback is given below.

METHODS and FEEDBACK ALGORITHMS

The electrocutaneous display consists of five elements spaced 40mm apart as illustrated in Figure 1. Each electrode has a 3mm dia. contactor surrounded by a 20mm dia. insulator. Double sided adhesive"EMG" collars provide mechanical fixation to the skin. The electrodes are monopolar and a large remote conductive pad serves as a common anode. The common anode arrangement is used because it mimics the situation that will exist when the system eventually uses an implantable stimulator (3) and subdermal implanted electrodes (4) under development in our laboratories.

Selection of the Mode of Grasp

At the beginning of the start up procedure the FNS system is in the IDLE state and there is no feedback activity. The user then depresses the chest switch and keeps it depressed to select a grasp mode. The FNS system toggles every 2 sec between the LATERAL GRASP mode and the PALMAR GRASP mode. These alternatives are signaled to the user by actuating all of the electrodes in the array with either of the following signals: Lateral grasp is indicated by presenting a series of single bursts (6 pulses; 50us pulse width; 1 kHz carrier) every 500ms. Palmer grasp is indicated by presenting three bursts in rapid succession and than repeating the treble burst pattern every 500ms. This is similar to the Morse code messages of DIT and DIT-DIT-DIT respectively. The user releases the chest switch to select whichever grasp mode was most recently displayed.

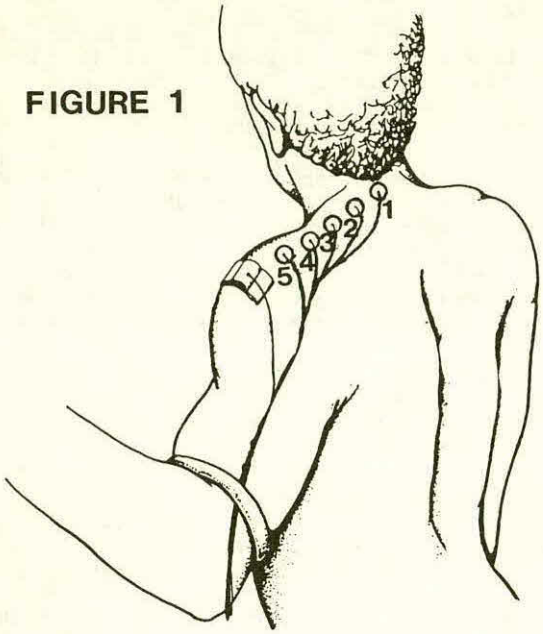
Selection of the Zero Command Position

The total protraction-retraction excursion that the user can produce is assessed in the laboratory when the user is initially fitted with an FNS system. At that time a portion (usually 40%) of the user's total range of voluntary shoulder motion is specified for the user to produce the proportional command signal that controls the grasp. This procedure eliminates the need for the user to constantly produce very large changes in shoulder position which can be fatiguing and can interfere with overall postural stability while seated in a wheel chair.

Additional flexibility afforded by the FNS system permits the user to respecify during each start-up procedure, the shoulder position (offset) that corresponds to the ZERO or starting point of the command range. The option to choose a different ZERO point is provided to compensate for slight positional changes in the fixation of the shoulder position transducer hardware each time that the system is donned and to allow the user to assume different postures and still be able to use shoulder position to control the FNS system.

As soon as the chest switch is released a timing interval of 3 sec commences. During this interval the user must move his shoulder to the position that will correspond to the ZERO or start position of the command range. The FNS system will assign as ZERO COMMAND whatever position the shoulder is in at the end of the time out period.

FIGURE 1



There is a danger that a ZERO point that is too protracted will be selected, and since the controller gain remains constant, the user would not be able to produce the upper limits of the command range when the shoulder is fully protracted. This problem which is illustrated in Figure 2 is avoided by providing the user with feedback about the absolute position of the shoulder during the timing interval for ZERO point specification as follows: Each of the five electrode positions in the feedback display, codes for 1/5 of the range of motion of the shoulder. At any given moment, one of the electrodes is active in accordance with the instantaneous position of the shoulder. The user can be assured that an adequate command range will be available by ascertaining that the ZERO point selection is made only when either the 1st (most lateral), 2nd or 3rd electrode sites are activated. A ZERO point selection when the 4th or 5th (most medial) sites are activated would imply that a command for maximum grasp would not be able to be produced.

Means are also provided that furnish the user with a confirmation of the shoulder position that the FNS system accepted as the ZERO point. This is accomplished by briefly increasing the freq. of the electrode that was active within the display at the time that the ZERO point selection time interval expired. The high freq. signal (30 Hz) then ceases, and activity is shifted to the 1st electrode position of the array (at the base freq. of 7 Hz) since this then corresponds to the start (ZERO) position of the user's command range. If the user selects an inappropriate ZERO point, then the system START UP is begun again by depressing the chest switch to turn the system off, then depressing it and holding it down to begin again with the selection of the GRASP MODE and ZERO position. Assuming that this isn't required, then the user proceeds to ACTIVE CONTROL as described below.

Active Control and Command Feedback

At this point, the user is in the state of ACTIVE CONTROL of his/her FNS assisted hand. Increased shoulder protraction increases the command signal and causes the hand grasp to begin closing. The level of command as determined by the position of the shoulder is displayed to the user by activating successive electrodes in the feedback array in a manner proportional to the output of the shoulder position transducer. Note, however, that the five electrodes of the feedback display during ACTIVE CONTROL are used to code for the command range of shoulder motion only and not the total range of motion that the user is able to produce. Thus, there is a reassignment of each electrode with the absolute position of the shoulder when the feedback display is used during selection of the position and when it is used during the ACTIVE CONTROL state.

LOCKING and UNLOCKING the Grasp

The final application of machine state feedback concerns the initiation of the "LOCK GRASP" state

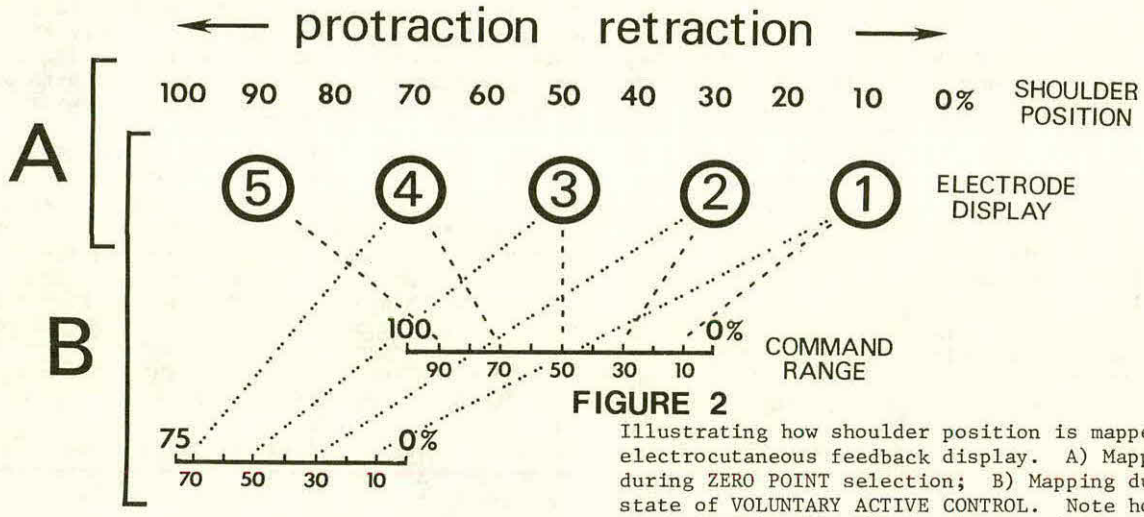


FIGURE 2

Illustrating how shoulder position is mapped onto electrocutaneous feedback display. A) Mapping during ZERO POINT selection; B) Mapping during state of VOLUNTARY ACTIVE CONTROL. Note how inappropriate ZERO POINT selection limits top end of command range in lower example.

and the user's subsequent task of "UNLOCKING" the grasp and regaining voluntary proportional control. LOCK GRASP refers to a condition during which the user can maintain the muscle stimulation parameters according to any arbitrary level of command and disengage the shoulder position controller. This is useful when an object is to be held for an extended period rather than just picked up and immediately put down, because the user can secure the grasped object and not have to devote further attention to control until it is desired to release the object.

The state of LOCK GRASP is easily achieved by the user raising his shoulder rapidly at the extent of protraction and hence, level of grasp force that is desired. After the LOCK GRASP state has been entered the feedback signal is turned off which keeps the skin from becoming excessively accommodated and serves as a message to confirm that the LOCK GRASP state has been entered.

UNLOCKING the grasp, however, can present some difficulty since the user's shoulder must be returned to the exact position that it was in when the LOCK GRASP state was entered. This is necessary to avoid an abrupt change in the level of grasp force when voluntary control is resumed: Any unexpected decrease in grasp force might allow a grasped object to slip, while a sudden increase in grasp could cause a fragile object to be crushed. Machine state feedback is provided to assist the user in finding the correct position (we call this the REALIGNMENT position) for UNLOCKING the grasp. We are evaluating the following algorithm: The user makes a rapid shoulder shrug to initiate the UNLOCK procedure. An electrode at either the lateral or the medial end of the display is activated to inform the user that the shoulder needs to be more protracted or retracted, respectively, in order to achieve REALIGNMENT. The user then slowly moves the shoulder in the direction specified by the REALIGNMENT ERROR SIGNAL.

Successful REALIGNMENT is signaled to the user by having the electrodes at both ends of the array be activated simultaneously at the 7 Hz base freq. If no adjustment is necessary because the shoulder position after the vertical "shrug" is allready at the REALIGNMENT position than both end electrodes are activated indicating no error. The user must maintain the REALIGNMENT position for at least 0.5 sec after which the UNLOCK task is automatically completed. Failure to maintain a correct REALIGNMENT position for the necessary time interval causes a return to the REALIGNMENT ERROR SIGNAL condition. Confirmation of a successful transition through UNLOCK to voluntary ACTIVE CONTROL is signaled as follows: The condition during which the two end electrodes in the array are active at the base freq. is changed to a transitional state where the two electrodes are briefly activated at a high freq. (30 Hz) for 0.5 sec. Following this, both of the end electrodes become inactive and this is replaced by activation of a single electrode at the 7 Hz base freq. in accordance with the position of the shoulder as is normally the case during the state of ACTIVE CONTROL of the grasp.

DISCUSSION

Space precludes a comprehensive discussion of the design considerations that have resulted in the present feedback algorithms, but a few comments may be provided: The separation of the feedback electrodes on the skin exceeds the minimum required for successive two-point discrimination by a factor of at least two, but our experience has shown that subjects can easily learn to rapidly identify the absolute locus of stimulation if the electrodes are 40mm apart and that this is very difficult to learn when spatial separations on the order of 20mm are tested. The "Morse code" style messages used during the selection of the GRASP MODE is more complex than would be the case if a low versus high freq. signal were used instead. However, we wish to make the feedback messages used during the FNS system start-up be substantially different from the feedback messages that are used during the REALIGNMENT tasks. This difference is further enhanced by having all five electrodes active at the same time during GRASP MODE selection, whereas only a maximum of two electrodes (the end most) are used during the REALIGNMENT and UNLOCK tasks. The later situation (i.e. having a maximum of only two electrodes active together) is also a constraint imposed by the hardware, because it can only service different electrodes sequentially. Since it is necessary for the stimulator to service the muscle electrodes during REALIGNMENT and UNLOCK there is insufficient time available for the stimulator to activate more than two sensory feedback electrodes while it is also required to maintain the muscle stimulation.

The provision of machine state and shoulder position command feedback should substantially enhance the ability of individuals with C5 and C6 spinal injury to grasp and release objects. Developmental efforts in our laboratory have demonstrated the efficacy of electrocutaneous feedback of shoulder position to assist individuals in producing more precise and more consistent command signals from the CWRU shoulder position transducer system (2). Evaluation of the feedback system in FNS system users in terms of its effects on functional tasks involving grasping have begun, and we look forward to reporting those results when they are available.

ACKNOWLEDGEMENTS
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EXTENSION-LIFT WHEELCHAIR FEATURING ACCESS TO GROUND AND FLOOR LEVEL ACTIVITIES

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ABSTRACT

This concept for a multi-terrain lifting wheelchair is a 3-wheel vehicle with a telescoping frame and lift mechanism. "Extension-Lift" raises and lowers the seated rider through a wide range of movement over a height of approximately 2 feet. In its low-level position, the Extension-Life wheelchair affords the user 1) access to floor and ground-level for activities and recovery of dropped objects, 2) stability and safety in traversing uneven terrain, and 3) a long sitting position in which the operator's legs are extended and his/her feet are elevated.

INTRODUCTION

The following analysis stems from the author's effort to develop a unique power wheelchair. In addition to its technical and functional merits, the Extension-Lift indoor-outdoor power wheelchair is presented as an example of a grassroots initiative to meet an individual's special need. This concept for a lifting wheelchair eventually became the senior project for the author's undergraduate studies in technical writing at the University of Minnesota, Duluth (UMD). In addition to writing tasks associated with filing patents for the Extension-Lift feature, project activities have comprised successive design and technical refinement for Prototype III, which is now undergoing shop trials.

BACKGROUND

This idea for a lifting wheelchair was conceived over the summer and fall of 1982 as a means to improve outdoor mobility and vertical access for individuals with physical handicaps. The Extension-Lift function evolved from the inventor's personal quest to extend his own functional limits as a post-polio quadriplegic.

Other designers have sought to develop a wheelchair that will lower its seated rider near the ground. Some innovators in this country and abroad are seeking to develop this use in new designs.[1] In the inventor's case, this design was inspired by his perceived need to get lower to the ground to play with his small children at their own level and participate in other outdoor activities that he had enjoyed prior to his disability.

A crude "backyard" prototype was first assembled in 1982. Shortly after, the concept was presented to students and officials of the Duluth Area Vocational Technical Institute, whose generous technical support over the next year produced Prototype II. This model proved the potential for Extension-Lift to bring the seated rider close to ground level. Over almost 200 hours of hard use in 1984 and 1985, it served reliably as a test chassis that was studied to determine changes for future development.

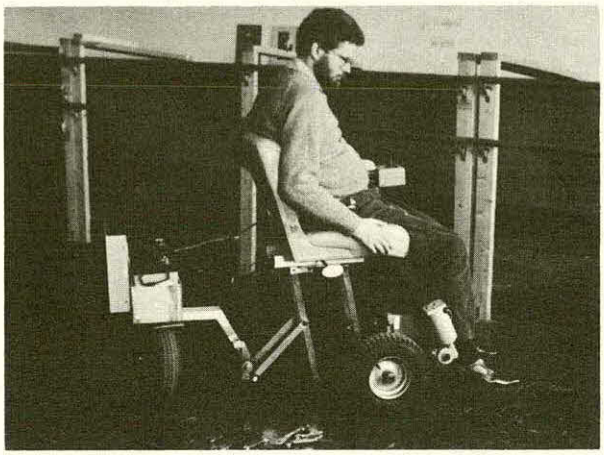

Figure 1.

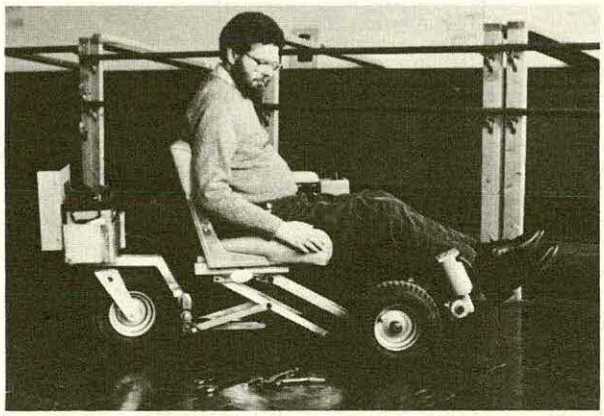

Figure 2.

Figure 3.

TECHNICAL DESCRIPTION

Configuration and components
A tripodal configuration (Fig.4) features two 12-inch front wheels pulling and one 12-inch rear wheel trailing on a castered pivot mount. The chassis comprises 1) a main frame, including an axle and spindle mounts for the two drive wheels; 2) an extension frame telescopically fastened to the main frame; 3) a parallelogram linkage that pivots to move downwards and upwards as a function of the extension movement of the frame; and 4) a rear wheel mounted in a caster on the rearmost end of the extension frame. (A swivel mount for the seat is incorporated at the top of the parallelogram linkage.)

Extension-Lift function. 1) A Motion Systems ballscrew actuator is used to extend and retract the telescoping portion of the frame 12 inches; 2) as the frame extends, a parallelogram linkage between the frame and the seat of the vehicle rotates down and backwards through an arc; 3) at 11 inches of stroke, the base of the seat touches the extension tube; whereupon 4) the extension-movement stops and the remaining inch of the actuator movement is used to kneel the back end of the vehicle the final 3½ inches to the ground (Figs. 1-3; Fig. 5).

This movement takes the rider from a normally high seated position (Fig. 1) down to a long sitting position within less than 10 inches of floor or ground level (Fig. 3). (A variety of footrest solutions are under development.)

Motors and drive train. Two 24-volt permanent magnet motors, each coupled through a wormgear and final roller chain assembly, power the front wheels. These motors are bolted to brackets welded to each of the drive wheel spindles. These motors are proportionally controlled from a joystick mounted on the armrest.

Figure 4.

Power Supply. Two 12 volt group 22 gel cell batteries from Globe, and two group 27 Absolyte (tm) 1260 batteries are being tested in the present model.

Batteries have been placed in various locations on the chassis, with the objective of leaving the most convenient access to work activities at various heights. In Prototype II, batteries were placed on the frame end to end on a tray behind the seat. This contributed too much to overall length. In Prototype III (Fig. 4), one of the batteries has been temporarily placed atop the rear wheel assembly, the other over the front axle, to facilitate testing of the new Extension-Lift mechanism. Optimal placement of batteries on Prototype III is still being evaluated (see discussion section).

Seating. Extensive seating and posturing studies have not been pursued for this vehicle. The author has tested three "off-the-shelf" utility seats from different manufacturers. These have provided adequately comfortable seating during extended hours of testing in prototpyes II and III.

Size. With the caster trailing fully behind, the overall size of the vehicle is 24 inches wide by 46 inches long in the upright position. Lowered, the vehicle is approximately 12 inches longer.

DISCUSSION

Success of the Concept
Three prototypes of this lifting wheelchair have been built and tested. Design and performance objectives have been evaluated as a qualified success.

Ground-level accessibility
Extension-Lift lowers the rider to within less than 10 inches of the floor or ground (Fig. 4). Gardening, retrieving dropped objects, and performing repairs on various equipment items are only a few of a vast range of endeavors that can be performed by many quadriplegics at this level.

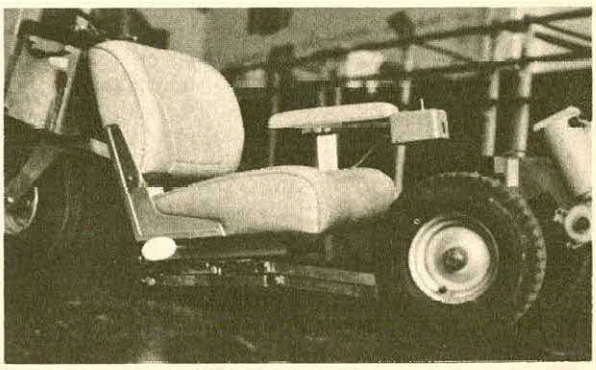

Figure 5.

Indoor and outdoor use
As a multi-terrain wheelchair, this concept's performance and features was intended to invite use outside traditional settings--off the sidewalk, in the woods, on nature trails, even through mud, sand, and snow, within reasonable limits. It has been discovered that the front wheel drive easily pulls the weight of the vehicle over difficult terrain; the steering caster follows after the weight is across. (In many conventional power chairs, the weight is thrust onto two undersized front casters, causing a lurching, jerky ride.)

The extended low-level position enhances stability, affording a wide margin of safety in traversing uneven terrain.

The tripodal arrangement also contributed to maneuverability, and compact design. The vehicle's maneuverability makes it more agile indoors than many other types of vehicles.

Transfers
Operators with nominal function in their upper body and limbs have available a wide range of height adjustment options for a level or downhill-angle transfer between the Extension-Lift wheelchair and a commode, shower chair, bath seat, living room sofa, or bed.

Design from stock components
The inventor and developers working with him have sought to construct the different prototypes using stock components wherever possible. Common implement tires, #35 roller chain, and stock motor and gear assemblies are being sought to avoid special tooling and design. This approach has also been directed toward reducing cost, assuring quality control, and speeding parts replacement.

PROBLEMS AND SOLUTIONS

Physical modeling with CAD
Early in 1985, the author gained access to a sophisticated microprocessor-based computer-aided design system. Since he began training on the system, he has started to apply this new design technology toward configuration and physical modeling problems that he has encountered during the building of Prototype III. The CAD system will be used to refine the final components placement for this model, which has been deliberately designed so that new parts can be readily retrofitted to reflect design changes.

Battery placement
The problem of battery placement on Prototype III will be solved by mounting the batteries end to end over the front axle and between the parallelogram lift arms of the vehicle.

Motors and Controls
New motors and controls are needed to enhance the vehicle's performance on gradients and uneven terrain. At this point, the author is negotiating with different manufacturers to build a programmable control unit and for design of a motor and gearbox assembly to match the performance parameters desired for the production prototype.

Appearance
CAD graphics will be used to model and refine a streamlined fairing and housing scheme to cover mechanical components that are presently exposed on Prototype III. Fabrication will be subcontracted to a plastics manufacturer.

CONCLUSION

The Extension-Lift wheelchair extends outdoor mobility and vertical access for persons with severe mobility impairment. Many of its useful features will be based on the user's sense of perceived need; new possibilities are unveiled with increased access to a world below the level to which one is limited by conventional wheelchair designs. Refinements will be made to Prototype III to further improve access and comfort. The author hopes that this initiative will stimulate efforts by others to improve vertical access for mobility-impaired individuals.

ACKNOWLEDGEMENTS

The author extends his gratitude to George Rapp, Jr., Dean of Science and Engineering at the University of Minnesota, Duluth, for his advice and support as academic advisor of the author's Interdisciplinary Studies Program during the development of Prototype III. The students and staff at Duluth Area Vocational Technical Institute are also acknowledged for their assistance in constructing the first prototype. Finally, grateful acknowledgement is expressed to the Ordean Foundation in Duluth, for contributing an initial grant to get development started.

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ELECTRIC WHEELCHAIR FUEL GAUGE TESTS

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ABSTRACT

Three electric wheelchair battery capacity gauges, or fuel gauges, were tested at the REC under typical wheelchair discharge duty cycle conditions using liquid electrolyte and gelled electrolyte deep discharge lead-acid batteries. All of the gauges gave reasonable results, and would be useful to aviod being stranded due to low battery conditions. A new circuit to improve analog voltmeter type battery monitors is described.

INTRODUCTION

The state of charge of the battery in an electric wheelchair is of vital importance to the user, especially when far from home. A few electric wheelchairs have a built in battery monitor that gives an indication of the available capacity of the battery. For those electric wheelchairs that do not have a built in battery monitor several "fuel" gauge meters which may be added to the wheelchair have recently appeared on the market. This paper reports tests on three new add-on battery meters which are appropriate for use on electric wheelchairs, and discusses some of the problems in using these battery meters.

THEORY

The most economical battery for electric wheelchairs is the deep discharge lead acid battery which is used for boats, golf carts, and other uses in which charging is intermittent [1]. The term "battery capacity" refers to the usable stored energy in terms of watt-hours (1 watt = 1 volt x 1 ampere). However, since the battery voltage during discharge does not vary much, it is usual practice to use ampere-hours as the unit of capacity for a given discharge current and temperature. Another term which is used with batteries is "depth of discharge" (DOD), defined as the percent of the total battery capacity that has been removed from the battery. For a given lead-acid battery on open circuit there are two ways, which are related, to accurately measure the state of discharge: (1) Measure the concentration of the electrolyte, which decreases linearly with discharge, and (2) measure the steady-state open circuit voltage , which also decreases linearly with discharge [2,3,4]. Both methods are only accurate after the battery is open-circuit for several days. Neither of these methods are convenient, and both methods are adversely affected by low temperatures. In the case of gelled electrolyte batteries the first method, which involves specific gravity measurement, cannot be used. A third method, which yields the approximate level of discharge, is to measure the battery voltage while operating the wheelchair on a smooth level surface. This method depends on having a relatively consistent duty cycle, and is also adversely affected by low temperatures. The duty cycle is the typical discharge history representative of wheelchair use over a several minute time period. Below room temperature, both the battery voltage and the total capacity of the battery under load are reduced, but above room temperature under typical climactic conditions the battery is little affected.

EQUIPMENT

Three battery gauges were tested at the REC using several different batteries. The two most common batteries used on electric wheelchairs, deep discharge liquid electrolyte and gelled electrolyte, were used in the tests. Batteries with capacities of 40 ampere-hours and 55 ampere-hours were used in the tests. The batteries were first discharged with a steady current drain, and then with a variable current drain. In the latter case the discharge corresponded to a typical wheelchair driving cycle. For a discussion of wheelchair driving cycles see reference [1]. All of the tests were carried out at room temperature. the three battery gauges tested are listed in Table 1, and shown in Figure 1.

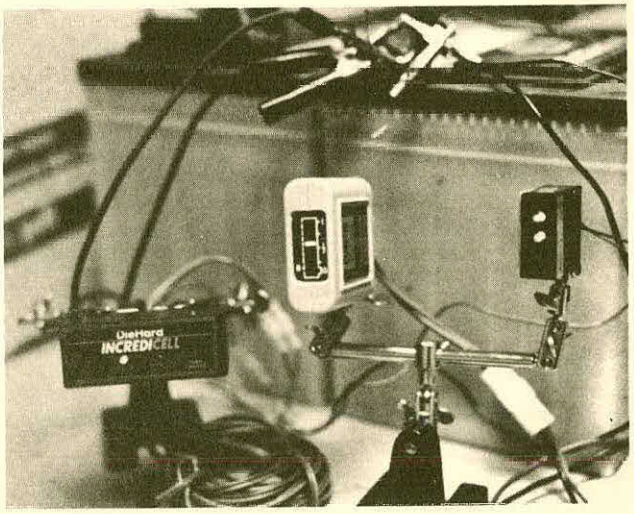


Figure 1: Electric Wheelchair Fuel Gauges L-R (1)Sears, (2)Curtis & (3)Motovator

Table 1 gives a brief statement of the measurement method used by the gauges. Gauge number one was designed for use on an automobile battery, but was found to be satisfactory for use on a wheelchair battery. It is attached across only one of the two 12 volt batteries usually used on electric wheelchairs. Gauges number two and three were both designed specifically for electric wheelchair use, and each includes instructions for wheelchair installation. For the installation of gauge one use the diagram on the last page of instructions as follows: (1) Connect the red wire to the positive terminal of one 12 volt battery, and the black wire to the negative terminal of the same battery. (2) Disregard or cut off the blue wire, and (3) Connect the grey/brown and grey wires of the temperature probe as shown in the diagram. The temperature probe wires can be run to the battery box, and the probe placed next to the batteries.

TABLE 1 : BATTERY GAUGES

1. Sears Battery Monitor 12 Volts Type: Voltmeter (temp. compensated) current consumption: 20-45 mA (15-11V)	Sears Roebuck & Co. Catalog Sales $11.99, 1985	Four lights < 11.8V undercharge 11.8-13.7 V ok and normal > 15.25 V overcharge
2. Fuel Gage - 900 W 24 volts Type: integrated circuit state of charge computer current consumption: 10-20 mA (24-22V)	Curtis Instruments 200 Kisco Avenue Mt. Kisco, NY. 10549 $75.00, 1985	Step Down LED Display 10 steps w/3 min delay max. 1 flashing bar @ 70% DOD. 2 flashing bars @ 80% DOD.
3. Range Indicator 12 volts / 24 volts Type: Voltmeter current consumption: 20-70 mA (25-24V)	Motovator 1733 Border Avenue Torrance, Cal. 90501 $32.95, 1985	Three lights Light 1, < 25.0 V 25% DOD Light 2, < 24.6 V 50% DOD Light 3, < 24.2 V 75% DOD

TEST RESULTS

A typical duty cycle discharge voltage curve for wheelchair batteries is shown in Figure 2. The battery voltage varies according to the current drain (in amperes) imposed by the duty cycle. For the three step duty cycle used in the tests, the voltage fluctuated over the range as indicate. Above the duty cycle voltage is the steady-state open circuit voltage which would be measured if the discharge according to the discharge cycle were to be stopped, and the open circuit voltage measured after at least a 24 hour wait.

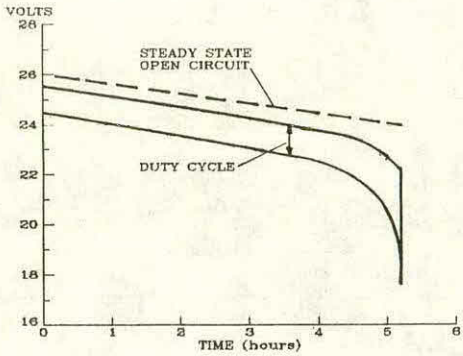


Figure 2
Typical Wheelchair Battery Discharge

All three of the battery monitors are designed to be used while the wheelchair is operating under normal loads. To use gauges number one and three properly, the user must realize that the battery voltage changes according to the instantaneous loading on the wheelchair. Therefore the user must only use the reading from the gauge when the wheelchair is under normal load on a smooth level surface. Gauge number two has the great advantage of indicating the state of discharge by a lighted bar position that does not change during a variable load cycle of less than three minutes, or during no load conditions. A typical plot of the performance of the three battery gauges during discharge of a 24 volt battery under wheelchair duty cycle conditions is shown in Figure three.

The Sears gauge has four lights with only one light indicating undercharge. When the undercharge light stays lit during normal load on a level surface, the depth of discharge has reached approximately 75%, and the battery should be recharged as soon as possible. Under heavy load the undercharge light may come on, but this is ignored when using this gauge for battery capacity.

The Curtis fuel gauge has 10 bar lights reading down to 80% DOD. The test results plotted in Fig. 3 are quite reasonable. On occasion this gauge read incorrectly at the beginning of discharge, but gave good results from 25% DOD on.

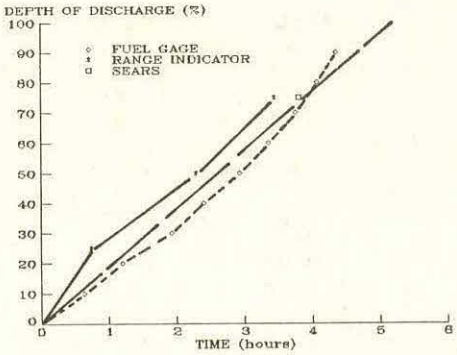


Figure 3
Typical Battery Fuel Gauge Performance

The Range Indicator gave reasonable results with proper interpretation. There are three lights set for 25%, 50%, and 75% DOD, all three lights can come on during heavy load. Only when one or more of the lights stay on during normal load on a smooth level surface is the reading correct. The plotted results for several tests were similar, except sometimes the 1st light would come on before 25% DOD was reached. The 2nd and 3rd lights gave good results for all of the tests.

RECOMMENDATIONS

A new fuel gauge circuit using analog voltage was developed that eliminates the problems of flickering lights due to loading. The problem with analog voltage measurement is that a heavy load can cause the battery terminal voltage to momentarily dip below the threshold voltage which would indicate to the monitor a lower % DOD than has already been reached, turning on the LED. Since the wheelchair duty cycle repeats itself after approximately one minute, a circuit which prevents the monitor from responding to a voltage decrease for more than a minute will alleviate this problem. The circuit shown in Figures 4 and 5 will not respond to a decrease in voltage for 90 seconds. The actual delay mechanism is a 22 microfarad capacitor in series with a 3.3 megaohm resistor in the LED driver. This capacitor is placed across the base-emitter junction of an NPN transistor (Q2) and four diodes. This means that the transistor will turn on and light the LED when the capacitor charges up to 3 volts. The charging is controlled by placing a transistor (Q3) with its emitter and collector connected across the capacitor. This transistor (Q3) is saturated when the battery terminal is above the predetermined threshold for X% DOD. This keeps the voltage across the capacitor at 0.02 V, the collector-emitter saturation voltage. When the terminal voltage falls below the threshold level, the transistor (Q3) turns off, allowing the capacitor to charge to 3 volts. This takes approximately 90 seconds. The trigger circuit is included to illustrate the entire monitor circuit. It is a simple comparator network based on an inexpensive IC that triggers three separate LED drivers as the battery terminal voltage falls past three threshold levels as it is discharged.

FIG. 4
BATTERY MONITOR TRIGGER CIRCUIT

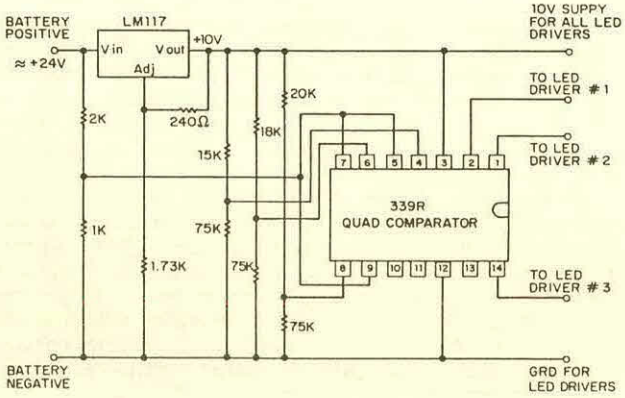
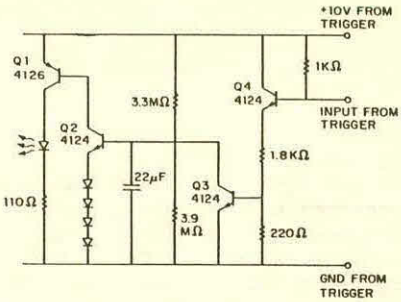


FIG. 5
LED DRIVER CIRCUIT W/DELAY



CONCLUSIONS

Three battery gauges for electric wheelchairs were tested and found to be reasonably accurate. Any of these gauges should be very useful in giving the wheelchair user an indication of the state of his batteries. They can be used in the same manner as one would use an automobile fuel gauge - to prevent strandings.

The Curtis Fuel Gauge was the most sophisticated and most expensive, but the easiest to use. The other two gauges require some interpretations in order for them to be used correctly, but a little experience should prove to be sufficient for correct usage.

The Sears Battery Monitor has automatic temperature compensation. The Curtis and Motorvator gauges will correct for low temperature since both battery voltage and capacity under load decrease with temperature, but this effect was not tested.

The design for a new fuel gauge with a delay circuit which will reduce the problem of using an analog voltage gauge is presented.

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COMPARATIVE EVALUATION OF TWO SWITCHING CONVERTERS FOR EFFICIENT DC MOTOR CONTROL

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ABSTRACT

Battery life is a very important consideration for powered prostheses & wheel-chairs, and can be extended by efficient motor control. The near ideal characterstics of MOSFETs are expected to yield improved motor and contoller efficiencies. Two configurations of MOSFET switching amplifiers are evalulated in this paper. Simple efficiency measurement experiments show that 80 % controller is achievable if parasitic affects are carefully minimized. Comparative advantanges & disadvantages as well as the suitabilty of each in rehabilitation applications, is discussed from considerations of size, weight, and the range of controller gain.

INTRODUCTION

Among other things, electric wheel-chairs & powered prosthetic devices can greatly benefit from an increased battery life. With the advances in low-power electronic components and with the better understanding of electrochemistry of batteries, etc, significant improvements in the battery life are now possible. For instance, with the use of lithium batteries and CMOS electronic devices, the battery life for digital watches and calculators exceeds one year. This is not the case with other systems that have moving parts. These systems require much more power and consequently would require "super batteries" of extrmemly high energy density, which neither exist presently, nor are expected in the near future. Only careful design, minimizing the power loss and possibly utilizing regenrative braking, can make enegetic portable systems acceptable for wide use. The objective of this study is to evaluate the suitability of DC motor controllers based on the near ideal characterstics of power MOSFETS, for wheel-chair and prostheses applications. General discussion why MOSFETs are well suited for such motor control applications can be found in [1].

BACKGROUND

Principle of Operation of Switching Contoller

To grasp the principle of operation of switching motor contollers, consider motor speed control by varying the motor voltage. One way to achieve this from (say) a battery, is to use a variable series resistance. Ofcourse this will be very wasteful since the series resistance will dissipate a large fraction of the total power. However, it can be noted that in two extreme cases, this power dissipation will be negligible, e.g., when the series resistance is zero or infinite. In the first case, the battery is directly connected, applying the full voltage to the motor, and in the latter case, the battery is disconnected and the applied motor voltage is zero. The switching controller rapidly alternates between these two extremes to achieve any average voltage.

Sources of Power Loss

The main sources of power loss for a switched-mode controlled motor are: resistive loss (copper loss in inductors, loss due to battery internal resistance, on-resistance of the MOSFET, etc.), core loss (hysteresis and eddie current losses in inductors) and switching loss. The selection of the proper switching frequency can minimize these losses. Let us look at it more closely. Because of the switching between two extreme cases in a switching controller, the motor not only sees an average (or DC value) of the current, but also a AC component. In general this AC component adds to the resistive loss in the motor without contributing to the torque and hence reduces the efficiency The AC component can be reduced by increasing the inductance in the circuit or by increasing the switchinag frequency. The inductor core losses increase with the switching frequency. The losses in MOSFET occurs only during the switching process when the resistance goes from a very high value to a very low value. Consider for example a typical MOSFET with a switching time of about 300 nanosecond. The switching loss is proportional to the switching frequency, the duration of switching, and the average power. Thus the percent loss at 20 Khz frequency is: 100(Tswitch)(fswitch) =0.6 %. The switching loss can be easily neglected unless the frequency of switching is in MHz. Thus switching frequency can be chosen from the consideration of inductor core loss and motor form factor loss.

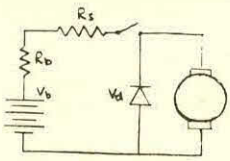
DISCUSSION

Many variations in the basic switching controller configurations are possible. The efficiency and the range of output voltage will differ from one configuration to another. Two of the more promising configurations are considered in this paper for evaluation, viz, noninverting buck converter & inverting boost-buck converter (refer to [2] for the terminology).

The two switching controllers under investigation are analysed using state space averaging technique. This is necessary due to the nonlinear switching process. In this technique the state space equations are separately written for the two switch positions and multiplied by the fraction of the time the switch is in the particular position. These weighted equations are then added to yield an average state equation, which accurately describes the phenomena with time constants much greater than the switching period. The analytical expressions of efficiency so obtained were compared with experimental values. The experiments consisted of measuring the input and output power by measurements of the motor voltage and current, for different values of pulse width (duty cycle) and at different power levels. The results for each controller are discussed separately first and then compared with each other.

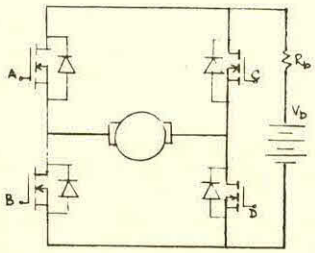
Non-Inverting Buck Converter

This is the simplest implementation of the switching control for a DC motor (see Fig 1). When the switch is on, the motor is connected to the power source, and when the switch is open, a diode (which can be an integral part of the MOSFET) is used to provide a path for motor armature current. Thus if D is the ratio of the time the switch is on, to the time it is off, the average voltage seen by the motor is given by, Vm = D (Vb).



Non Inverting Buck Configuration

Figure 1.



"H" Bridge, 4 Quadrant Configuration

	A	B	C	D
MOT +	⊓⊔	OFF	OFF	ON
MOT −	OFF	ON	⊓⊔	OFF
GEN +	OFF	⊓⊔	OFF	ON
GEN −	OFF	ON	OFF	⊓⊔

Figure 2.

This converter can thus only achieve motor voltage smaller than the battery voltage, hence the name "buck converter". This basic idea can be used to build a four-quadrant converter as shown in Fig. 2. A quadrant can be chosen by modulating the correct MOSFET. Note that the generator quadrant greater than unity gain is possible and can be referred to as a boost converter. If Rm, Rs, Rb are the resistance of the motor, the on-resistance of the MOSFET, and the internal resistance of the battery, respectively, Vb, Vd are the battery voltage and diode drop, and Im is the motor armature current, then the efficiency of the buck converter can be written as,

$$\eta = 1 - \frac{(1-D)Vd}{D\ Vb} - \frac{\{Rb+Rs(1+D)\}Im}{D\ Vb}$$

The experimental measured efficiency of a DC motor is plotted in Fig. 3 for free running and stalled condition. The corresponding theoretical predictions based on the above relation is also shown. The efficiency under the stall condition is smaller since the current is much larger and the second

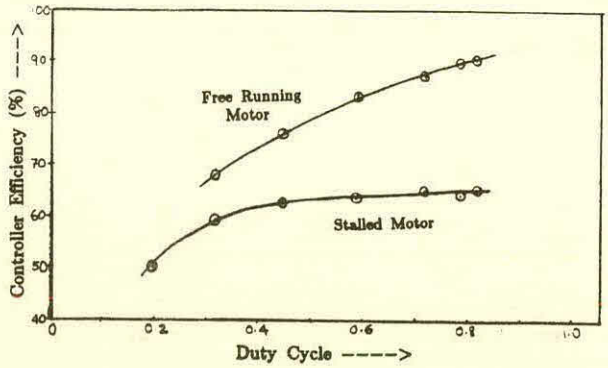


Figure 3.

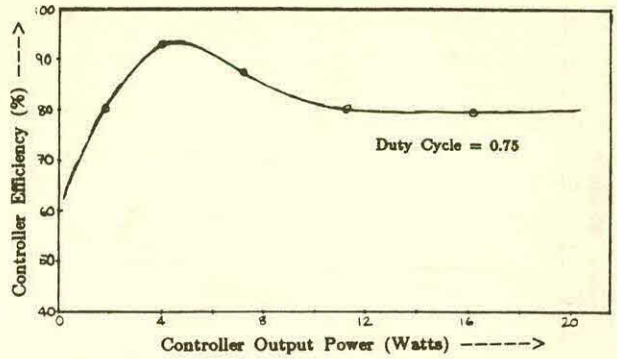
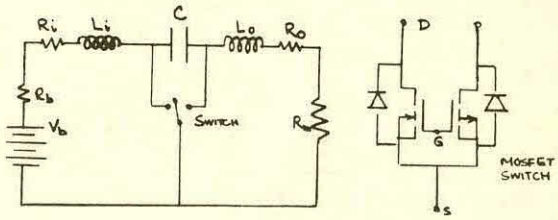


Figure 4.

dissipation term in the above expression becomes significant. Fig. 4 shows the converter efficiency plotted against output power. At very low power output, D is very small and therefore the first dissipation term is large (voltage drop in the diode is comparable to the motor voltage). For large power outputs, motor current is high and for constant battery the efficiency goes down linearly with power. Thus there is good agreement between the theoretical prediction and the experimental results.

inverting Boost-Buck Converter

As the name suggests, in this configuration, the polarity of the output voltage is inverted. Also this converter can both step-up (amplify) & step-down (attenuate) the input voltage. The two pole double throw switch shown in the schematic (Fig. 5) is implemented by a pair of N & P-channel MOSFETs. When the switch is in position 1, the input inductor is shorted accross the battery.



Inverting Boost-Buck Configuration

Figure 5.

The inductor current builds up until the switch changes to the second position, then the current decays as the intermediate capacitor is charged. When the switch moves to position 1 again, the intermediate capacitor is discharged to the load through the output inductor. In the absence of parasitic affects in the inductors and switches, the gain of the amplifier is given by the ratio of the time the switch is in the first position, to the time it is in the second position. In real cases, when the inductors have resistance, the behavior of the converter significantly differs from the ideal case as shown in Fig. 6. Whereas the ideal gain approaches infinity as D approaches 1, the real gain approaches zero. This difference is due to large resistive loss when current levels are momentarily very high. The real gain and efficiency, in the presence of parasitic inductor resistance is given by,

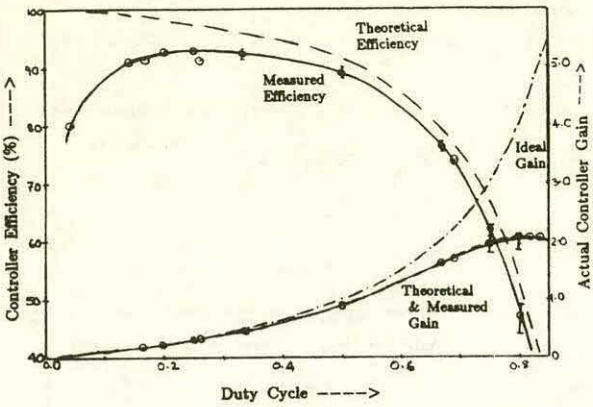


Figure 6.

$$\eta = \frac{1}{1 + (Rl1\,\rho^2 + Rl2)/Rm}$$

$$\frac{Vo}{Vin} = \rho\eta \qquad \text{where, } \rho = \frac{D}{(1-D)}$$

Fig. 7 shows the comparision of theoretical and experimental result. The two solid curves show the drastic effect of inductor resistance on the converter efficiency and on the range of gains. The lower curve is the characterstics with lower resistance but at higher switching frequency of 23.5 KHz. This clearly shows that the core losses become very important at high switching frequency.

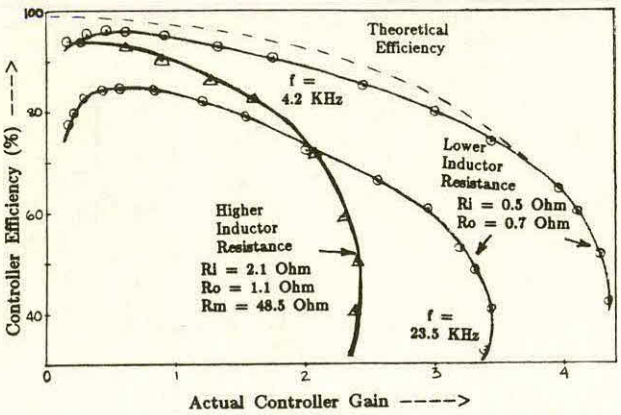


Figure 7.

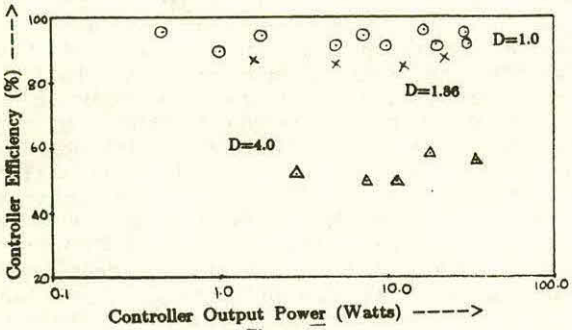


Figure 8.

Therefore if higher frequencies are to be used, which is desirable, (since both the inductor size and audio-frequency noise decrease with increasing frequency), careful inductor design is essential. These core losses account the main difference between theoretical and experimental results. Fig. 8 shows that in this configuration, the efficiency of the converter is independent of the power.

CONCLUSIONS

The experimental results are consistent with the theoretical prediction of the controller. The peak efficiency of the two controllers is approximately equal.

The output ripple can be reduced by increasing the switching frequency in the ultrasound range, thereby, moving the motor performance up to the level of purely DC operation. Such high switching frequency is made feasible using MOSFETs.

Buck converter can be used with advantage where weight and size is premium, since it does not require any inductive element other than the motor armature inductance itself. It is thus well suited for prosthesis applications. The limited range of controller gains, however, make it necessary to use a higher voltage battery. It is also suitable for 4-quadrant applications, where both motor directions and regenerative braking may be used.

The inverting boost-buck converter, on the other hand, has the advantage of wider gain. This range, however, is very sensitive to the of parasitic effects. The inductors, therefore, have to be carefully designed, minimizing the core and copper loss. This is likely to result in big and heavy components and may be unsuitable for devices that have to be carried, but may be quite acceptable for wheel-chairs.

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DESIGN CONSIDERATIONS FOR AN ELECTRIC VEHICLE AUTOMATIC TRANSMISSION

James B. Reswick

ABSTRACT

It is shown that a torque responsive automatic transmission that increases its speed ratio with increasing load can markedly increase overall efficiency of an electric motor driven vehicle operating under high load at low speed when compared with a vehicle having a fixed speed ratio drive. A new variable speed ratio transmission called the "RESATRAN" is discussed from the point of view of the optimization of its design parameters. Theoretical operating characteristics of a RESATRAN equiped vehicle are presented.

BACKGROUND

Because of their light weight and high efficiency, permanent magnet (PM) direct current (DC) motors are used to power many small vehicles including, especially, those designed for use by disabled persons.

The ideal torque, current and efficiency vs. speed characteristics of the PM-DC motor (which are the same as for the DC shunt motor) are simple and well known as shown in Figure 1.

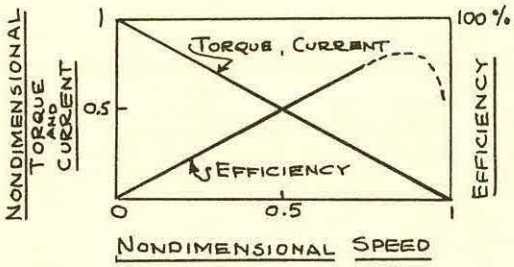

Figure 1. Torque, Current and Efficiency vs. Speed for a PM-DC Motor at Constant Voltage

While a relatively high efficiency of over 80 percent may be obtained, it occurs only near the maximum motor speed where torque values are low. Vehicles for handicapped persons are usually designed to operate at their highest speed around 80 percent of maximum motor speed. They are, therefore, relatively efficient on hard level surfaces at full speed with overall efficiencies of nearly 50% reported (Ref. 1). Almost all of these electric powered vehicles employ a direct drive using belts and/or gears between the motor and the wheels. Thus, the overall efficiency at low speeds on the level and especially on slopes is very low.

A variable speed torque responsive automatic transmission has been proposed by the author as a means to improve the efficiency of PM-DC motor driven light vehicles. A prototype, called the "RESATRAN", has been built and tested by him in a three wheeled vehicle. Test results showed an increase of about 100% in efficiency when climbing grades and an average increase of 60% in distance traveled over a certain test course for a given battery charge (Ref. 2).

The purpose of this paper is to further elucidate certain design factors related to automatic torque responsive transmissions of the RESATRAN type.

THE AUTOMATIC TRANSMISSION

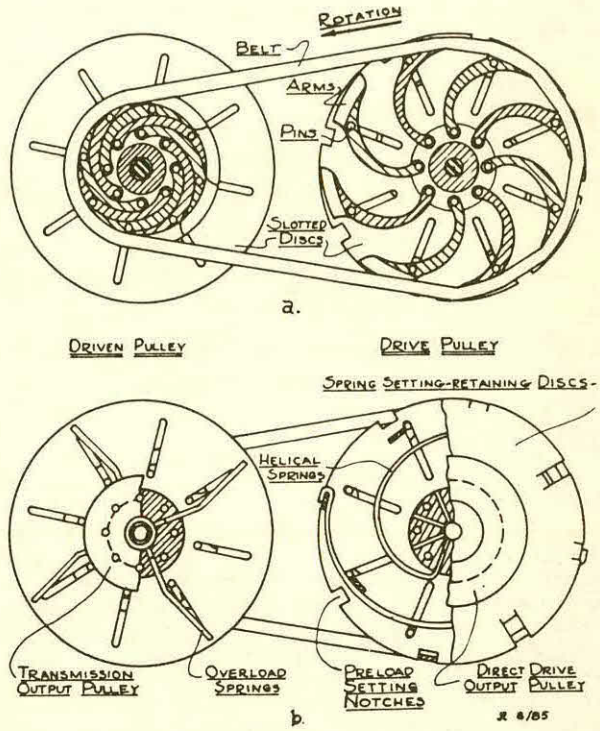

Figure 2. RESATRAN Automatic Variable Speed Transmission. a. Section. b. Side View.

Figure 2 is a drawing of the RESATRAN with different sections cut away to show its principal parts. The reader is referred to reference 2 for details of its operation. For the purpose of this paper, it may be understood as a transmission that maintains a constant output to input speed ratio of 1.73:1 for an input torque between zero and "A", where A is a certain percent of the maximum input torque. When the input torque exceeds A, the transmission speed-ratio becomes variable and is inversely proportional to input torque reaching a value of about 0.58:1 at an input torque of "B" where B is a larger (than A) percent of the maximum input torque. Speed-ratio is constant at about 0.58:1 for all torques higher than B. Overall speed-ratio changes by a factor of 3.

A and B are the main design parameters to be discussed. For the RESATRAN as pictured in Figure 2, A is the initial bias torque produced by the springs holding the arms in their outmost position and B is the torque exerted by the springs when the arms just become nested in their minimum diameter position.

In the following, the drive pully will be assumed to be attached to the drive motor so that the maximum reference torque is that produced by the motor when stalled with full voltage applied. In equation form, the above becomes:

$R=1.73$ for $0<T_M<A$

$R=1/(B-A)x(1.73B-.58A-1.15T_M)$ for $A<T_M<B$

$R=0.58$ for $B<T_M<T_{MAX}$

where:

R=output speed - motor speed

T_M=motor torque

A=motor torque at which transmission begins automatic change

B=motor torque at which transmission completes automatic change

T_{MAX}=maximum motor torque at full voltage and zero speed.

One may ask, "why get so complicated - why not have the transmission change its speed ration over the entire motor speed range?" A glance at figure 1 shows that there is no need for speed reduction at low motor torques where efficiency is high. In this region it is best to use the motor's characteristics without alteration. Thus, value A should be not less than about 20% of maximum motor torque. At high torque demand it is desirable to have the maximum speed reduction available over a range of speeds since it is undesirable to operate a motor near zero speed. Thus B should be not greater than 50% or 60% of maximum motor torque.

A second question may then be asked, "Why not use two sets of gears and simply shift between them at some middle range torque to achieve the same objective?" Such a system is quite feasible and is, in fact, used in some electric powered prosthetic hands, but the transition from low motor speed to high would be rapid and disconcerting to a vehicle user. From a smooth control point of view, it is much better to spread the speed-ratio change over a substantial torque range. Experience with the RESATRAN has shown that it tends to "smooth out" rapid acceleration changes.

Thus, in the case of the RESATRAN, the spring design depends on parameters that will place A above 20% and B at below 50% or 60% of maximum motor torque.

THEORETICAL OPERATING CHARACTERISTICS

Most motorized electric vehicles are controlled by means of pulse modulation of current. The motor characteristics at a fixed pulse modulation value are the same as if the voltage of an unmodulated current were kept constant. Thus the effect of pulse modulated control may be understood through examination of constant voltage characteristics plotted for different voltages.

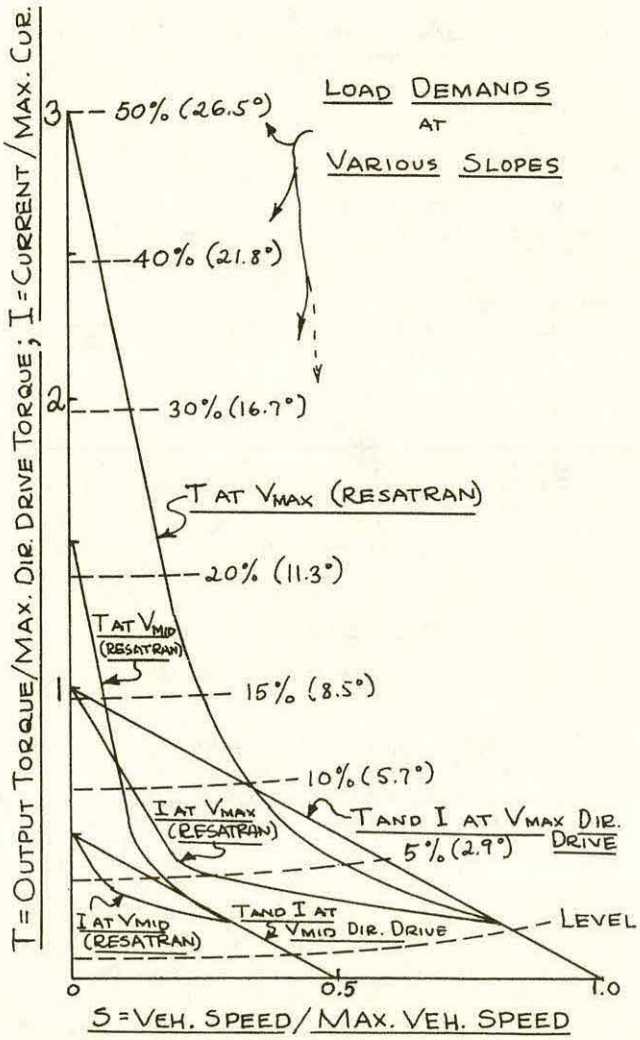

Figure 3. Theoretical Operating Characteristics of an Electric Powered Vehicle with and without RESATRAN Automatic Transmission. Load Characteristics are for Low Speed and Based on an Appropriate Motor Choice. A=0.2 and B=0.4 for V_{MAX} and A=0.4 and B=0.8 for V_{MID}. $V_{MAX}=2V_{MID}$.

To gain an idea of the operation of a vehicle equipped with a RESATRAN automatic transmission, operating curves have been drawn for two constant voltages, V_{MAX} and V_{MID} where $V_{MAX}=2V_{MID}$. Based on V_{MAX}, A has been set at $0.2T_{MAX}$ and B is put at 0.4 T_{MAX} where T_{MAX} is the maximum motor torque at V_{MAX} volts. Choice of these values and the curves themselves are the result of a computer study. Because A and B are different for V_{MID} the curve for V_{MID} is very different in shape than that for V_{MAX}. The fact that the curves are all nonlinear and do change their shape with different applied voltages leads to interesting and perhaps surprising over-all operating characteristics. The curves assume 100% transmission efficiency and "ideal" motor characteristics.

Figure 3 shows the ideal characteristics for the two voltages V_{MAX} and V_{MID}. The straight lines represent both the current and torque characteristics for direct drive without the RESATRAN transmission. All values are non-dimensional and related to the maximum output torque with direct drive, maximum current and maximum vehicle speed at the higher V voltage. On the same graph are plotted typical load torque requirements for travel on the level and up grades of various steepnesses. The motor is "chosen" so that it's maximum stalled torque would equal that required to hold a vehicle with a RESATRAN on a 50% (26.5^{o}) slope. Speeds are low so that windage losses are negligible. A number of interesting conclusions may be drawn from figure 3.

Operation on a level surface at constant speeds is shown on the "level surface" line. The transmission is inoperative and currents and torques are the same as for a direct drive.

On a 5% grade at V_{MAX}, the vehicle will climb somewhat slower than with a direct drive but with about 50% the current. At V_{MID} it will climb faster than with the direct drive and with about 60% current.

On a 10% grade at V_{MAX}, the vehicle will climb slightly faster than with a direct drive and draw 53% current. At V_{MID} the vehicle could not climb a 10% grade with direct drive but can easily do so with the transmission, drawing only a little less current than at V_{MAX}.

On a 15% grade the improvement is dramatic. With direct drive V_{MAX} the vehicle would bearly move at near zero efficiency. With the transmission it reaches nearly 25% speed at 40% of the direct drive current.

Over 15% grade at V_{MAX} a maximum starting torque equivalent to that required to hold on a 50% grade can be achieved. While such grades are rare (and dangerous) this high torque is useful for curb climbing and starting up over obstacles.

CONCLUSION

With the RESATRAN, or similar automatic transmission, an electric powered vehicle may be equipped with smaller motors than with a direct drive and be expected to operate more efficiently and thus go farther on a single battery charge. Design parameters for optimum performance have been derived and performance curves for these parameters are presented.

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22.5

A MULTI-PURPOSE SIMULATOR USED IN POWERED MOBILITY ASSESSMENT & TRAINING

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INTRODUCTION

Due to extremely limited strength and range of motion, many persons with high-level quadriplegia may appear to be unable to operate a conventional powered wheelchair. These physical limitations have led to the research and development of many sophisticated alternatives in powered wheelchair actuation and control interfaces, i.e., sip & puff, head actuation, chin control, voice command, etc.

Little attention has been paid toward fostering the physical restoration process by encouraging utilization of weak or limited muscles to control the chair. Many attempted resolutions opt to utilize a more immediate or convenient form of wheelchair actuation, irregardless of its contribution to atrophy, and the subsequent reduction in overall ADL abilities. With the proliferation of technology in the rehabilitation process, it is imperative that we do not jeopardize a patient's potential to regain multiple functions, by our well-intentioned efforts to remedy the single problem of wheeled mobility.

BACKGROUND

Patients with two distinct clinical pictures have benefitted from assessment and training with the multipurpose simulator. The first is associated with spinal cord injuries at the C-4 level resulting in innervation of facial, levator scapulae, trapezius, upper cervical paraspinals, and sternocleidomastoid muscles. This patient may also exhibit trace strength of the C-5 myotome, specifically deltoids and biceps. Typically, the maximum level of wheelchair mobility attained by this patient is with utilization of sip and puff, mouth stick, or chin control device. An alternative to this scenario has been a therapeutic emphasis on hypertrophy of existing musculature, muscle reeducation, and the laying of ground work for possible nerve root return through power wheelchair propulsion training with a hand control.

Due to the paucity of muscle activity available for achievement of this act, the set-up for successful activation of a joy stick control must maximize and optimize utilization of remaining musculature. Thus, the training position with the client has been in an upright position in a power wheelchair. The arm may be supported in a functional driving posture (See Figure 1) by a sling-suspension with supporting cuffs under the proximal and distal forearm, an elbow splint (if required), wrist splint, and joystick interface (knob, T-bar, etc.), initially strapped to the wrist splint.

The position of maximal support allows the direct translation of all shoulder and neck muscle activity into the joystick control. Weight-shift and body mechanics are also recruited to learn to

FIG. 1 Patient driving with arm support

drive. Within the first day of training, a client may begin to learn the fundamentals of driving. As progress continues with increased strength and control, external support mechanisms may, and will be weaned to the point where one is driving independently with only a joystick interface upon the joystick. Conventional methods of strengthening are an adjunct to the driving training, yet the functional position of the upper extremity in training for power wheelchair mobility facilitates learning the effective use of available musculature and body mechanics for successful driving.

The training method, as described, is dependent upon appropriate staffing and power wheelchair availability. Both of the elements of safety and feedback may be limited at times in the clinic. Such a situation fostered the development of the multi-purpose simulator as an adjunct to power wheelchair mobility training with a hand control. (See Figure 2)

FIG. 2 Patient learning to use joystick

The second clinical picture of a patient benefitting from use of the simulator is also associated with spinal cord injuries. Individuals with C5-C6 lesions of the spinal cord have innervation of biceps and deltoid musculature (C5), and wrist extensors (C6), and may exhibit a complicating condition of high tone or spasticity. Such tone is typically dominant in the flexor muscles of the biceps. Contractures at the elbow

FIG. 3 Alternative positioning of the joystick

joint may result, rendering typical posturing of the arm for functional hand control power wheelchair control an impossibility. Alternative positioning of the joystick control box for successful achievement of driving then becomes necessary. (See Figure 3)

METHOD

Joystick Alteration & The Development Of An Adjustable Mounting Stand

A standard E&J 3P joystick drive control box was modified by installing a ball and socket swivel joint between the drive control box and the extension slide tube. This allowed for up to 30 degrees angulation of the box in any desired direction.(See Figure 4) Gravitational goniometers (mounted on the front and side of the control box) are used for determining position references, when duplication of this position is required on the

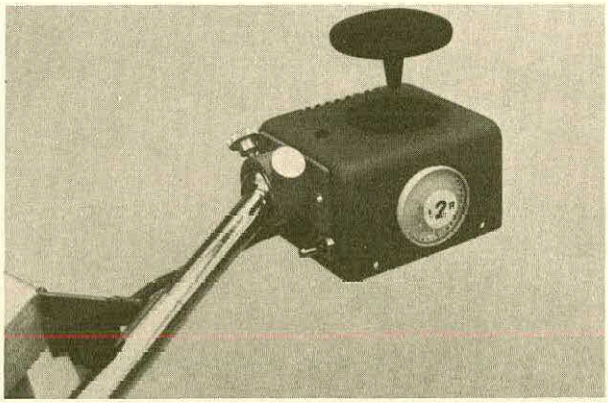

FIG. 4 Modified joystick control

client's personal wheelchair. The extension slide mounting bracket was mounted on a weighted pedestal to allow for up-down/in-out movement of the control box and extension slide tube. (See Figure 5) The joystick mounted on this adjustable

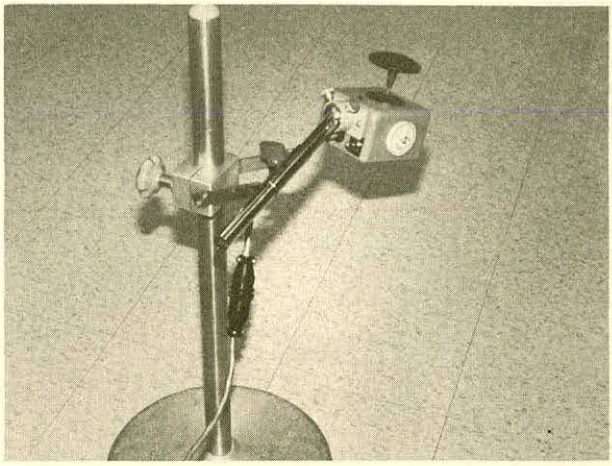

FIG. 5 Adjustable mounting pedestal

stand becomes portable, and the therapist is afforded the option of placing it to either side of the patient's wheelchair to begin determining the position which facilitates the most controlling capability. This eliminates the need to transfer the patient into different powered wheelchairs for the purpose of evaluation, and also affords immediate "trial and error" positioning possibilities.

The angling capability of the drive control box allows virtually limitless attitudes of the joystick. For instance, while the patient is endeavoring to maneuver a conventional joystick in a straight-forward direction, it may be determined that he is repeatedly directing the joystick into the left quadrant. In this case, the drive control box could be angled to the left in order to provide straight-forward mobility, while pushing the joystick in this same direction. In a similar situation, the patient may lack the strength to maintain a straight-forward direction. This can be remedied by tilting the drive control box downward to utilize gravity assistance in this direction. (See Figure 6)

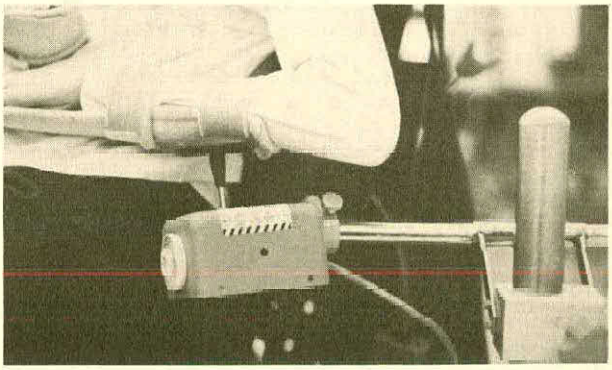

FIG. 6 Tilting box for gravity assist

Development Of A Visual Feedback Interface

A Visual Corresponding Coordinate Display (VCCD) unit was developed which provides four opposing LED bar graph displays to simulate direction of travel. The bar graphs are proportionally illuminated by the directional displacement of the joystick. (See Figure 7) The self-contained power

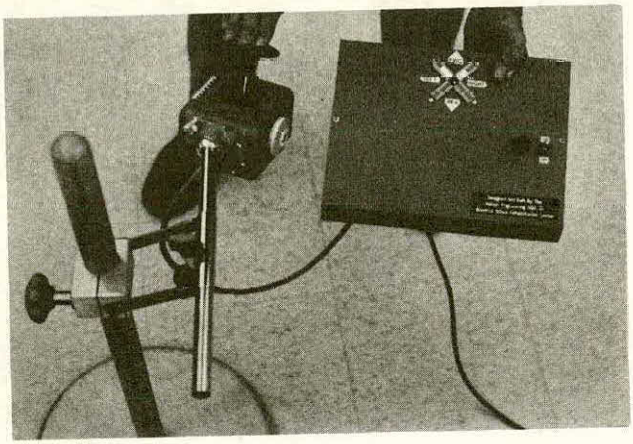

FIG. 7 Joystick simulator demonstration

supply and related circuitry, housed internally in the VCCD unit, provide variable voltages which feed the four IC display drivers, causing the LED bar graphs to segmentally light in step with the tilt of the joystick. The drive control box is coupled to the VCCD by the same type of connector as used on the 3P wheelchair. This allows the clinician the ability to disconnect the control box from the VCCD, remove it from it's stand, and connect it directly to the wheelchair for a real test-drive situation, once proper location is determined from the simulator mode.(See Figure 1)

RESULTS

A partial list of benefits derived from the development and utilization of this project are as follows:

To The Client

(1) Patients transferred to this type of wheelchair manipulation have demonstrated improved self-image.
(2) There has been noted hypertrophy of existing muscles, thus facilitating gain in other ADL activities, i.e., feeding, typing, etc.
(3) Barring spasticity, most users have gained higher levels of functional independence.
(4) Reinforcement of a more natural sitting posture.
(5) Much safer for early stages of training.

To The Clinic

(1) Less staff involved to monitor training.
(2) The visual display provides more easily quantifiable data.
(3) The articulating control box allows virtually unlimited variations in positioning, thus quicker set-ups.
(4) Prior necessity to have a powered wheelchair for evaluation purposes is reduced.

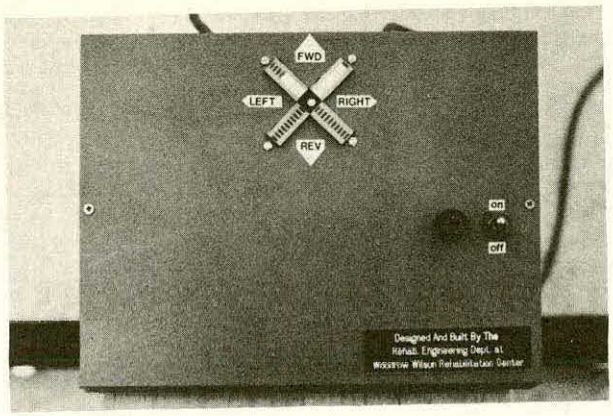

FIG. 8 VCCD Unit

CONCLUSION

VCCD units could be inexpensively produced and sold by all powered wheelchair manufacturers. In addition to assisting as clinical assessment and training devices, VCCD's would aid in proper wheelchair prescription. It is further suggested to all manufacturers that the articulating feature of our modified joystick control box, and the stacking (in-line) power switch, be offered as optional features on all powered wheelchairs of this similitude. The stacking power switch (of the click-on/click-off variety) could be remotely positioned by the clinician for head or elbow activation, thus affording total independence of the wheelchair user.

ACKNOWLEDGEMENT

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MANAGING COMMUNITY MOBILITY IN LONG TERM BRAIN INJURY REHABILITATION

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Lenox Hill Nursing and Rehabilitative Care Facility's Community Team Program provides comprehensive physical, cognitive, and psycho-social rehabilitative services to brain-injured adults in order to prepare them to achieve their maximum independence in alternative living environments and/or vocational-educational settings. With functional independence as our goal, it is important to provide the brain-injured population the mobility skills necessary to function as an actively involved in the mobility group to:

1. increase the residents' awareness of community facilities,

2. increase the residents' awareness of the accessibility rights,

3. facilitate independence of mobility-impaired residents within the community, and

4. facilitate advocacy in issues pertaining to community accessibility.

The group meets weekly to review accessibility rights and laws and plan actual public visits pertinent to community living. Staff accompany residents into the community to assess each resident's knowledge base, advocacy skills, and ability as related to community mobility. In turn, the staff can report problems and progress to interdisciplinary team for the proper remediation through inpatient treatment and proper discharge planning A poster presentation reviewing the group's purpose, goals and treatment sessions will be provided.

Brain Injury is a diffuse diagnosis lending the individual to variable levels of function if he or she is able to withstand this severe form of insult. The human brain is a vast and individualized organ making signs and symptoms of injury unique to the patient as well as to the location and extent of brain damage. Overall areas of dysfunction seen are behavior, cognition, communication, and sensorimotor function. Physical therapists, as members of the multidisciplinary team, must work to effectively retain and educate the patient to achieve their optimal level of functioning.

Many clients become significantly impaired in their ability to move so that adaptive equipment, such as wheelchairs, is used to aid in their performance. The physical therapy departments of many hospitals and rehabilitation centers serving brain-injured clients are held responsible for addressing clients' mobility skills within the facilities, and upon discharge, within the community. In an effort to ensure independence, the community mobility group was organized.

Community team mobility group was started to fulfill the needs of a handicapped population that is actively pursuing discharge into the community. To instruct residents to maneuver and manage themselves within the community as well as advocate for their rights as handicapped individuals is the group's main purpose.

To be a member of the mobility group, each resident must meet the following criteria: have active interest in pursuing community discharge; have an impaired mobility status; attend the group 100% of the time and behave appropriately in the community. If necessary, acceptions to the criteria will be made to involve team members that are enthusiastic about learning their rights as handicapped individuals.

The group meets twice a week and at the beginning of each month a calendar of events is prepared. The calendar includes two trips into the community, the rest of the time being designated for research purposes.

To learn about the environment surrounding them, the group discovered that first they must be able to identify themselves in various settings. Thus, a trip to the registry of motor vehicles for picture identification was organized. After being able to prove their existence, the residents proceeded on to learning the specifics about accessibility. A population of rules and regulations complied by the Architectural Barrier Board in the Commonwealth of Massachusetts was the primary source used to educate the residents about their rights. The mobility group participants focused on areas concerning administrative issues and architectural design requirements in specific buildings and facilities. With knowledge of the accessibility regulations, the residents ventured into public buildings to review the facilities' access and to pursue any necessary options.

Action was first taken after visiting the public library. It was discovered that the library provided a hydraulic lift for its handicapped population. Contrary to the library's claim of providing handicapped access, not only was the operation of the lift impaired, also the library did not possess a proper permit to use the lift.

Affirmative action was taken by writing a letter to the appropriate city officials requesting repair of the lift. Measures were taken by the city within a matter of a week to restore and license the lift for safe and efficient use.

The community mobility group continues to accress issues concerning their human rights as residents of the community. Their accomplishments have made a significant difference in their living environment. The group's main goal is working towards total independence, mainstreaming back into society!

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A program of the New Medico Head Injury Rehabilitative System.

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SUBJECTIVE RATING OF CLIENTS' ABILITIES AS PREDICTORS OF COMMUNICATION SUCCESS

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ABSTRACT

A questionnaire was developed which asked clinicians to rate clients whom they had seen for augmentative communication, in respect to several cognitive and attitude traits, and to note how long it took for the client to become a comfortable user of a nonvocal communication aid. Analysis of responses for 70 user-device pairs showed that success in learning to use a communication device correlated with two cognitive ratings, and all three of the attitude ratings. For those who were successful, rapidity of learning correlated with both user and receiver attitude to the use of nonvocal communication, while only one cognitive rating, 'Memory,' did so.

INTRODUCTION

As part of our work on the development of a prescription guide, to serve the clinician in making the choice of the communication device likely to provide maximum benefit to her or his client, we have tried to develop predictors of several indices of user-device performance. One of the indices of performance we are concerned with is the practice time required for the client to become a comfortable user of the device. By 'comfortable user' we mean that the demand for visual search and conscious attention to the device during communication has levelled off, i.e., they will not decrease substantially further.

If the client's cognitive and attitude characteristics are rated at levels that we see reliably associated with slow, difficult learning or even abandonment of the use of the device, we can be guided to restrict our consideration to less complex devices. Conversely, if we find evidence of correlation between particular traits and successful device learning, we need not exclude more demanding devices from consideration for clients with high levels of these traits.

MATERIALS AND METHODS

In order to identify predictors of success in learning to use communication aids, a questionnaire was developed to gather clinicians' retrospective reports on their clients' experiences in learning nonvocal communication techniques The population of interest here was people with normal receptive language and cognitive abilities, who could make use of systems based primarily on written language.

The questionnaire asked the clinician to recall patients who needed communication augmentation, and to rate each patient on a scale of 1 (very low) to 6 (very high) for each of 4 cognitive characteristics, called 'Intelligence', 'Spelling', 'Verbal Memory' and 'Memory for Locations' (spatial memory). They were also asked to rate, on the same scale, 5 other characteristics: 3 concerning the patient, and 2 concerning the patient's family and caregivers; these were patient's attitude to technology, patient's drive to communicate and patient's willingness to use a communication aid; and the levels of support provided by the patient's family and/or the people closest to him or her, and by the people responsible for the patient's care. The questionnaire also asked for the patient's age, gender and etiology, the time it took the patient to accept the clinician's recommendation to use a nonvocal communication technique, and the time period between the start of training and achievement of 'comfortable' user status.

Clinicians were being asked to rate cognitive abilities and attitudes on the basis of their subjective impressions at some distance in time, since they were in many cases reporting on patients with whom they were no longer in contact. We therefore included with each rating a 3-point scale on which the clinician was asked to indicate her or his degree of confidence in each judgment, as high, medium or low.

Responses were received for 38 male and 20 female patients, representing 70 user-device pairs which were appropriate for our study, i.e., which involved use of written language, and for which learning time was reported (some individuals used more than one device). Etiologies included cerebral palsy (approximately 30% of individuals); and ALS, head injury, brain stem stroke, anoxic injury and other acquired conditions. Of the 70 devices used, 50 were electronic and 20 were not. Clinicians' ratings of clients' cognitive abilities were examined for intercorrelations. 'Intelligence' was highly correlated with 'Verbal Memory,' suggesting that these may not have been independent traits. Both also correlated significantly, but at a lower level, with 'Spelling'. The meaning of 'Location Memory' was not consistently interpreted by our respondents. It was either treated as identical with 'Verbal Memory' or not rated; and confidence ratings were low. It is therefore omitted from further consideration.

Clinician ratings of the 5 other variables were also examined for intercorrelations. All correlated signficantly with each other. However, only two pairs correlated at a level that accounted for about half of the variance: family support with caregiver support (not

surprisingly, since in some cases family members are also primary caregivers); and again, understandably, the patient's general attitude towards technology with her or his willingness to use a nonvocal communication technique. Our clinician-respondents generally indicated a high level of confidence in their ratings.

RESULTS

Clinicians' ratings were regressed against learning time as reported for each patient-device pair. The learning time reports consisted either of a time period at the end of which the client was a comfortable user of the device or of the statement that the client gave up the device without becoming a comfortable user. Learning time data was therefore treated in two ways. Ratings were regressed against time as a continuous variable; and were also regressed against the binary variable "succeeded" versus "did not succeed". The following two Tables present values for the correlation of each rating with Learning Success (Table 1) and with Learning Time (Table 2).

		r
ATTITUDE	WILLING NVC	.629*
	DRIVE	.340**
	ATTITUDE TO TECHNOLOGY	.447*
ENVIRONMENT	FAMILY SUPPORT	.140
	CAREGIVER SUPPORT	.199
COGNITIVE	SPELLING	.122
	INTELLIGENCE	.425*
	MEMORY	.474*

*p < .001; **p < .005

Table 1. Correlation of Ratings with Learning Success ("succeeded" vs. "did not succeed")

		r
ATTITUDE	WILLING NVC	.391*
	DRIVE	.340***
	ATTITUDE TO TECHNOLOGY	.488*
ENVIRONMENT	FAMILY SUPPORT	.599*
	CAREGIVER SUPPORT	.478**
COGNITIVE	SPELLING	.043
	INTELLIGENCE	.184
	MEMORY	.344***

* p < .001; ** p < .005; *** p < .05

Table 2. Correlation of Ratings with Rapid Learning

Clinicians' ratings of 'Intelligence' and 'Memory' correlated significantly with learning success (vs. abandonment of the device) (Table 1). Among those patients who became successful users, however, cognitive ratings had little to do with rapidity of learning. Only 'Memory' was correlated, and just barely, at the .05 level.

Clinicians' judgments of their patients' attitudes all correlated significantly with success and short learning time. A positive attitude to technology and in particular to nonvocal communication technology, as perceived by the clinician, is related to success in learning to use a device. Finally, the attitudes of support people, the family and care-givers, correlate well with rapid learning but not with success as opposed to giving up on the device.

It should be pointed out that since data was pooled across devices, this report does not touch on how device characteristics affect learning time. Disparities in learning time among devices are considerable, with periods over one or two years reported for some devices, and only one or two days needed for others.

Before proceeding to interpret the user trait findings and assess their potential usefulness to making prescription decisions, it is necessary to deal with an unavoidable problem associated with this data, the potential contamination of ratings by the clinician's

awareness of how successful the patient was at learning to use the device. In other words, the patient may have been judged 'intelligent', etc., partly because she or he learned to use the device rapidly. It is not possible to refute this criticism entirely. Nevertheless, the fact that not all factors, (all of which are generally held to be important) predicted outcome at a significant level at least supports the interpretation that clinicians responding to our questionnaire did not simply rate their clients' abilities on the basis of these clients' device learning success. Indeed 'intelligence' itself did not correlate significantly with rapid learning.

A fundamental problem with interpretation of these results remains - is there any justification for inferring causality from these correlations? Did positive attitudes influence success or were these attitudes the result of initial positive experiences with the device? To the extent clinicians were accurate in reporting on the status of the client's attitude before device training began, we can credit the patient's willingness to use nonvocal communication technology with some role in his or her device success.

Examination of the correlations obtained from receivers' attitudes provides some further weight in favor of giving credence to the clinician reports. Neither family nor caregiver support correlated significantly with success vs. failure in device use. For those patients who did succeed however, the support of the people in their environment did correlate significantly with rapid learning of the device (in days). It is not farfetched to infer cause here. The increase in opportunities to practice afforded by supportive receivers can be expected to enable the patient to become a comfortable user in less time.

Our hypothesis before carrying out this study had been that the patient's cognitive abilities affect how long it takes for him or her to become an 'expert user' of a communication device. This study did not constitute a direct test of that hypothesis. No objective test scores were available; ratings of patient and environment characteristics cannot be considered entirely free of contamination by outcome, and, further, time is measured in elapsed time, not hours of practice. Determining the relationship between cognitive test scores and actual practice time required to reach expertise was therefore beyond the scope of this study. We obtained the somewhat surprising result that the clinician's judgment of the patient's intellectual ability did not relate to how rapidly, in terms of elapsed time, expertise was acquired; and that user attitude and particularly environmental attitude did relate to rapid acquisition of expertise.

CONCLUSIONS

The results of this survey provide support for including the clinician's ratings of the nonvocal patient's and primary receivers' attitudes towards use of nonvocal communication in the patient's needs and abilities profile. Low levels of willingness to use an aid on the part of the patient and/or the primary receivers dictate a need for, and raise the match score of, a simple, low-learning-demand device. High positive attitude ratings suggest that ease of learning and lack of complexity are not vital and that other criteria (e.g., predicted rate of communication) may be accorded relatively greater weight in the prescription decision.

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23.2

TRAINING SEVERELY IMPAIRED APHASICS ON A COMPUTERIZED VISUAL COMMUNICATION SYSTEM

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ABSTRACT

At the VA Rehabilitation R&D Center in Palo Alto, CA, work is underway to develop and evaluate a microcomputer implementation of a visual communication system for severely impaired aphasics. The current system incorporates a vocabulary of representational and ideographic icons, a language-like syntax, and a specially designed interface which uses a "mouse" computer pointing device to control events on a high-quality graphics display. Two globally aphasic individuals are currently being trained in the use of the system. This paper reports on: the use of technology in the current approach; the neurologic speech disorders of the two current subjects; their training histories in this project; and several observations which illustrate, among other things, specific subject adaptation to the computer based system and performance improvements attributable to the introduction of the computer.

INTRODUCTION

There are 84,000 new cases of aphasia each year, due primarily to cerebral vascular accidents (CVA) and to traumatic head injury.[1,2] Because of improved treatment and better medication, individuals are surviving CVA and head injury in greater numbers, and subsequently are living longer. However, despite current speech rehabilitation work by dedicated neurologists and therapists, approximately half the new cases of aphasia each year remain severely language impaired. The efficacy of speech therapy appears modest in comparison with the effects of spontaneous recovery[3], and no particular method of speech therapy has been demonstrated to be superior in controlled studies.[4] Aphasic individuals not helped by existing therapies within the first half year remain in need of some new form of communicative assistance.

CURRENT APPROACH

The current approach draws upon modern technology to construct an alternative communication system, which can be used by some chronically severely impaired aphasic individuals.[5,6] The system is designed around the residual strengths of the subjects, to bypass damaged functions which have not been restored. It relies heavily on subjects' visual processing capabilities — their ability to identify objects and conventional representations visually — and their abilities to put them in proper spatial sequence and to assign meanings to the resulting constructions that refer to events in the world. Subjects trained in the use of the system have shown an appreciation of the communicative potential of the system.

The work builds upon and extends earlier studies conducted in New York[7] and Boston[8,9], and shares some similarities with recent work abroad.[10] Our work is based most directly on the *VIsual Communication* (VIC) system described by Gardner[9], which used representational and ideographic icons drawn on index cards as vocabulary items. The Boston study demonstrated that some appropriately screened aphasic individuals can master the fundamentals of the of the VIC system, using it to: 1) answer questions; 2) receive commands; 3) describe on-going events; and 4) communicate simple wants and feelings.

An important difference between our work and previous studies is our incorporation of contemporary microcomputer technology. This technology affords a new set of advantages and opportunities as a communicative orthosis for aphasic patients. A high resolution screen allows the construction of easily recognizable iconic images. A display of "cards" on the screen eliminates the necessity for manual manipulation of large stacks of real index cards and the placement of such cards on a table — operations which are difficult at best for hemiplegic and apraxic patients. Instead, patients use solely the "mouse" pointing device — operable with one hand — to "spread" the stacks of cards on the screen, select individual cards, and arrange them into meaningful strings in communication spaces on the screen. Six stacks of cards are portrayed on the screen, containing from left to right: punctuation, proper nouns, verbs, prepositions, common nouns, and modifiers. Translations of the symbols are displayed below the icons on the screen, and primitive word-for-card translations appear on request. This implementation of the visual communication system on a microcomputer significantly diminishes the motor and organizational demands made upon the subject. This relatively early stage of the VIC computer implementation has already demonstrated the power of the approach and and has pointed the way for future developments, which will more fully realize the potential of high technology components.

SUBJECTS IN TRAINING

We are currently training two severely aphasic patients to use the system. The first subject, JN, is a 45 year old, right-handed man, who was in good health until 1974, when he suffered a head injury in a motorcycle accident and subsequently underwent a partial frontal lobotomy. Since that time, JN has suffered from a severe expressive and receptive aphasia. Despite diligent work on the part of the patient in four separate speech therapy programs, all attempts at traditional speech therapy have been completely unsuccessful. On the Boston Diagnostic Aphasia Examination (BDAE), JN shows subscores near or at the zero level for naming, reading and repetition. His Profile of Speech Characteristics reveals profound impairment in fluency of free speech and absence of melodic line. No scores for phrase length, grammatical form, paraphasias in running speech and word finding were obtainable, due to profoundly impaired articulation. Auditory comprehension is somewhat less impaired, with a Z score of −1. Spontaneous written expression is confined to highly routinized sequences, such as the patient's name and address, while the patient exhibits moderate ability to copy, occasionally replacing printed forms with cursive script forms. There is no comprehension of written material.

The second subject, JS, is a 54 year old, right handed man whose left hemisphere CVA in April 1985 left him with a severe global aphasia. Examination of the patient following the left CVA revealed an old right CVA near the Sylvian fissure of unknown onset. Intensive traditional speech therapy between April and October 1985 produced mixed results. PICA subscores for Auditory comprehension during that period improved from 5.96 (23rd percentile) to 8.33 (41st percentile), while verbal ability declined over the same time from 5.00

(21st percentile) to 3.00 (5th percentile). Subscores for gestural and graphic abilities improved over the period, the former from 6.91 (20th percentile) to 11.18 (52nd percentile), the latter from 5.33 (37th percentile) to 8.08 (74th percentile). JS's expressive deficits are profound, with no comprehensible verbalizations whatever and severely impaired articulatory agility. Examination reveals no useable word finding or word repetition ability, and no facility for grammatical form phrase length or melodic line. He can write some stereotyped phrases, such as his name, and can copy directly with some accuracy, but shows no comprehension of written material. He has a very severe receptive deficit. His recent Z scores on the BDAE are less than −2 for auditory comprehension. He possesses excellent writing mechanics and discrimination of written symbols, but has essentially no comprehension of written material.

SUBJECT TRAINING HISTORIES

Our first subject, JN, has been in training on the computerized system for approximately one year. He comes to the Center twice a week, and trains an hour each visit. He has thus received approximately 100 hours of training on computer based VIC. Prior to introduction to the computer, JN trained for several months using the card-based system developed by Baker[8] and Gardner[9]. JN has a deck of VIC cards of his own which he uses to practice in the system with his father during the days between training sessions. He became quite proficient in use of the card-based system, and lessons learned by the researchers in teaching him this system allowed the researchers to extend the inherited VIC system in ways which were carried over into the computer implementation. JN is unique in that he provides the only instance of a VIC card user transferring over to computer VIC. The ease with which this transfer was made was a matter of interest to the authors, and is discussed below.

Our other subject, JS, is currently an inpatient at the Palo Alto VA Medical Center, receiving intensive VIC training five days per week, two hours per morning. At this reporting, he has completed twenty hours (two weeks) of VIC instruction. He has been trained on the computer from the outset, has never handled the VIC cards (although the trainer does during parts of the training sessions), has no deck of VIC cards of his own, and does no VIC practice between training sessions.

Despite differences in the training time profiles of these two subjects, the order in which VIC communications have been introduced to the subjects and the ways in which they have been drilled, the ease or difficulty with which they master and make use of a given construction, and their communicative strategies are broadly similar.

SUBJECT PERFORMANCE OBSERVATIONS

Several significant observations on subject performance have emerged from the training activities. Four of these are:

1. Both subjects can perform using VIC at a level far exceeding their capabilities to perform using natural language, responding to simple and complex commands, for example, with near 100% accuracy. Both patients can discriminate between, and respond appropriately to, commands, statements, yes-no questions and content (*wh*- word) questions, responding to the latter with declarative statements. In the case of our earlier subject, JN, these declarative statements include the use of conjunction, direct and indirect objects, prepositional phrases, quantification and expression of simple personal feelings and desire; training in these areas has just begun with our newer subject, JS.
2. The ease of learning and using VIC lexical items appears to be correlated with the part of speech. Common nouns, proper nouns and numerals are mastered most quickly and retained most accurately. Fig. 1, showing early training data from our first subject, JN, illustrates this point graphically. In the first four sessions, during which approximately twenty nouns were in use, JN's accuracy of association of common nouns with their icons and vice versa ranged between 90% and 100%. Accuracy remained essentially at 100% thereafter. The introduction of new nouns required no reinforcement of previously acquired nouns, and absence of training over time did not result in extinction. A similar pattern has held true during the training of JS, the second patient reported in this study. Fig. 2 shows a different sort of performance, one which seems characteristic of verbs and prepositions. With these items, accurate performance seems to be acquired only over time, with drills designed specifically to differentiate between items. In these categories, the introduction of new items leads to generalized degradation in performance with previously appropriately used items, and overall performance improves slowly, as a result of exercises to drill the differences between the new and all old items.

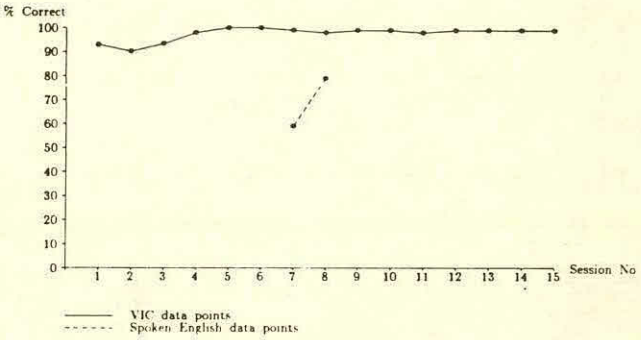

Fig. 1: Common Noun Acquisition

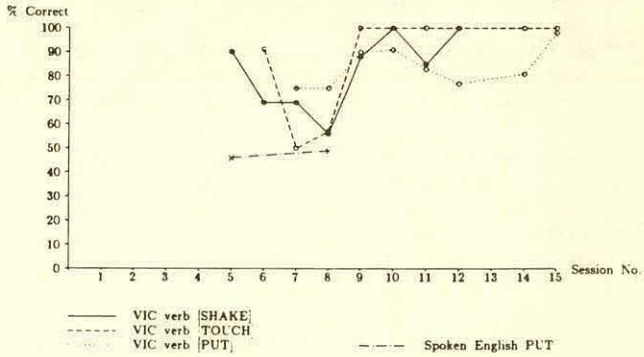

Fig. 2: Verb Acquisition

3. Our subjects have been able to master the principles of the computer interface operation both quickly and accurately. This requires the ability to select icons from a type of "pull-down menu," similar to those on Apple's Macintosh computer, and to conduct icon searches through such pull-down menus, which represents an extension of the

standard Macintosh interface philosophy. These tasks are conducted quickly and accurately by our subjects. JN, our first subject, was a proficient VIC card user at the time of introduction to the computer program. In two hours on the computer VIC implementation, he was able to master the operation of the interface, transferring his knowledge of VIC to the new medium with ease. Significantly, as he gained facility in using the interface, he developed strategies which demonstrated an understanding of the nuances of its operation. For example, he would minimize search times by preparing the interface's icon presentation (according to his expectations) in advance of transactions, and by skimming through icons on the screen whenever experience told him that a significant distance remained to the desired item. Our second subject in this study, JS, has been trained on the computer from the outset. He also acquired the fundamentals of interface operation within two training sessions, and he also handles its complexity without difficulty.

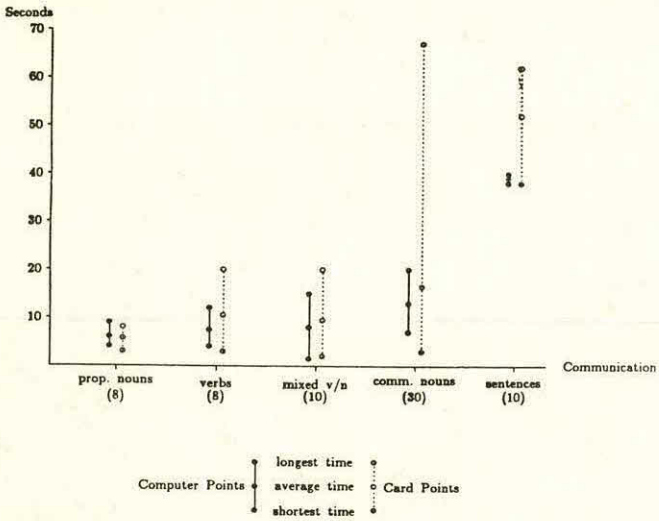

Fig. 3: Times — Cards vs. Computer

4. JN's performance demonstrates that the time required to access and correctly place an icon using the computerized system is both more consistent and — on average — shorter than the time required using index cards. Fig. 3 illustrates this point graphically. The performance improvement stems from the increased ease with which icons can be manipulated on the screen. In the computer system, all operations can be conducted with one hand, moving the mouse to position the cursor on the screen, and pushing the single mouse button to select an icon, transfer it into a communication space, or erase it. When icons are being scanned for selection, eight are presented on the screen at a time, thus utilizing parallel visual processing, and another eight can be brought up in a fraction of a second, simply by releasing the mouse button and depressing it again. Selection of an icon requires only positioning the cursor on the icon so that it appears in reverse video (indicating that it is available for selection), and releasing the button. This contrasts dramatically with the awkwardness of patients, even those with only a mild hemiparesis, when handling stacks of up to 30 index cards, shuffling through them, selecting individual cards, laying them out on a table top, and reorganizing the cards into proper files once a transaction has been completed. The relative ease of using the computer system is noticeable during therapy sessions, encouraging the subjects towards more frequent and more complex communications.

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Application of Multiple Wordstudy Analysis in the Development of Vocabularies for Augmentative Communication Systems

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ABSTRACT

While many studies of word frequency use have been conducted with normal individuals, it is difficult to determine the extent to which this reflects the speech of a given client or situation. Even the top fifty most frequently used words may differ by more than 50% between different word frequency studies. Although some words are consistently high frequency across all settings, frequent variation in vocabulary production is reported between population samples discriminated by age, mode of output, task, setting, sampling method, and units of analysis. A wordset for an augmentative communication system is ideally tailored to a client's age and educational level but applicable across settings and communication modes; frequency information for a wordset should then be based on an appropriate sample(s) and augmented by information on vocabulary variation across relevant settings/modes. This paper outlines considerations and available resources for vocabulary selections utilizing frequency data, based on an overview of a bibliography collected at the Trace Center, Madison, WI, of over 350 studies related to vocabulary frequency in normal individuals.

The most frequently quoted reference in psychological or educational research for relative frequency of vocabulary use is still Thorndike (1944). However, Thorndike and many other large scale studies (Kucera & Francis, 1967; Carroll et. al., 1971; Zettersten, 1978) are derived from analysis of printed texts rather than productive written or spoken interaction. These types of text analyses are appropriate for many educational tasks, such as estimating readability, but are not intended to be estimates of word frequency in conversation or written interaction. Studies of spoken or written production (such as Howes, 1966; Wepman & Hass, 1969; Schonell et al., 1956; Fitzgerald, 1934) more closely relate to output of communication systems, but tend to analyze a single setting and a much smaller dataset and may be limited in their application across settings. Any single study is limited by size and representativeness of sample, and even the Beukelman et. al. (1984) sample of output by communication aid users has limited utility for recommending wordsets for augmentative systems, because of the high variability between individuals, settings, and other linguistic factors.

A demonstration of the degree of potential variability between wordsets is provided in Table 1. For this study, the Thorndike list was used as a base. Ratios indicate the percentage of the top 50 words of 13 wordset studies as compared to the top 50 words of the Thorndike study. The 50-word samples from the 13 studies differed by 14-54% from the wordset derived from Thorndike. More importantly, the words which contribute to the variation differ according to the population sample (child, college

TABLE 1

% SIMILARITY IN TOP 50 WORDS
BETWEEN THORNDIKE AND OTHER WORD FREQUENCY STUDIES

Frequency Study	Sample Characteristics	% Agreement
Kucera & Francis (1967)	A/ta/1,000,000 words	86%
Dewey (1923)	A/ta/100,000 words	84%
Black & Aursherman (1955)	A/s/college student speeches	78%
Horn (1926)	A/w/personal and business leters	76%
Rinsland (1945)	C/w/children's school themes	72%
Beukelman & Yorkston (1984)	A/s,w/augmentative communication aid users	68%
Berger (1976)	A/s/college student conversations	64%
Jones & Wepman (1966)	A/s/picture descriptions	64%
Wepman & Hass (1969)	C/s/picture descriptions	60%
French, Carter & Koenig (1930)	A/s/telephone conversations	60%
Fitzgerald (1934)	C/w/4th-6th grade personal letters	58%
Edwards (1964)	C/w/free writing	48%
Cameron (1970)	A/s/college student conversations	46%

A = adult
C = child

s = spoken production
w = written production
ta = text analysis

students, adults), mode (speech, writing, text), setting (school, home, office, public places), task (free writing/speech, conversation, formal themes/speeches), and other sample characteristics. For example, some of the words unique to the top 50 of different populations include: contractions used in slang, e.g. "I'll, You're," and profanity (college student speech), words like "play, gonna, boy, put, house" (children's speech), and words like "please, seem, would, because" (adult writing). Even a combined wordset derived from word frequency studies across all settings has limited application for communication aid selection: it would both exclude high frequency vocabulary unique to a given population or setting, and include vocabulary unrepresentative of client or sample characteristics

The following are a series of observations and resources for normative data to assist adaptation of vocabulary sets, based on review of over 350 sources related to vocabulary frequency collected at the Trace Center. (Complete copies of the bibliography are available from the Trace Reprint Service).

The top 10-20 words, such as "you", "a", "and", "the", and "in", tend to be consistently frequent across wordsets from all communicative settings. Most words within the top 100 are also frequent across different wordlists, but include unusually frequent words characteristic of a given sample or setting (such as profanity, or words about play). Beyond 100 words, it becomes increasingly difficult to predict relative frequency of any words across wordsets, except within a setting utilizing limited vocabulary (such as writing letters about farming equipment). Because of the high variability of human linguistic output, word frequency lists are best used only as a sample vocabulary rather than sole determiner of a communication aid wordset.

For some clients, supplemental wordsets for limited professional or educational purposes may be appropriate. Normative data is available listing the most frequently used words for specific professional tasks such as technical writing, business letters, classroom speeches, and other vocational purposes (e.g. Johansson, 1978; Hartsough & Laffal, 1970; Cole, 1940). Also, some frequency or core vocabulary studies provide suggestions for vocabulary related to such disabilities as CP, MR, deafness, and cultural disadvantage (e.g. Irwin, 1966; Mein, 1961; Walter, 1978; Sherk, 1973).

If part-word units are utilized as spelling aids, the applicability of frequency ranking of words changes, even if only 50 words are coded in a device. For example, introducing "the" as a spelling aid as well as a word alters the usefulness of including the following high frequency words: the, then, them, they, their, there, and these. Data available on frequent letter/phoneme sequences can help predict which phoneme sequences occur most frequently, and which of those represent linguistically relevant units (see Carterette & Jones, 1974; Hanna, et. al. 1966; Roberts, 1965; Solso, 1979).

Most top 50 lists also include items with closely related content, such as have/had/has. Wordlists such as Basic English propose a representative set of words to fulfill a range of related meanings in order to reduce such perceived redundancy (Ogden, 1932; West, 1953). While Basic English is controversial as sole medium of communication, principles may guide training to minimize redundancy of wordsets and number of words used to express ideas.

Determining vocabulary for communication aids must balance between flexibility and ease of use. Ideally, all possible sentences or units of sentences could be easily coded and retrieved, but the cognitive load of remembering codes would eventually interfere with performance. This is true however the vocabulary is coded. Various studies for measuring cognitive load of a language sample may be useful for determining size and arrangement of wordset. Pertinent information includes readability (Chall, 1958), syntactic units (Corson, 1982), encoding strategies (Paap & Ogden, 1981), and other processing demands (Schwartz, 1976; Schneider & Shiffrin, 1977).

Communication devices must often serve as both speaking and writing aids for clients. This makes vocabulary selection more difficult, because the processes and vocabulary uses are different between these two communication modes: e.g. speech is typically faster and more often redundant than written communication. Studies such as those by Chapanis et. al.(1977), Weeks (1976), and Earl & Goff (1965) describe differences in vocabulary and communication strategies between speech and writing, as well as analyzing the effects that different communication devices have on language content and use.

For augmentative system users, relative speed is reflected in the number of keystrokes or motions necessary to convey a message. The longer and more frequent a word, the greater the keystroke savings if that word is represented by a reduced code. Current frequency studies reflect only frequency rankings, and need to be reorganized by length/frequency information to facilitate calculation of keystroke efficiency.

Greater keystroke savings may also be accomplished by coding the most frequent phrases or word combinations as well as words. Hart, Walker & Gray (1977) provide normative data on the most frequent 2- and 3-word phrases used by children, which include: "I'm going", "I'm not", "I don't know", and "I want to". Even if some phrases or sentences provide fewer objective keystroke savings than words, it may be appropriate to include them on communication aids. Particularly in conversation, the most important words/phrases may not be what is said most often, but what is needed to say most quickly.

It is certainly possible to derive a list of 50, 100, or 200 frequent/useful words without utilizing any frequency or related information. However, individuals are reported to be poor at identifying or estimating frequency of words, with a tendency to note unique utterances or omit functor words. In deriving a communication aid wordset, clinicians must try to balance normative data on word frequency with clinical impressions and iimited samples of linguistic output from a client. One of the functions of research is to provide maximum information about the application of word frequency information across different settings, to minimize clinician effort to fit the communication aid to the person, situation, or task.

In summary:

- We have more resources available for wordset selection than the straight word frequency studies we have been using; other available information includes normative data on technical vocabularies, vocabularies for special populations, differences in language and strategies between communicative modes, frequencies of letter/phoneme sequences or conversational phrases, semantic reduction, or cognitive loading of different coding strategies.

- Single vocabulary frequency studies alone are limited in application to the sample they study, and combined wordlists

across different settings often obscure important differences between age groups, settings, tasks, or sampling techniques. Instead, comparative information derived from frequency and related studies can provide data on group differences to justify and apply clinical impressions of individual differences in vocabulary use.

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WORD MENU REDUCES COMMUNICATION RATE

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ABSTRACT

Two naive subjects, training with two devices which differed in menu size and arrangement, were videotaped and timed. Message production rate improved with practice for the user of the simpler device. The other participant made no improvement in rate during 3 hours of practice. Rate worsened when he selected whole words from the menu. The need for use of a standard terminology and assessment protocol in the discussion and analysis of device features, and for acquisition of user-device performance data, is discussed.

INTRODUCTION

In order to identify characteristics of people and of communication aids which affect success and speed of learning, a study was planned in which naive participants (not actual or potential device users) would each train and practice with devices chosen to represent extreme values of device parameters which were likely to affect 'learnability'. We wanted to allot each participant enough practice time with a device to become the equivalent of an experienced user. In an earlier paper, the two senior authors proposed a model of how experienced users operate non-predictive keyboard devices (i.e., devices that do not offer predictions that the user needs to accept or reject)(1). Once most of the 'cognitive' learning has taken place, that is, once the user no longer has to search for menu items, communication rate is largely determined by motor factors, i.e., the interaction of user motor abilities and device menu layout and geometry. Dependence of the operator's movement time on the physical parameters of the device is what the motor assessment technique we have developed (2)(3)(4)(5) is aimed at characterizing. Predictions of rate derived from this technique therefore can be validated only by observation of rate in actual message production after completion of 'cognitive' learning.

We were therefore interested in obtaining some data which would help us estimate the time we would need to allow for our study subjects to reach 'expertise'. Accordingly, we asked two individuals to participate as subjects in a preliminary study, in which each was to undertake learning to produce English text by means of a particular communication device. We selected keyboard devices which were at opposite poles in terms of menu size and complexity of organization of the menu. One was the Canon Communicator, the other a predecessor of the current Vois devices, the Handivoice 110. The Handivoice featured a large menu of words arranged according to 'semantic' principles, in addition to an alphabet and a phoneme alphabet. (Note that the newer Vois devices offer menu options which can be used instead of the one described here.) The devices also differed in visual complexity, the Handivoice items being displayed in small print on small background areas in four colors.

Of course, there are techniques in existence which are considerably more complex in concept than the Handivoice, such as Minspeak, which entails the individual user's invention and memorization of a menu, picture-code and layout. In order to measure the 'learning time' needed for a study participant to become a fluent Minspeak user, it would be necessary to have this person design his or her Minspeak system - an endeavor which was beyond the scope of this preliminary study.

We expected the Canon to be the more quickly learned of the two devices, and to mark more or less the lower bound of the amount of time a subject would require to reach expertise with a device. We were prepared to find that the large-menu device, with its non-alphabetically ordered words, and visually imposing overlay, would take longer. If communication rate achieved with this device was found to level off within a feasible time frame, we would be in a position to plan our device-learning study realistically.

METHODOLOGY

In order to keep our data free of delays due to composing (making up) messages, we gave our two participants English text to copy. The text was excerpted from a book on etiquette. The language was simple in style and vocabulary and dealt with everyday situations. To reduce demands on reading and memory, and any contributions to time as a result of them, we presented the text on cards, one line per card, using natural divisions in the text to end each line. Each participant was instructed to read, remember and reproduce the line of text, then flip the card over to get the next line of text. It was acceptable to consult the card in mid-line, but this rarely proved necessary. Since timing was done only within a line, from the first key press to the last key press for each line, we could be confident that reading time per se had no appreciable effect on time measurements.

Subjects' device operation during training sessions was videotaped, with the camera sighting over the subject's shoulder to give a clear view of the hand and the keyboard. A

protocol was developed for coding subjects' keystrokes, errors, edits and omitted edits. This protocol allowed calculation of actual keystroke rate, time per output character (the inverse of effective rate) and C, the number of language units (items from the language menu) selected on average per word. Five-minute samples of performance were transcribed at regular intervals. Elapsed time was read off a timer placed next to the device and included in the videotape. Transcription and coding versus sample duration required approximately a 5:1 time ratio after some initial practice. The transcriber measured time per line of text (i.e., for each period of continuous message production), and then divided the elapsed time by the number of keystrokes the subject had produced (minus 1), to obtain time per keystroke. It was necessary to distinguish between time per keystroke and time per correct output character, since keystrokes could include errors and edits, and, in the case of the word-menu device, level selections and entire words.

It was also of interest to calculate C for each five-minute sample of performance with the Handivoice. While C for a device menu can be considered relatively stable for large samples of standard spoken English, this is of course not the case for samples as small as those the new learner produces in the course of five minutes. Since the Handivoice menu included words which occurred rarely in the samples, it was expected that C would vary greatly from one sample to another. A word which could be produced by means of a single menu item could reduce C for the sample in which it occurred and, therefore, reduce calculated time per character. For example, production of a 5-letter word by one key press adds 5 to the denominator and only 1 movement time to the numerator, in calculating time per character.

RESULTS

S1 -- Canon Communicator
S1 spent 3 forty-five minute sessions using the Canon to transcribe text from cards as described above. All sessions were videotaped. Six five-minute samples were transcribed and timed in each of the first and third sessions. Average time per character in session one was 1.03 seconds; by session three, average time had been reduced to .76 seconds. The slowest time/character calculated for a sample in the third session was still significantly faster than the fastest sample average in the first session. It was clear that significant gains had been made by the time of the third practice session. However, a range of .25 seconds between best and worst third session sample average suggested that we could not be confident that 'expertise' had been reached, since we would expect expert performance to be characterized by, among other things, a low level of variability. Thus, even for this 'best case', 3 forty-five minute sessions were not clearly adequate to achieve stable, expert performance.

S2 -- Handivoice
S2 spent 4 forty-five minute sessions transcribing text from the prepared cards. Six five-minute samples were transcribed and timed in each of the four sessions. This device offered a menu which included whole words, phonemes and an alphabet. We chose not to ask S2 to learn to use the phonemes, but we did encourage him to use whole words when the ones he needed were available. When we plotted average time per character for each sample we were surprised to note that there was not much of a trend at all unless we included the fourth session, and what trend there was was decidedly towards slower performance! In spite of sustained efforts, S1 was not managing to achieve any reduction in time per character with practice. While during session one he averaged 2 seconds/character, during the next two sessions his times ranged between 2 and 4 seconds/character. During the last session -- ever, as far as he was concerned -- times ranged from 4 up to 6 and finally 12 seconds/character for the last sample.

We then calculated actual C for each of the samples and computed the correlation between C and time per character for the 12 samples in sessions two and three, after initial exposure to the device but before complete discouragement had set in. Note that our expectation, as discussed above, was that these two variables would correlate. Indeed, the correlation was significant; however, it was in the opposite direction: -.52. Selection of whole words from the device menu significantly slowed S2's rate of message production. Review of the videotapes made the reason for this effect evident: S2 was spending lengthy periods searching the overlay for the locations of desired words. It was evident that for this highly cognitively demanding device, three hours of practice amounted to a mere drop in the bucket.

Confirmation of the wide range in learning times experienced by users of these devices was obtained via a questionnaire distributed to a number of clinicians (6). While learning time reports of a couple of days or less were not infrequent for devices like the Canon, there were periods of over a year reported to be required before some users could operate their device without spending time in visual search of the interface to locate desired menu items. Clearly, there is more than one device 'improvement' which, for some of its users, entails a learning effort and investment of time which seem hard to justify.

DISCUSSION

A great deal of effort is currently being directed, in many research and clinical settings, as well as on many home computers of

friends and family members of people with augmentative communication needs, towards improving these systems. Many modifications and techniques are presented with accompanying claims for the improvements they represent and the benefit they afford the user. In the absence of systematically collected performance data, to relate these features to such measurable aspects of performance as communication rate, amount of use (not just of the device; but of the innovative features themselves) and learning time, these claims need to be treated with caution.

It would be helpful if systems could be discussed and evaluated in terms of a generally applicable terminology and framework. A framework has been developed within which any system can be considered in terms of the menu, code and layout, and the motor demands placed on the user (7). A proposal has been made that this terminology and the analysis it entails be used to develop criterion scores based on a standard user model (8) (or a set of standard user models representing etiological, severity or simply motorically distinctive subtypes). Research pursued in this framework has resulted in a procedure which allows calculation of 'motor-determined maximum rate', the upper limit to the performance which could be achieved by a user-device pair. Testing and continued development of this procedure are in progress. Efforts will be directed in this further development at enlarging this framework to accommodate quantitative assessment of the effects of user cognitive abilities and device cognitive demands on performance. What is required is the acquisition of data on learning and on expert use of devices to characterize how user abilities and device parameters determine learning rate, and how they impinge on expert performance. Modification of our prediction equation for motor-determined maximum rate by inclusion of appropriate, experimentally determined terms to account for device cognitive demand would allow us to come closer to predicting actual in-use rate. In other words, our expectation is that mental load in expert use entails a price in sustainable communication rate. 'Automatization', i.e., overlearning, of the device operation rules and menu overlay represents minimum mental load and thus should make it possible for the user to achieve an actual rate, at expertise, which is as close as can be to to the predicted motor-determined maximum. Measurement of mental load in expert use by appropriate adaptations of 'traditional' mental load measurement techniques or use of trained, able-bodied study participants who could carry out standard mental load assessment tasks (9) (10) would provide another source of experimental data which could significantly improve prediction of actual functional performance.

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DILEMMA OF PREDICTION IN COMMUNICATION AIDS AND MENTAL LOAD

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ABSTRACT:

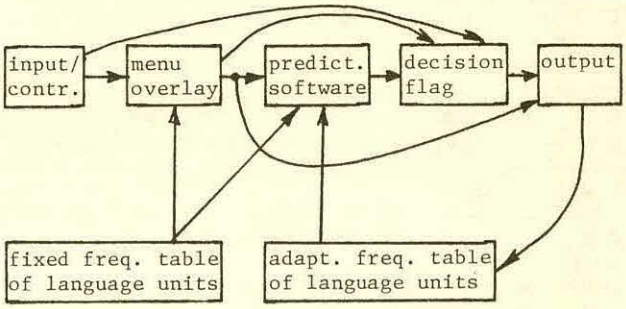
The use of prediction techniques in communication for the disabled is claimed to provide a major enhancement of communication speed. The problematic aspect of prediction is that interaction between the (computerized) communication system and the user becomes more complex: the user has to decide after each prediction whether the machine proposed character is what he or she intended or not. Thus, while prediction may bring about a savings in keystrokes per word, it is responsible for an increase in mental load. This problem is discussed, a model for prediction presented and new research initiatives proposed.

PROBLEM DEFINITION:

The design of augmentative communication aids is discussed in many articles, especially since the possibilities of microcomputing have become readily available. One of the most severe problems, for many people who experience motor and speech impairment, is the low speed of communication they can attain. While device users may often be quite motivated to take the time which is needed to construct elaborate sentences, conversation with a communication partner is almost impossible to sustain when rate is below about 3 words per minute (1). Many research efforts are therefore directed to the problem of improving the input interface for communication system users. It will not be necessary in this paper to describe the various input systems as they are used in practice. The essential issue is that a motorically handicapped person often can only use a very small number of keys or switches, or operates very slowly with the normal set of keys. Improving the input interface is a complex problem which can be addressed in several ways. One of these is the "optimization" of the physical interface, i.e. the type and number of switches, the type and characteristics of a joystick or a specially designed device with respect to the abilities of the patient (2). Secondly, the structure of the language representation and coding system (menu overlay) can be of major importance (3,4,5,6). For example, if a scanning system is used, the average time for selection of language units can be greatly diminished by arranging the units on the basis of their expected frequency of occurrence. Another type of approach is to predict language units on the basis of previously produced text. The basic idea here is to take advantage of redundancy which is inherent in normal text and spoken communication. The assumption is always in this case that a string of units can be seen as a sequence of events with a probabilistic behavior (7). Prediction means, subjectively, that the computerized device tries to present the next letter or word on the basis of whatever information is available to it at a particular moment. In a theoretically optimal system, with a 100% prediction success rate, it will appear to the user that the next language unit is just a calculated outcome based on the preceeding units and environmental context. In this case prediction becomes equivalent to a system of a large number of encoded messages larger for example than that of Speedkey (8). In contrast a poorly functioning prediction system which seldom anticipates correctly can be characterized as a noise generator. However both of these systems have at least one attractive aspect: the user does not have to be concerned with the system's errors. In the first system no errors are made, while in the second system the user will just ignore all predictions. A severe difficulty arises with systems which have a correct prediction rate lying between the above-mentioned extremes. The user of such a predictive system has to consider each new produced element consciously: is it right or wrong. The trade off between the cost of decision making (right or wrong) and the benefit of keysaving will be different for rapid ten finger typing and in case of a severe motor handicap.

A MODEL OF PREDICTIVE COMMUNICATION:

Since small computers have become available at low prices new communication device development has taken advantage of these machines. This has made the use of rather complex techniques quite feasible and also means that the use of long tables and lists is no longer a problem. The figure below represents the use of prediction in a computer-based communication aid.



The two blocks, labelled input and menu overlay in this figure, are already indicated above as areas in which work has be done with the goals of improving communication function. The third block of the diagram is the predictor software. This block uses either a fixed frequency table of probabilities of language units or an adaptive table which is created automatically during the use of the system, or could use both tables. The fourth block is the decision flag. The user has to evaluate the predicted language element and signals the machine, either with a special key or by means of the menu, to process the predicted unit or cancel it. The last block, output system, can be any appropriate display system. Although the type of display is important for composition of

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the other blocks in the system it is not relevant in this paper. The fixed frequency table is most often used as the basis for design of language menus. The use of an adaptive table in this case entails that during text production the menu has to be altered. It is evident that such a situation is undesirable since it would dictate a need for constant relearning.

ASPECTS OF USER DEVICE EVALUATION:

The problem of **communication rate prediction** has been extensively discussed elsewhere, e.g., (3,9,10). Incorporation of a predictive component in the device offers the possibility of reducing **L, the number of motor acts per language unit**, since entire words might be produced via only one key press each, denoting acceptance of the machine's prediction. However the need to attend to the machine's predictions before each motor act is carried out carries the possibility of increasing **T, the time needed per motor act.** The resultant effect on the product of these two terms and on rate is not a priori predictable.

Another aspect to be considered is that speed is not the only consideration in communicative function. Certainly, in defining a measure for the usefulness of a communication system speed is of major importance, but also **the ability of the user, and the possibilities of the system** to expres the intended message as well as possible. This includes the availability of a variety of expressions which are adequate with reference to communicative abilities of the user and current communication situations.

A further aspect is the **"ease of operation".** This aspect will be discussed in more detail in the following paragraphs.

Anticipator

MENTAL LOAD AND PREDICTION: A CONTRADICTION ?

Bernstein

Prediction has been employed in a variety of devices:

The **ANTIC system** developed at Tufts University (12) uses letter prediction such that 50 % of keystrokes can be saved.

The **ANTICIPATOR** (13) does letter prediction resulting in a 20 % keystroke saving. Experiments with combining prediction with typing on a standard keyboard showed a slower typing speed, presumably due to the additional decision load (mental load).

The **PAL system** from Dundee (14) was originally designed for use in simultaneous subtitling for the deaf. Due to predictive word completion, a 40 % keystroke saving is obtained with a normal keyboard.

The **PREDICT system** (7) is designed to accelerate programming speed by using wordprediction. Thus, the contents of the frequency tables cover only "computer vocabulary" (keystroke savings of up to 50 %).

MICROPECH (Heckathorn)

Communication by means of an augmentative communication aid can be seen as production of a series of language units which is a **stochastic process of events** occurring at varying intervals. In the case of prediction we have to take into account another stochastic process which is strongly related to the first one: a series of right and wrong events. These events require an acceptance decision by the user in the case of correct prediction. Incorrect prediction also takes some of the user's attention. Referring to the literature of mental load, this situation is well known as a dual task situation (15,16,17,18). The primary task in this situation is the selection of language units, while the secondary task is the judgment of the predictions. Depending upon the details of task execution, it is possibly that the primary and the secondary task can strongly interfere with each other. This can be the case, for example, if the same key or input signal is used for selection of language units as well as the acceptance decision, or in case the independently entered language units are presented in the same way as the predicted units. To get more insight into this problem it is clear that research has to be carried out to assess levels of mental load during judgment of predictions as well as during independent entry of language units. Available **methods for mental load measurement** (15,19) are not necessarily applicable due to the deficits of this population. The method of filling up spare mental capacity with an additional task (in the case of a predictive system, the use of a third task) may be difficult to use for want of a performable additional task. Also physiological measures in current use, such as heart rate variability changes, breathing rate changes and skin conductance can be problematic due to the fact that neuromotor deficit can be accompanied by abnormalities of the autonomic nervous system. An alternative approach to estimate mental load levels is to simulate the situation with non-handicapped subjects. However the predictive validity of such an experiment can be questioned. Also, an information theoretical approach (20) to the mental load measurement problem might fail. Although from the viewpoint of a stochastic process which produces an event series, information theory is applicable, it is not well understood how semantic aspects of the communication can be related to bit rates and redundancy.

Schwijda
Hunnicut
Woltosz

RESEARCH NEEDS:

Derived from the problems mentioned above, there are some areas of research which can be indicated as follows:

Criteria for efficiency or usefulness of a communication system are needed; it might not be sufficient to predict the communication rate of a person-machine combination although it appears a major aspect of usefulness. The definition of usefulness should also include a suitable mental effort to produce the messages, and the suitability of the language system with respect to the user's possibilities. This problem is already indicated in the available literature.

The estimation of mental load in a predictive system has to be studied; it is expected that this will give more insight in basic mechanisms of predictive communication. From the literature on mental load it is seen that a desirable level is hard to define. Some subjects might feel comfortable at rather high levels, others prefer very low levels. A general problem is that large variations in the levels of mental load are annoying to almost anybody. Especially while functioning on a rather low level, it is very disturbing to experience an unpredictable high level. This could be the case if normal routine is interrupted by unpredictable events.

Furthermore, elaborating the usefulness of communication systems with a higher speed, one shouls also consider other schemes which enhance communication.

The word finding software (21) for patients with minor anomia is very interresting. Based on a number of features (clues) it could also be used to **produce complete sentences** having a correct grammar. Software which is developed for analysis of sentences is based on the same software which is needed for this automatic sentence production system. A preliminary solution is the storage of a large amount of sentences in a computer memory which can be recalled on the basis of given clues.

Due to the very strong interaction among the different elements of a communication system, input interface features, menus, prediction schemes and output configuration, it is necessary the that above mentioned research has to be seen in the context of the whole system and the abilities of the potential users.

CONCLUSION:

It is obvious that prediction in communication devices will get much attention due to advances in software which makes it possible to exploit language redundancies in useful and creative ways. These issues are complex; it is therefore not expected that much progress will be noted on a short term. It is however possible to anticipate these developments by performing laboratory simulations of what is expected. Availability of much more language data concerning a corpus of child language, and a corpus of sign language will stimulate these developments.

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OPEN-LOOP CONTROL OF THE HUMAN KNEE DURING SWING

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ABSTRACT

Electrical stimulation open-loop control strategies are oftentimes successful in restoring ambulation tasks in paraplegic persons in the research environment. A laboratory stimulation system has been developed permitting easy definition of stimulation control sequences for two antagonist muscles. The efficacy of the system has been demonstrated, and certain issues of controlling knee angle of motor-complete paraplegic persons have been preliminarily assessed. Future studies using the system will focus on feedback-control of knee angle. Experience with this system is expected to provide design insights for a clinical programming system to support our implantable 8-channel stimulator.

INTRODUCTION

Electrical stimulation of paralyzed and paretic lower extremities can restore standing and ambulation in the research environment (1,2,3, 4,5). There are various electrode systems for activation of paralyzed muscles, but common to all systems is a requirement to program stimulation sequences to achieve coordinated functional movements. The stimulus intensities and timing for each muscle must also be coordinated with any existing voluntary motor control. To evaluate various issues of controlling stimulated skeletomusculature, we have developed a laboratory stimulation system to regulate leg motion about the knee under controlled laboratory conditions.

Our computer-controlled stimulation system provides up to 32 channels of stimulation (6,7), but only 2 are used in this study to control the antagonist muscles about the knee. Interactive graphics enable the user to define any commanded knee angle trajectory and to define the stimulation sequences for each channel. Stimulation can cycle repeatedly and the resultant average knee trajectory can be displayed.

Various issues of the stimulation system and of stimulated muscle have been evaluated: the efficacy of the control sequence selection procedure; system response repeatability; and changes in muscle recruitment and dynamic properties. Furthermore, this laboratory system has provided ideas and specifications for the design and development of a clinical programming system to support our implantable 8-channel RF-coupled stimulators (8,9) for the restoration of ambulation tasks in paraplegic persons.

SYSTEM DESCRIPTION

A Digital Equipment Corporation MINC PDP 11/23 laboratory computer forms the core of our stimulation system. The computer controls our Z80-based host-slave stimulators, communicates with a Modgraph graphics terminal, receives input from a Houston Instrument Hipad Digitizer, and samples knee angle signals in real-time.

The Z80-based host-slave stimulators generate regulated current rectangular pulses to bipolar electrodes (7). The output stages utilize latches and timers to define amplitudes and pulse durations for monophasic or biphasic pulses. A complete stimulation pulse is initiated by a single command pulse from the host processor. Monophasic and biphasic waveforms of up to $\pm$ 120 mA (59 uA/bit resolution), and 33 mS PW (0.5 uS/bit resolution) can be generated.

The user can define the commanded knee angle trajectories and the stimulation amplitude envelopes for both channels using the Modgraph terminal and the digitizing pad. For both the commanded trajectory and the stimulation sequences (amplitude envelopes), a mouse is used to "sketch" the desired waveform. Once a stimulation sequence is defined, it can be redrawn, scaled up or down, or shifted in time.

METHODS

Six motor-complete paraplegic subjects (injury levels T6-T12), with no obstructive spastic tone, participated in our study. To evaluate their degrees of spastic tone, we assessed their muscle responses to quick stretch, assessed their straight-leg-raising angle while supine, and utilized the Pendulum Test (10) before and after the first two sessions.

The subjects were seated in a special test chair that positioned both the hip and the knee in 45 degrees of flexion at rest. This rest position defined the zero knee angle. The lower leg swung freely and an electogoniometer transduced the knee angle. Cutaneous stimulation of the quadriceps and the hamstrings muscles was used to achieve two trajectories, extension-flexion and extension-only. For the extension-flexion trajectory, the leg extended to +30 degrees, flexed to -30 degrees, then came to rest at 0 degrees in a smooth fashion within a 2 second period. For the extension-only trajectory, the leg extended slowly to +30 degrees, then returned to 0 degrees within a 3 second period.

Symmetrical-biphasic stimulation pulses at 30 Hz and 300 uS PW were used to activate the muscles. Amplitude modulation was used to regulate muscle activation.

Bipolar carbon-rubber electrodes of 2 sizes were used. The quadriceps electrodes were both large (4 cm x 9 cm), and the hamstrings electrodes were either large and small (4 cm x 5 cm) or both small. Quadriceps motor points were found to be broad and required no special attention to placement; however, the hamstrings motor points were determined each time by probing (11).

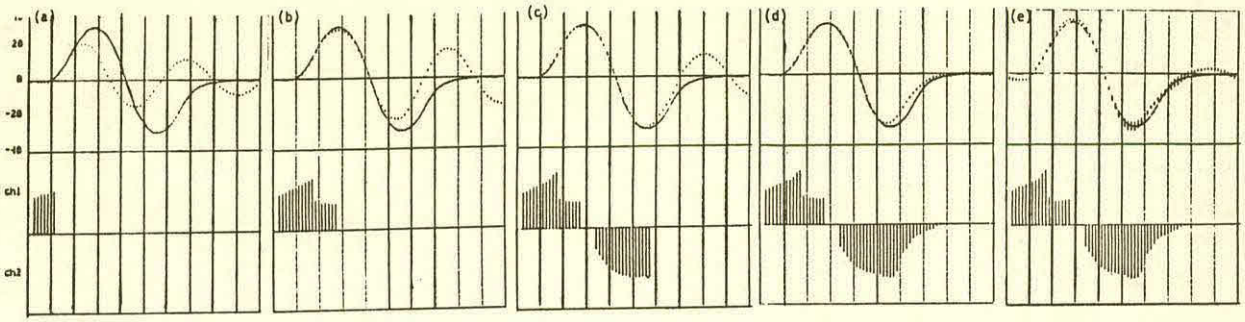

FIGURE 1: Five selected iterations illustrating the control sequence tuning procedure. The commanded and actual responses are indicated by the solid and broken lines, respectively. The control sequences were modified to match the commanded trajectory. Iterations: a) 2nd; b) 8th; c) 15th; d) 19th; e) average of 10 cycles. The lower half of the figure shows the stimulus amplitudes for the quadriceps (top) and the hamstrings (bottom).

The amplitudes were sketched and were updated, or tuned, in a trial-and-error fashion to minimize the error between the actual and the commanded responses. The average absolute error was calculated and used as a measure of the match between the actual and commanded responses. The control sequences were developed first by matching the beginning part of the commanded trajectory and iteratively improving the match over latter parts of the trajectory (Figure 1). Once good matches were achieved, 5 single cycle responses were repeated every minute. Three multiple responses (at least 10 continuous cycles) were repeated every 2 minutes.

RESULTS

The efficacy of the tuning method, and response repeatability were both demonstrated. The average number of iterations to achieve good matches (less than 3 degrees of average error) were 20 and 13 for the extension-flexion and the extension-only trajectories, respectively. Single cycles were generally highly repeatable (Figure 2). Multiple cycles were repeatable within each trial, as evidenced by the small standard deviation bars, but showed large variations between sets (Figure 3).

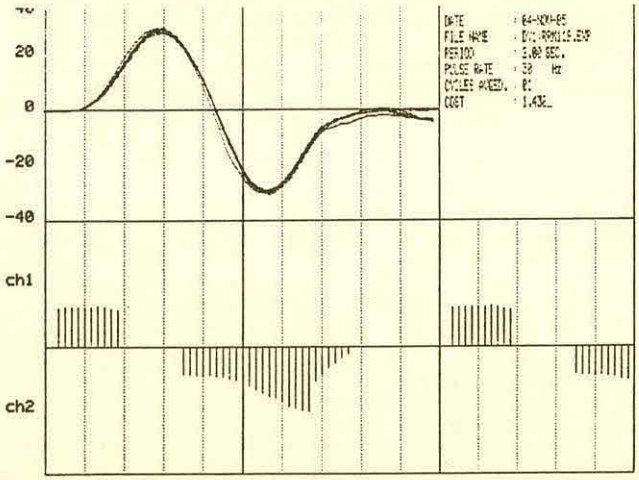

FIGURE 2: Five consecutively repeated responses of the extension-flexion trajectory for one subject. The minimum and maximum amplitude axis values for the quadriceps and hamstrings were 30 and 90 mA, and 30 and 100 mA, respectively.

DISCUSSION

The laboratory stimulation system with interactive graphics permitted easy definition and modification of the stimulation control sequences. The trial-and-error method of tuning these control sequences permitted good matches between the actual and the commanded trajectories within 30 to 60 minutes. These times can be reduced by one-third with a faster graphics system.

An interesting phenomenon, observed in all subjects, is shown in Figure 3. The average responses of the multiple cycle runs decrease with time, yet are repeatable within each trial. Perhaps this indicates that metabolic by-products of muscle activity continue to accumulate even after muscle action ceases.

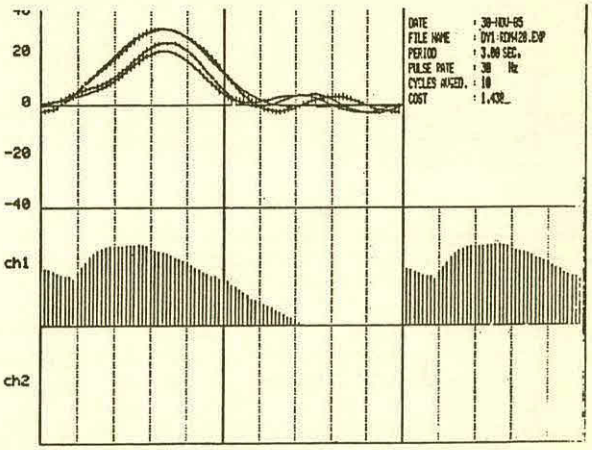

FIGURE 3: The average responses of 3 sets of 10 continuous cycles of the extension-only trajectory for one subject. Vertical lines represent 1 standard deviation. The minimum and maximum amplitude axis values for the quadriceps were 40 and 100 mA.

Response repeatability, and thus, the tuning method can be affected by disturbances to muscle input-output properties, most notably fatigue in the unconditioned muscle. Three reasons account for high rates of fatigue in electrically stimulated muscle: 1) disuse atrophy results in low oxidative capacities of the muscle fibers; 2) the synchronous activation of fibers requires functional stimulation frequencies higher than naturally occurs in the intact neuromuscular system (12); and 3) the large, strong, easily-fatigued muscle fibers can be electrically recruited before the small, weak, fatigue-resistant fibers (13). Electrical conditioning of atrophied muscle improves the contractile strength and oxidative capacities of the fibers (14). Moreover, the dynamic properties change requiring lower stimulation frequencies for fused contractions (14). For these reasons, electrically conditioned muscle would be expected to provide a more stable system to control than unconditioned muscle.

The effects of conditioning on control problems and muscle properties are being assessed. The areas of evaluation include: the control sequence tuning procedure; system response repeatability; muscle recruitment changes; and muscle dynamic changes.

To compensate automatically for disturbances to muscle input-output properties, feedback control schemes will be investigated to regulate knee angle during the swing phase of gait.

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TRACKING PERFORMANCE OF PATIENTS CONTROLLING FES HAND GRASP ORTHOSES

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ABSTRACT

Pursuit tracking tasks are used to elaluate quadriplegics' ability to control hand grasp orthoses employing functional electrical stimulation. Tracking tasks are performed in both force and position.

INTRODUCTION

Quadriplegics have been using functional electrical stimulation (FES) hand grasp orthoses for over a decade. Some of these systems are used regularly by the patient to assist in activities of daily living. The objective of this research is to evaluate quantitatively how quadriplegics operate these orthoses.

The hand grasp system employs electrical stimulation to activate paralyzed muscle. The patient uses a proportional controller to modulate the stimulus parameters to provide graded muscle activation appropriate for a given task. Proportional control, in the patients we have studied, is achieved by transduction of shoulder movement.[4]

Normal voluntary movements are controlled by cutaneous, proprioceptive, and visual feedback. As movements are practiced, motor programs are developed and refined through sensory feedback, with the new motor program stored in long term memory. In parallel with the refinement, there is less need to rely on sensory cues and the movements require less attention. In the intact hand, programs are associated with hand muscle activation and can take advantage of cutaneous and proprioceptive feedback directly involved in the task.[1]

In contrast, the patient using the hand grasp system must develop motor programs for shoulder movements that provide command signals to the stimulator. Direct sensory feedback concerning shoulder movement is available from proprioception at the shoulder. Sensory feedback about hand function is only available by visual monitoring of hand movements. Thus, refinement of the shoulder motor programs requires the patient to extrapolate the relationship between shoulder movement and hand function.

The first step in our evaluation is to test the patient's ability to perform a given task accurately. Next, learning of motor programs at the shoulder to produce hand movement is tested. Finally, the ability to use the experience gained from the practice of one task to generate a new motor program for a similar task (i.e. transfer) is tested. The tasks chosen for this evaluation are tracking tasks because they have been established as good measures of manual control and lend themselves to quantitative evaluation.

METHODS

To date, the subjects involved in these experiments are two quadriplegic patients and four persons with normal muscle function (normals). The normals were included to form a data base of tracking performance to compare to quadriplegic tracking performance in similar tasks. Neither of the patients involved in this study were regular users of the orthotic system at the time of their involvement.

Pursuit tracking tasks were used to evaluate movements in both normals and patients. The objective of a pursuit tracking task is for the subject to match his/her response to the target track that the experimenter controls.[1] The response is the output of a force or position transducer. In the case of the patients, the response is produced by their personal hand grasp system (altered so that there is no time delay between command generation and stimulation). Normals use their physiologic system (intact hand) to produce a response. The track is generated by a PDP-11 computer system in real-time. The same computer system also performs the data collection. The track and response can be displayed on a color monitor situated approximately at the subject's eye level. (Figure 1)

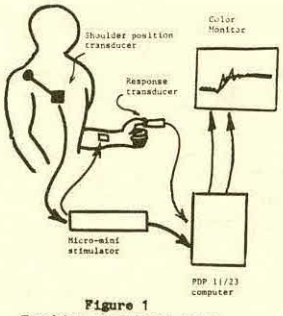

Figure 1
Tracking experiment setup

Initial experiments were performed to test the ability of patients and normals to perform a tracking task accurately, i.e. with a minimum difference between the track and the response. These tests were all done with visual feedback provided to the subject during the tracking task.

Subsequent experiments were performed to test the ability of subjects to learn motor programs and transfer the motor programs from one task to another similiar task. In these experiments subjects were trained on one track (track A) and tested on a similar track (track B). Then the tracks were reversed and track B was trained and track A was tested. During the experiment and before training, both tracks were tested with and without visual feedback to determine the initial level of tracking performance.

Training consisted of three trials with visual feedback to orient the subject to the track. Next, visual feedback during the trial was removed so that the learned motor program, unaltered by sensory adjustments would control the response. After each trial, the complete track as well as actual performance were displayed so that the subject could make corrections to his motor program on the next trial. The exact number of trainning trials performed was determined subjectively by the experimenter based on tracking accuracy. However, there were always more than ten and less then twenty, and the number of training trials for track A was equal to that for track B.

Finally, testing was done without visual feedback during or after the trials. Visual feedback was not given so that no corrections to the motor program would be made. Testing was performed first on track B then on track A. Again a number of testing trials on each track were performed.

A number of different tracking tasks have been tested thus far. Step tracks represent the worst case for control, since they are impossible to track precisely. Ramp tracks simulate more natural movements. Bell shaped tracks simulate grasping and releasing an object. Tracking trials ranged in length from twenty to sixty seconds.

When the experimental subject is a patient the following data are collected: track, response, and the input command to the hand grasp system. When the subject is a normal only the track and response are collected since the input command is not accessible.

The accuracy of tracking is determined in two ways. First, average error (AE) is computed as the mean absolute value of the track minus the response over the duration of the trial. Average error gives an overall measure of accuracy. Second, accuracy as a function of time or type of track is determined from plots of response and track vs. time. An example can be seen in Figure 2. The straight line represents the track and the other line represents the response. The response versus command plot (Figure 2) shows the muscle input/output relationship during the test. Rise time and overshoot calculations are performed on step tracks. Rise time is calculated as the time the response takes to go from twenty to eighty percent of the step value. Overshoot is the amount that the response exceeds the track, divided by the amplitude of the track, expressed in percent.

RESULTS

Patients

When tracking tasks are performed with simultaneous visual feedback it is found that patients can track accurately. The average error is low and the response achieves the target in most cases. However, Figure 2 shows that the resonpse does not reach a steady state, even though the step duration is five seconds. The patient is continuously correcting the response. Rise time is slow for low forces and there is not much overshoot in response or command. At higher forces the overshoot is larger, but the rise time is faster.

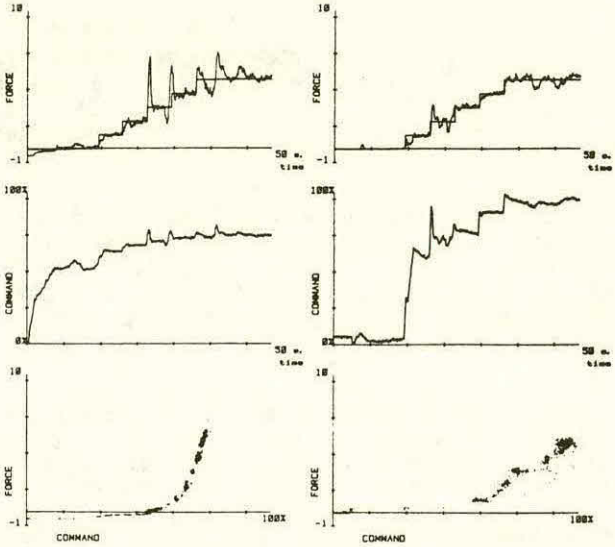

Figure 2
Patient tracking trial with visual feedback. Force is measured in newtons.
A.E.=.3580 newtons

Figure 3
Patient tracking trial with visual feedback. Force is measured in newtons.
A.E.=.1997 newtons

Overshoots in response are accompanied by overshoots in command.

Figure 3 is an identical trial to Figure 2, performed by the same patient during the same experiment about an hour later. There are not as many overshoots, but rise times are still fast. This is due to the gain of the hand grasp system. In Figure 2 the slope of the force versus command curve is steep indicating a high gain. The slope in Figure 3 is not as steep. The gain of the system has decreasesd due to muscle fatigue. Average error is lower in Figure 3 than in Figure 2 indicating more accurate tracking. Tracking is improved by the lower system gain. In Figure 2 a small change in command results in a large change in response (force), but in Figure 3 a larger command change is necessary to produce the same change in force. The patient has better control of force due to finer command resolution.

Although, there is a decrease in the average error as seen in Figures 2 and 3 this decrease is not consistant. A consistent decrease in the average error as more experience is gained with a tracking task would indicate that the subject is learning. However, this decrease is probably due to improved control and not to learning.

No evidence of learning is found when tracking tasks are performed without visual feedback either. however, patients do not make many corrections to their response. Figure 4 shows that the command the patient generates is steady, i.e. command maintains a constant level for a period of time. The response over the same period of time is not steady. It increases due to potentiation. Potentiation causes additional force to be generated by the muscle as the time of stimulation increases, although the stimulus input remains constant, i.e. constant command.

Normals

Figure 5 shows that tracking with visual feedback during the trial is accurate, but corrections are

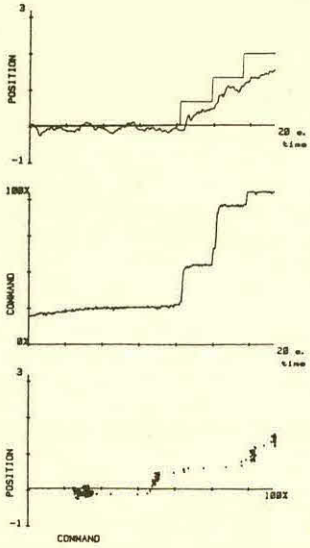

Figure 4
Patient tracking trial without visual feedback. Position is measured in centimeters.
A.E.=.3795 centimeters

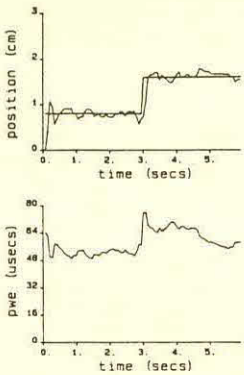

Figure 7
Step response of closed-loop controller using the extensor pollicis brevis muscle.
Taken from [5].

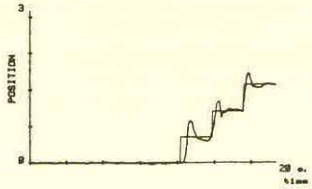

Figure 5
Normal tracking trial with visual feedback. Position is measured in centimeters.
A.E.=.0778 centimeters

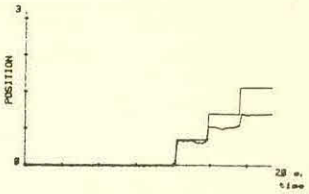

Figure 6
Normal tracking trial without visual feedback. Position is measured in centimeters.
A.E.=.2225 centimeters

constantly being made. Unlike the patients the normals seem to reach a steady state. Large overshoots occuring after the beginning of the step are present. Rise times are rapid.

Figure 6 shows tracking by a normal when visual feedback is not available during the trial. The response is steadier than the response in figure 5. It reaches a steady state almost immediately. There is no overshoot, although rise times are still rapid.

Like the patients normals do not exhibit any evidence of learning with or without visual feedback. Average error does not decrease consistantly, and there is no significant difference in the average errors with or without visual feedback, although the average error is larger in Figure 6 than in Figure 5. This increase in average error without visual feedback does not occur over the entire sample of tracking trials.

DISCUSSION

When average errors of patients and normals are compared there is little difference. This indicates that patients can operate their hand grasp systems as accurately as normals can operate their intact hands when each is provided with the same visual feedback. The gain of the hand grasp system has an effect on the tracking performance, but the patient can adjust his motor program to overcome this effect and track accurately. With visual feedback patients and normals make successive approximations to maintain the target.

The results of this study show no evidence of the learning of motor programs by either normals or patients. Evidence of learning would be represented by a decrease in average error as the amount of training increased.

Although learning was not evident in the patients they did demonstrate an ability to produce steady commands with their shoulders. However due to the open-loop properties of the hand grasp system the response produced by these commands varies. Since open-loop systems map command to stimulus input, internal disturbances cause a time varying response. This makes it hard for the patient to control hand movements resulting from shoulder movements without visual feedback. A control scheme that maps command directly to response (force/position) would improve the hand grasp system repeatibility resulting in improved hand control. Closed-loop control systems being tested in this laboratory provide direct command response mapping. Figure 7 shows the step response of one of these control systems. Pulse width (stimulus input) varies greatly immediately following the step, but the position remains at the target level. By varying the stimulus input the closed-loop control system compensates for internal disturbances. This compensation would reduce the patient's need to continuously monitor output and adjust command.

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CONTROL OF GRASP BY FORCE AND POSITION FEEDBACK

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ABSTRACT

A combination of force and position feedback is used to regulate grasp during functional neuromuscular stimulation. Antagonistic muscles are cocontracted over part of the operating range to provide continuity of control and resistance to external disturbances. The cocontraction is established by a predetermined mapping scheme relating controller output to the pulse width of the stimulus applied to each muscle. Operation in unloaded position regulation and in loaded force regulation is demonstrated in the hands of quadriplegic patients.

INTRODUCTION

Functional neuromuscular stimulation (FNS) systems have been developed for restoration of hand function in patients with quadriplegia due to spinal cord injury at the C5 [1] and C6 [2] level. Two types of hand grasp have been provided by electrical stimulation of finger and thumb flexors and extensors: lateral pinch (key grip) and tip pinch /palmar prehension. Feedback regulation of hand function is desirable because of both physiological and technological limitations of controlling electrically stimulated muscles. First, the relationship between the stimulus parameters and the force and/or position output is highly nonlinear. Second, there are significant time dependencies of muscle activation and force production. In the present open-loop systems that are used clinically, the user must compensate for errors by adjusting the command signal while visually monitoring performance. Although some correction for nonlinearities can be made on a predictive basis, the time dependencies have not been sufficiently characterized so that this is possible. In principle, feedback control could automatically regulate the overall input/output properties of the hand system, providing linearity and correction for time-varying properties [3,4]. Linearity and repeatability would improve performance by making it easy to learn the system behavior, thus making it possible to benefit from previous experience.

FEEDBACK CONTROLLER FOR HAND GRASP

For automatic regulation of the input-output properties of grasp, two physical parameters are of central importance: the opening of the hand and the forces exerted on the grasped object. A feedback control system employing a combination of force and position feedback (Fig. 1) can provide regulation of grasp under a wide range of mechanical loading conditions. The regulated property is stiffness (K), the ratio of force change to position change, rather than force or position alone. Example relationships between the command (C), force (F), and position (P) for isometric, unloaded and compliantly loaded situations are graphed in Figure 2.

The control system has two features that are important for good control. First, the properties of the load control the relative contributions of force and position feedback. Thus, the controller need not decide whether to operate in a position or a force feedback mode. Second, only a single command signal is needed to control force and position. In the unloaded condition, the command directly controls position, and in rigid loading, the command directly controls force. Both parameters will vary with the command when the load is compliant.

Two antagonist muscles are used to control movement at a single joint. Each muscle is controlled by a train of rectangular stimulus current pulses of constant amplitude. The duration of individual pulses are varied to modulate the strength of the contraction. Results of single muscle activation studies demonstrated that regulation could be achieved with passive antagonistic loading. Thus, theoretically, there would not be any need to activate more than one muscle at a time to control the full range of joint motion. However, coactivation of antagonists would still be desirable for two reasons: elimination of control deadbands secondary to joint angle dependence of muscle threshold, and rejection of dynamic external disturbances.

METHODS

Experimental tests of the stiffness control system were carried out in quadriplegic patients that had received the clinically deployed hand grasp system. The muscles that were stimulated in the controller tests were selected from among those that already had percutaneous electrodes implanted. Thus, muscle pairs were not always true antagonists. The hand, forearm and elbow were placed in a padded fiberglass cast and secured to a table at a comfortable height. The patient was seated in his own wheelchair, next to the table. The joints of the finger or thumb were splinted to reduce movement to the carpo-metacarpal joint of the thumb and the MP joint of the finger. Position was measured as spatial displacement of the end of the digit. Force was measured by a stationary, rigid transducer placed under the pad of the digit. The muscle activation controller was first order with a pole at one to ensure zero steady state error for step inputs, and a zero at 0.6 (z-domain). The gain was adjusted for acceptable performance, without much concern for being optimal.

Performance was evaluated on the basis of responses to step and ramp changes in command level, stability during maintained constant commands, the smoothness of control when switching between the region of coactivation and regions of single muscle activation, and the smoothness of control when switching between isometric force and unloaded position control. The robustness of the

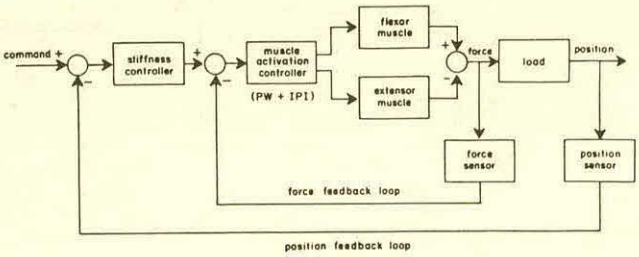

Fig. 1. Combined force and position feedback system with antagonistic muscle activation.

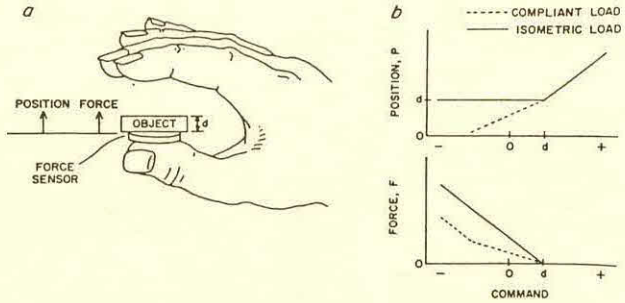

Fig. 2. (a) Configuration of hand for tests, and for definition of measures. (b) Relationships for force and position for different loads.

control schemes was not evaluated in these studies- the goal was to investigate possible control schemes and tuning procedures.

RESULTS

Coactivation of Antagonist Muscles

Coactivation of antagonists was achieved by a fixed scheme of mapping the output of the controller to the pulse widths of the antagonistic muscles, as illustrated in Figure 3. The mapping scheme accomplished four objectives: 1) saturation of PW at some maximal value to prevent electrode damage or excitation of neighboring (undesired) muscles; 2) avoidance of using pulse widths that are known to be below threshold to avoid charge injection when it is not effective; 3) control of the degree of coactivation; and 4) establishment of different gains for the regions of single muscle extension, coactivation, and single muscle flexion.

To simplify setup and to eliminate redundancy, the values of controller output at which the switch between coactivation and single muscle activation takes place are arbitrarily set to +/- a constant. This reduces the number of independent parameters to: controller gain, and for each muscle, threshold pulse width, pulse width at coactivation limit, maximal pulse width, and the ratio of slopes for single muscle activation and coactivation.

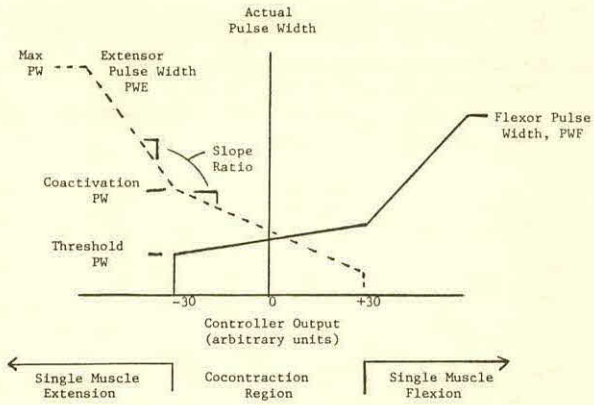

Fig. 3. Controller output to pulse width mapping scheme for coactivation.

The threshold settings for each muscle are set slightly below the lowest value of threshold actually measured over the full range of motion of the joint. This prevents oscillatory behavior that could occur as the controller output passes the estimated threshold. The coactivation pulse widths are initially selected to be the values needed to achieve nearly full range of motion without coactivation. These values are then modified to take in the following criteria: 1)they should be high enough to allow a minimum value of stiffness for rejection of dynamic disturbances; 2) they should be high enough to guarantee that they would not be reached unless the position gain is low as a result of loading (anatomical limits or external loads); 3) they should be large enough to prevent control deadbands due to underestimation or position sensitivity of thresholds; and 4) they should be low to reduce muscle fatigue as a result of wasted activation (generation of forces that are used only to oppose active antagonists). The maximum pulse widths are set in accordance with the usual criteria of safety and spillover of excitation to unwanted muscles.

The system is first tuned in the region of coactivation, and then the ratio of slopes is adjusted to get good performance in each region of single muscle activation. Each ratio is selected to permit the same controller gain range obtained during coactivation.

Control of Joint Position

Three studies demonstrated that this tuning scheme does give stable position regulation. Thumb extension (with gravitational loading in the direction of flexion) were controlled with coactivation of the EPL/EPB and ABP muscles of one patient and the EPB and ABP muscles of a second patient. Results from the third study are presented and results from the first two studies were in good agreement.

Unloaded position control during truncated ramp commands of 1.5 cm at three velocities is

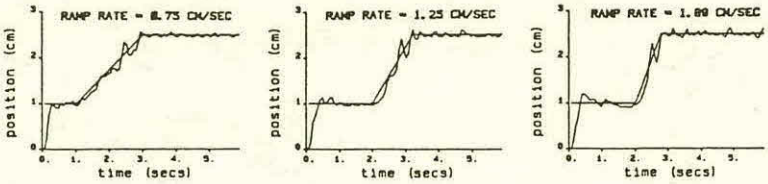

Fig. 4. Truncated ramp responses of isotonic position regulation using coactivation (ABP and EPB muscles; controller gain = 0.04; controller zero = 0.6; command = 2.5 cm; stimulus period = 70 ms).

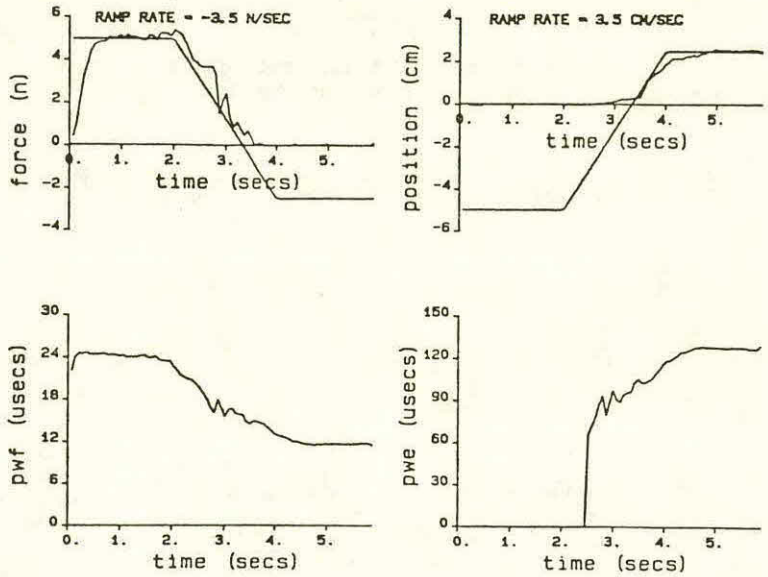

Fig 5. Truncated ramp responses showing the transition from isometric force regulation to unloaded position regulation using coactivation. (Same parameters as in Fig. 4; controller stiffness = 1 N/cm).

illustrated in Fig. 4. Control is stable in the steady state, and the overshoot is negligible. Throughout the experiment, coactivation of the antagonist was verified by repeating trials with the flexor stimulator cable unplugged.

Control of Joint Stiffness

The transition between isometric force control and unloaded position control is illustrated in Fig. 5 during the same experiment as the position control shown above. In this case, the command was set initially for a position of -5 cm, followed 2 s later by a ramp to 2.5 cm at a rate of 3.5 cm/s. Since zero position was defined as the position with the thumb resting on the force transducer with positive position in the direction of extension, a negative position was impossible, and a force in the direction of flexion was produced initially. The magnitude of the force was defined by the regulated stiffness (1 N/cm) and hence, a flexion force of 5 N was produced. As the command ramped up, the force decreased to zero and the thumb began to extend. There was some overlap between force and position control near the zero command level, due to the compliance of the thumb pad on the transducer. The transition from isometric force control to unloaded position control was smooth, as was the transition from single muscle activation to coactivation.

CONCLUSIONS

The success of our efforts to date indicates the feasibility of the proposed control systems. The performance of the system is still being studied, with particular regard to the robustness of the performance, and with the aim of incorporating wearable transducers.

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PREDICTION AND EVALUATION OF ANKLE MOVEMENT WITH IMPLANTABLE PERONEAL STIMULATORS

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ABSTRACT

A quantitative method for the prediction and evaluation of ankle movement with implantable peroneal stimulators was investigated. The method consists of a determination of desired EMG responses of the tibialis anterior, peroneus longus and soleus muscles to single surface stimulation pulses as well as gait dynamics with and without stimulation prior to implantation. The EMG records were then used as a guideline when positioning the electrodes during surgery. Gait parameters and EMG activity of subcutaneously stimulated muscles were measured and compared with the results obtained before implantation. Eleven hemiparetic patients with previously implanted stimulators were evaluated and predictions were done for one clinically completely paraplegic patient.

INTRODUCTION

In the last few years in Ljubljana over 30 hemiparetic patients have had the peroneal underknee electrical stimulator implanted (1). The electrodes are attached to the nerve during a minor surgical procedure under local anesthesia with the patient lying on the hip of uninvolved side. The electrode position was selected by moving the implant along the exposed common peroneal nerve close to the lateral head of the fibula under the knee. Clinical observation of the stimulated ankle movement during surgery did not prove to be a satisfactory criterion for positioning. Clinical evaluations of patients with implants has shown that in several cases the quality of gait correction is not satisfactory (2). Although in certain cases gait correction is still functional, an excessive eversion can be observed. Similar results were reported in (3).

Improvement of the quality of gait control by implantible peroneal stimulators motivated us to enhance the methodology of implantation and checking implant position over time. The enhancement consists of complementing the qualitative evaluation of foot movements with the monitoring of EMG responses of the tibialis anterior, peroneus longus and soleus muscles caused by stimulation as a function of electrode position of the implant during surgery.

METHODS

The method of ankle movement prediction can be divided into three phases:

1. Preimplantation phase. Responses to single pulse stimulation of the common peroneal nerve were monitored in the ankle dorsal flexors, evertors and plantar flexors in patients while standing, sitting and lying down. Single pulse surface stimulation of 1 Hz and 0.2 ms duration was used on standing patients in order to obtain a strong activation in the tibialis anterior and a moderate one in the peroneus longus. Triceps surae was monitored to exclude the unwanted plantar flexion. The responses were then monitored with the patient sitting and lying on the uninvolved side, as required by the surgical procedure. Besides the single pulses, a 30 Hz train of pulses with the same amplitude and duration was applied during gait in order to estimate ankle dynamics with the electrodes in the same position. Gait dynamics without stimulation at the same gait velocity was also measured.

2. Implantation phase. The records of EMG while lying down from the previous phase were used during surgery as a guideline for determining implant position. Muscle activation was monitored during surgery while moving the electrodes along the nerve and the responses compared to this EMG. The responses were tested again after the wound was covered and after the skin sutures were removed in order to check for eventual displacements.

3. Postimplantation phase. Two to three weeks after surgery, quantitative gait evaluation with subcutaneous stimulation and EMG activity of stimulated muscles groups were measured and the results compared with those obtained before implantation. This phase is used to follow up on a patient's gait quality over time.

INSTRUMENTATION

Recording was done by three pairs of In Vivo Metric Ag-AgCl surface EMG electrodes, placed 2,5 cm apart over the muscle bellies, and a wraparound ground electrode between the stimulation and detection side. EMG signals were amplified by three Tektronix AM502 differential amplifiers, displayed on a Tektronix 7633 oscilloscope with two 7A18 dual trace amplifiers, and recorded by a Tektronix C-5C oscilloscope camera. The oscilloscope was externally triggered by the stimulation pulses. Ankle dynamics were estimated by the information obtained from ground reaction measuring shoes (5). The surface stimulation was

provided by a constant current stimulator with continuously regulated amplitudes from 0 to 50 mA, frequencies from 0.7 to 50 Hz and duration of rectangular monophasic pulses from 0.05 to 0.5 ms. Monopolar wet felt pad electrodes with the neutral 10 by 5 cm rectangular anode in the fossa politea and the active 2.5 cm in diameter round cathode at the lateral head of the fibula were set at 1 Hz and 0.2 ms while the amplitude was adjusted for the desired response. Subcutaneous stimulation was done by an implanted receiver with bipolar platinum electrodes touching the nerve close to the lateral head of the fibula (1).

RESULTS AND DICUSSION

Eleven hemiparetic patients with previously implanted stimulators were tested by the described method. Different combinations of M waves in the tibialis anterior, peroneus longus and, to a smaller extent, in the soleus were recorded with slight H reflexes in some cases. Sometimes the responses varied considerably between standing, sitting and lying on the flank for the same patient. There was no general rule as to whether the responses while sitting or lying were closer to the standing ones. The combination of M waves in the observed muscles is fairly correlated with the direction of the isotonic ankle movement. Three of the 11 patients showed good dorsal flexion with moderate eversion. Two other patients had similar responses when the implant was pressed against the nerve. The remaining 6 patients showed various stages of excessive eversion with or without dorsal flexion as well as plantar flexion in one case. Ground reaction force, ankle torques as well as clinical gait observations were used for gait evaluation. There are good correlations between the EMG muscle responses and ground reaction forces measured during walking with subcutaneous stimulation (the conclustions were similar for 90% of the observed population). Illustrative examples are shown in Figs. 1, 2 and 3. In Fig. 1 the M waves of four superimposed respones in the tibialis anterior (top), peroneus longus (middle) and soleus (bottom) are presented for a dex hemiplegic patient with dorsal flexion and moderate eversion during stimulation with implanted electrodes in the stance phase. A low amplitude H reflex of the tibialis anterior can be observed. In Fig. 2 average vertical ground reaction force F and its point of action along the X,Y axes with and without stimulation of the same patient are shown. It is obvious that the transfer of body weight onto the impaired leg is raised by 30% and stride time is shortened by 40% due to stimulation. The trajectory of the point of action of the impaired leg is shifted from the lateral side (excesive inver-

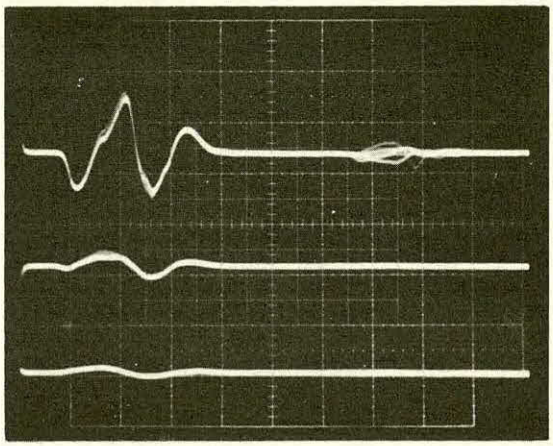

Fig.1 EMG responses

sion) to the middle of the shoe sole, similar to the trajectory of the unimpaired leg. Using this data and joint angle functions, ankle torques around the X and Y axes were computed and are shown in Fig. 3. There is a significant change of the Mx ankle torque of the impaired leg from dictinctive inversion (15 Nm in the push off phase) to slide eversion (-5Nm at the beginning of the stance). This can be explained by the activity of the peroneus longus during stimulation in the swing phase. Also, there is a significant change in the My torque at the beginning of the stance phase, which is a consequence of prolonged activity of the tibialis anterior. Normal leg torques do not show significant changes due to a large variation of the measured data, the only exception being the My trajectory in the push off phase.

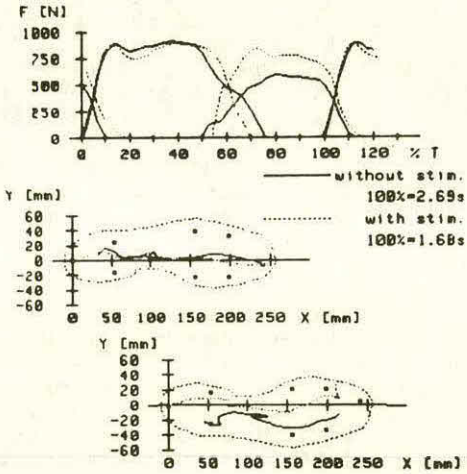

Fig.2 Ground reaction

EMG monitoring was applied during the implantation of a peroneal underknee stimulator in a clinically complete spinal cord injury patient (Th 5) to achieve the withdrawal flexor reflex. EMG

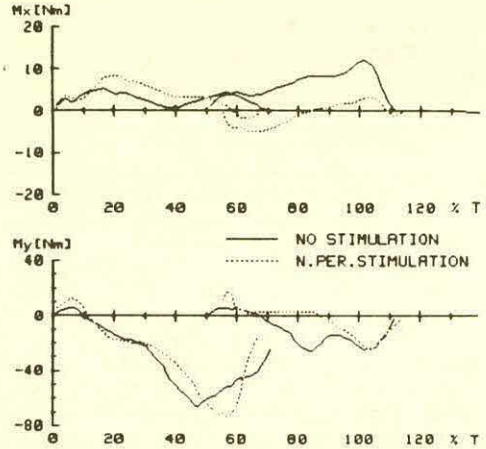

Fig.3 Ankle torques

responses of the tibialis anterior, peroneus longus and soleus muscles with strong dorsal flexion and moderate eversion in the ankle during stimulation of the common peronel nerve by surface stimulation prior to implatation were obtained. During the implantation under local xylocain anesthesia, the patient displayed only dorsal flexion in the ankle. EMG monitoring revealed a somewhat lesser activation of the peroneus longus which did not result in eversion. During tests a month after implantation, strong dorsal flexion with moderate eversion was achieved with responses practically equal to those before implantation.

Ankle torque, ground reaction force as well as kinesiologic gait evaluation and EMG response of stimulated muscles gives good information for the prediction and evaluation of ankle movement under the influence of subcutaneus electrical stimulation. The problems of different movements while standing, sitting and lying are solved by comparing the previous EMG responses of the desired stimulation while lying down with the monitored responses during surgery.

High correlation between ankle dynamics and EMG responses of stimulated muscles suggests the possibility of simplifying the methodology to EMG monitoring only.

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ELECTROMYOGRAPHIC STUDY OF THE SIDE STEP FOR DEVELOPMENT OF ELECTRICAL STIMULATION PATTERNS

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ABSTRACT

Electromyographic (EMG) activity in a normal subject during a side step were recorded and studied for the purpose of developing an electrical stimulation pattern for side step in paraplegic subjects. A preliminary pattern based on the EMG activity was developed and tested.

INTRODUCTION

Functional neuromuscular stimulation (FNS) has provided selected paraplegic subjects with the ability to stand up, walk forward and walk up and down stairs (1,2,3). However to make the FNS system functional in daily activities, the system must also be capable of providing paraplegics with the ability to step sideways and backwards, maneuvers which are important in opening doors and getting around obstacles. One of the major difficulties in developing stimulation patterns for different activities is determining which muscles to stimulate and when and by how much they must be stimulated. Considerable insight can be gained by analyzing the electromyographic activity in normal subjects performing specific tasks. In this abstract we present results of the muscle activity in a normal subject during side steps as determined by surface EMG electrodes.

METHODS

We studied the activity of the trunk, hip and leg muscles during a side step using external EMG preamplifiers placed over different muscles of a normal male subject. The EMG signals were amplified, band pass filtered (20Hz - 2KHz) and recorded on a fiber optics chart recorder (Honeywell model 1858 CRT Visicorder). Because the number of electrodes was limited, two series of tests were conducted, each using seven electrodes. The electrodes were placed over muscles which were found from preliminary testing to be the most active during side stepping. In the first series, electrodes were placed over the surface of the left and right erector spinae, the right gluteus medius, semitendinosus, vastus lateralis, medial gastrocnemius and tibialis anterior. In the second series electrodes were placed over the left and right gluteus medius, left semitendinosus, rectus femoris, vastus lateralis, medial gastrocnemius and tibialis anterior. Hip flexors were not considered since their EMG signals are not readily detected by surface mounted electrodes. Muscles included in this study were limited to those that are currently available for stimulation in the functional neuromuscular stimulation system at our laboratory. Hip and leg muscles were stimulated by intramuscular electrodes (1) while surface electrodes were used for stimulation of trunk muscles.

Foot/floor contacts from the heel, medial and lateral aspects of the feet, and force plate data were recorded on a MINC 11/23 (Digital Equipment Corp.) and synchronized with the EMG recordings using a trigger pulse. In the experiments the subject stood beside two force plates (AMTI Model OR6-5-6) and took a series of lateral steps across the plates. A number of trials were conducted, with both right and left side steps, on different days to ensure repeatability of the data.

The right step cycle was divided into four phases which were termed double support feet together (FT), right swing (RS), double support feet apart (FA) and left swing (LS). The cycle began with the subject standing in the normal standing position with the feet approximately 10 cm apart (FT). The subject then shifted his weight, and moved his right leg laterally about 60 cm (RS). At right foot contact (FA), the subject simultaneously began pushing off with the left foot into the left swing (LS) phase, which ended with left foot contact in the feet together (FT) position. From this position, the step cycle repeated. The approximate periods for each phase were 0.30 seconds for the FT phase, .45 seconds for the RS phase, .25 seconds for the FA phase, and .40 seconds for the LS phase. The timing of the whole cycle was regulated with a metronome.

To make a preliminary evaluation of the ability of paraplegic subjects to make a side step a stimulation pattern based on the EMG results was developed and programmed into a 32-channel portable stimulator. The pattern was tested on a paraplegic subject who stood with the aid of FNS and a parallel bar for balance.

RESULTS

The preliminary study to determine which muscles were active in the right step maneuver indicated that the anterior trunk muscles (rectus abdominis) and the adductors of the hip did not show any significant activity, and were therefore not considered in further studies.

In the following, results are shown for only the right side step, as we found that the results for the left side step were symmetrical. We found that the EMG signals were very repeatable from trial to trial on different days, both in terms of the muscles that were active and their timing. Figure 1 shows the raw EMG signals from two representative trials and indicates the activity of the various muscles tested. We found that the right tibialis anterior was most active just prior to RS

and during LS. It also showed some activity during the FA phase. It displayed its least activity during the RS phase. The left tibialis anterior was most active during left swing and least active during both double support phases (FT and FA).

The right medial gastrocnemius exhibited its greatest activity during loading of the right foot at the beginning of the FA phase, as well as during the LS phase. The left medial gastrocnemius was most active during the FA phase, immediately following right foot contact.

The left and right vastus lateralis showed some activity during most of the cycle. However, the left vastus lateralis was inactive during most of left swing, while the activity of the right vastus lateralis decreased during right swing. The left rectus femoris showed almost identical patterns of activity as the left vastus lateralis.

The right semitendinosus was most active at the end of RS and during loading of the right foot during the FA phase. It also showed some activity during the FT phase. Otherwise, it showed very little activity. The left semitendinosus was active only during the FA phase and, to some extent, during LS.

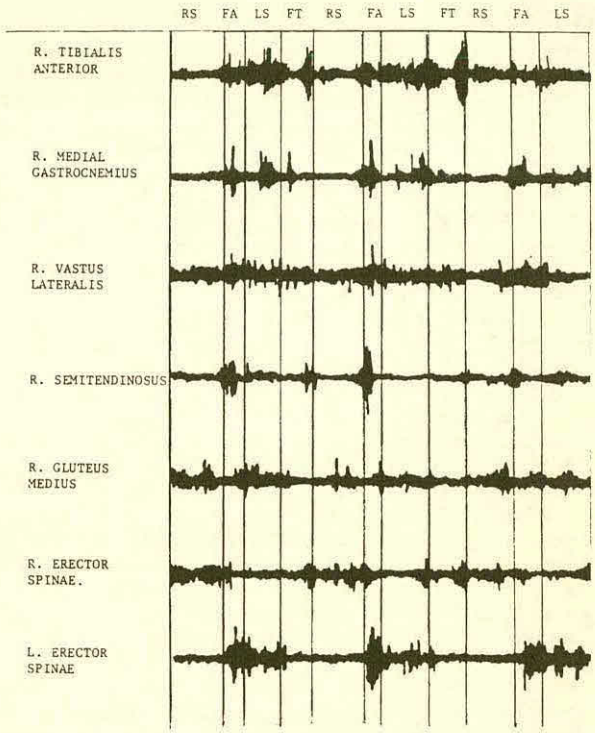
Both the left and right gluteus medius were active throughout most of the cycle. The activity of the right gluteus medius started with the RS phase and continued through the FA phase and LS, appearing to peak near the beginning of left swing. There appeared to be little activity during the FT phase. The left gluteus medius was most active just prior to RS and showed reduced activity during the RS and FA phases.

The left and right erector spinae acted out of phase with one another. While one was active, the other was generally inactive. The right erector spinae was active toward the end of the FT phase and during RS, while the left erector spinae was active during the FA phase and LS.

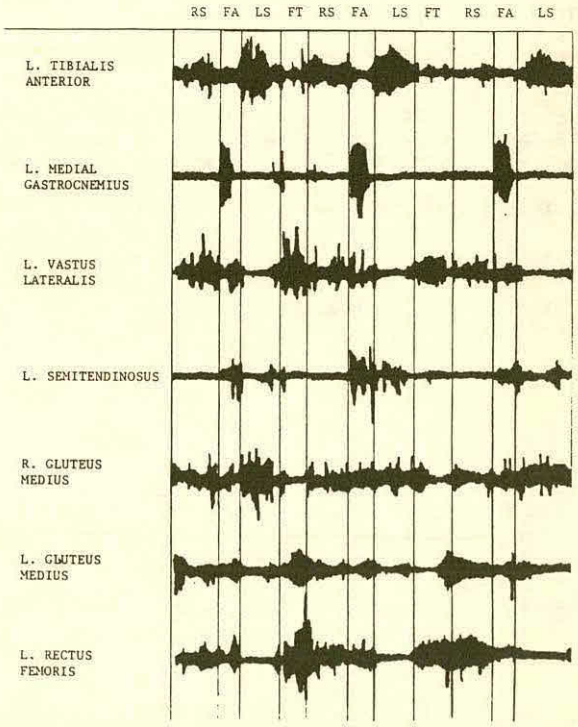
From preliminary tests with the quadratus lumborum muscles we found that these muscles were very active during most of the cycle. The right quadratus lumborum was most active during right swing and became inactive at right foot contact. The left quadratus lumborum was most active at right foot contact and just prior to right swing.

Analysis of the force plate data showed detectable patterns of weight shifting corresponding to the actions of the gastrocnemius and gluteus medius muscles.

In testing the stimulation pattern based on the above EMG results, only slight modifications were necessary to accomplish the side step in a paraplegic subject with electrical stimulation. The modifications involved bilateral stimulation of adductors during left swing, which was not observed in the normal subject.



(a)



(b)

Figure 1. Muscle EMG activity in the first series of tests (a) and second series of tests (b). Signal gains were adjusted to show activity but not relative intensity between different muscles.

DISCUSSION

We found that the muscle EMG activities during a side step are considerably different than those of forward walking. The double support periods are longer in the side step than in forward walking. Foot contact usually begins with toe contact, rather than heel contact. Just prior to right swing, the left gluteus medius, right erector spinae and right tibialis anterior appear to act together to lift the right foot from the floor. Then the left gluteus medius shifts the center of gravity while the right gluteus medius regulates the step length. Once the right foot contacts the ground, the left gastrocnemius acts to shift the center of gravity to the right leg while the activity of the right gluteus medius increases to hold the left side of the pelvis up during left swing. During right swing, the right erector spinae helps to keep the right side of the pelvis up, while during left swing the left erector spinae keeps the left side of the pelvis up. At right foot contact, the right gastrocnemius acts to soften the impact, and then the gastrocnemius and tibialis anterior both act to stabilize the right ankle.

Results of initial testing of the stimulation pattern indicate that the side step can be achieved in paraplegic subjects using electrical stimulation. The modifications necessary in the stimulation pattern indicate that more careful study of the role of the adductors in the side step may be necessary.

The results reported in this abstract, and results of further analyses of functional tasks in normals are currently being used to help improve the functional abilities of paraplegic subjects using FNS.

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ABSTRACT

Current methods of producing customized seating for handicapped individuals are time consuming and outdated, or create ineffective seats for wheelchair bound individuals. A prototype system for rapid (one-hour) manufacturing of custom contoured foam cushions was developed. Patient's buttocks are measured in a uniform pressure device which is then used as a mold cavity for the cushion molding process. Topics covered include design concepts, materials research, and viable processing techniques.

INTRODUCTION

The field of orthotics and prosthetics is a very diverse and rapidly expanding field. Current technological advancements include lightweight composite leg braces and cam-controlled exoskeletal walking devices for paraplegics. However, one area of this field that hasn't received a great deal of attention until recently is the field of customized contoured seating for wheelchair bound individuals.

There have been several isolated studies completed on different types of foam, air or gel cushion systems; however, each of these systems has drawbacks and cannot to the authors' knowledge be made in any rapid manufacturing process. Based on observations made at Newington Children's Hospital, the demand for such customized seat inserts is great.

The present method of fabrication of custom seating at Newington Children's Hospital is tedious, time consuming, and often cost prohibitive. Many of the patients who could benefit from sitting in an upright position in a wheelchair are patients in state facilities who do not have the funds for a custom fitted wheelchair.

Newington's present system makes use of a phenolic resin back support with a hand carved foam insert. The foam used is polyethylene Ethafoam. The insert is custom tailored to the individual patient with the use of a plaster cast model. The plaster cast is taken from a beanbag impression of the patient prior to the Ethafoam carving.

The advantages of the present system are that it does produce acceptable cushions and that the final insert effectively offers comfort and support to the patient. However, the disadvantages of the system in many cases outweigh the advantages. The initial plaster cast model taken from a beanbag is a primitive process that does not provide for an even pressure distribution in the final seat.

Technicians must wait for red spots to appear on the patients skin indicating high pressure areas in the chair which can causes complications. The patients may not feel the discomfort because of their paralysis, but if the redness persists the patient will develop bedsores which lead to skin cell death (necrosis) and infection.

Another disadvantage of the system is that skilled individuals are needed to hand carve, readjust, and assemble the seat. These individuals must not only have the technical skills to make the seat, but they must be familiar with the medical needs of the patient as well. Finally, one of the main disadvantages of the system is the actual manufacturing process. The handcarving sequence itself is very time consuming. This creates a long wait on the patient's part for a much needed customized seat.

Considering these factors as well as the needs of the patients who require such customized seating, it is the goal of this project to develop a rapid manufacturing system to produce custom contoured seating fixtures for multiply handicapped patients.

SYSTEM CONCEPT AND DESIGN

Research indicated the optimum system would be one in which the contour of the seated patient could be measured quickly and subsequently modified if deemed necessary by a physical therapist. The design chosen (see Figure 1) consists of an array of moveable rods and a foam-in-place (FIPS) molding system. A typical manufacturing sequence includes:

1. The patient is placed in the measurement chair. The rods are then displaced by the weight of the patient. A pressurized air bladder underneath the rods insures an even pressure distribution on each rod.
2. The rods are locked in place.

3. The patient is removed from the measurement chair.
4. If desired the physical therapist can make final adjustments on the surface contour.
5. The air bladder is removed from the lower cavity and a foam molding bag is inserted.
6. Foam is injected into the bag beneath the rods. After foam expansion the patient's contour is duplicated.
7. The foam insert is then removed from the cavity, trimmed and covered with fabric, and inserted into the wheelchair.

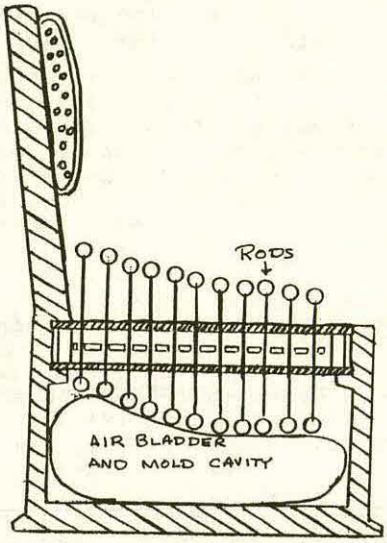

FIGURE 1

The design for this mechanism has been fully developed into the prototype stage and consists of four major components:

1. The displacement rods which are steel columns with spherical ends, placed in a one rod per square inch array to provide comfort and optimal contour reproduction.
2. The locking mechanism which is designed to instantaneously lock all the rods once the patient is in the optimal position. This mechanism can subsequently used to loosen individual rows of rods to allow for manual adjustment of the cushion contour.
3. The air bladder is the means of uniformly supporting all the rods during steps 1 and 2.
4. The most important component is the foam to be used for the seat cushion.

Through research it was decided that the optimum foam would have properties similar to Ethafoam and also capable of being foamed-in-place in a hospital environment. A major foam manufacturing corporation is presently assisting in the development of a commercial urethane formulation to satisfy these specifications.

RESULTS AND CONCLUSIONS

Small scale prototypes of each component of the mechanism have been manufactured and tested. A full-scale prototype is scheduled for completion by mid-May. Tests show that the design is functional and will be a method for rapidly producing low cost contoured seating cushions. The prototype presently under construction is limited to the buttock region. Future systems will also include a section to reproduce the contour of the upper torso. It should be noted that one rod chair system is capable of producing at least one custom contoured cushion per hour, thus greatly improving productivity.

The advantages of this design are as follows:

1. One-hour manufacturing and processing.
2. Improved cushion contour and pressure distribution.
3. Cost efficiency - foam costs are minimal (approximately $10.00 per cushion) and an estimated amatized mechanism cost of less than $10.00 per cushion.
4. Finally, this design makes seating fixtures for a wide range of applications such as the multiply handicapped, geriatrics, and various consumer groups.

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A METHODOLOGY TO ASSIST IN THE PRESCRIPTION OF WHEELCHAIRS

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ABSTRACT

A simple method has been developed to assist in the prescription of wheelchairs. The use of a Measurement and Positioning Device (MPD) enables the prescriber to physically move the wheelchair user in space to an "ideal" position and so work with the user in order to determine the most appropriate positioning. Measurements can then readily be taken. Also proposed is a complimentary system, Ffor manufacturers to use, to describe the seating and other support surfaces. This information will displayed in a standard format between all manufacturers and will match the format of the measurements obtained from the MPD.
The prescriber will thus be able to quickly match the requirements of their client across a wide range of available wheelchairs.

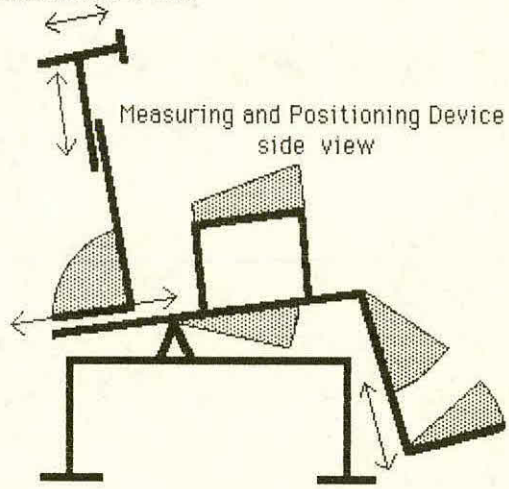
INTRODUCTION

RESNA[1], with the support of the VA[2], the PVA[3], ANSI[4] and a number of other agencies and organizations, has facilitated a series of working meetings that has enabled wheelchair users, prescribers, evaluators, designers, and manufacturers to come together and seek answers to questions related to wheelchair prescription.
In a previous article the development of Wheelchair Standards, of which this work is part, has been discussed at length[5].

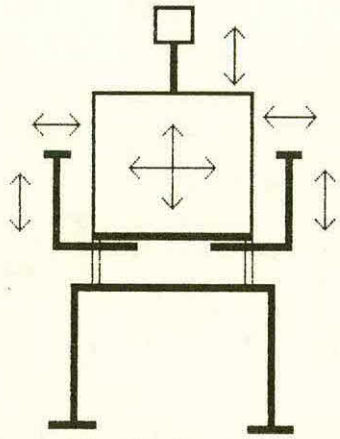
METHODS

The starting concept was the desired outcome: a wheelchair user in a subjectively good posture(s) and some associated description of desired support and fit. The criterion for this posture, support, fit and functional mobility is left to the descretion of the user and the users prescriber. A method of quantifying this posture was developed based on a simplified model of the human form supported by imaginary foot, leg, seat bottom, and seat back planes.

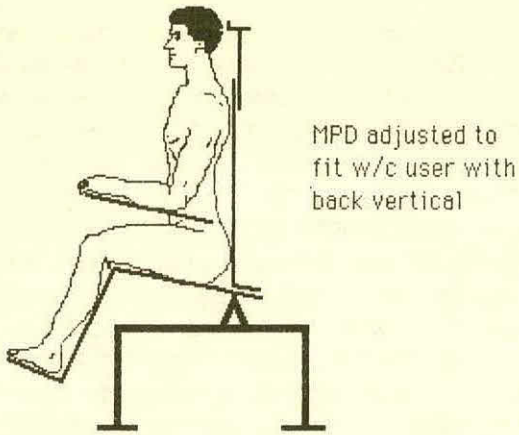
The intersection of these planes, and projection of the foot onto the foot plane , the pelvis onto the seat bottom plane, and the head and torso onto the seat back plane served to define length measurements analogous to foot length, leg length, thight length, back height, hip width, and head position. Direct anatomical measurements were minimized in this conceptual approach. A primitive plywood mock up of an adjustable device was fabricated to test this conceptual model, and based on a positive outcome of this test, McLaurin of the University of Virginia REC designed a Measurement and Positioning Device (MPD) that retained the simplicity of the support planes of the conceptual model, provided reference and adjustable support surfaces for forearm and head, and incorporated sufficient adjustments and mechanical advantages to facilitate a series of positioning and measurement trials.



Note:
Grey shaded areas represent approx. angular adjustments. Arrows represent linear adjustments.



Measuring and Positioning Device - front view



The simulation of an 'optimum' position is arrived at by means of an iterative procedure in which the 'user' is positioned on the MPD with the appropriate cushions and body support devices in place. The seat width, depth and inclination; backrest height, width and angle; armrestposition (height, width, inclination, etc.); foot and leg rest positions (angles, distances, etc.); headrest (if applicable) position (angle, placement, etc.) are adjusted to the positions which are consistent with maximum comfort (and support,if appropriate) in the subjective opinion of the 'prescriber' and user. This positioning process is based on an initial 'rough adjustment' followed by successive iterations in the various adjustments until the 'optimization' process is completed.

The optimization of ergonomic function with respect to wheelchair propulsion and handling is determined by simulation of the position of the propulsion interface. Since most wheelchairs are propelled with handrims, this determination is a function of the placement of the mainwheel axle position with respect to the shoulder axis position of the user and the respective diameters of the wheel and handrim and the effective position of the combined C.G. of the wheelchair (not including the main wheels) and user. The position of the hub axis of the propulsion interface (e.g., wheel and handrim) is located in an initial position and then successively re-positioned until an 'optimum' placement is determined.

The MPD has proved useful both as a positioning tool and as measurement reference, since the majority of the user measurements are taken directly from the adjusted MPD and not from anatomical landmarks which are hard to define and more time consuming to make. The MPD is mounted on floor scales which allow the center of gravity to be calculated based on the measured weight shift with a specific change in postural angle.

The number of measurements was reduced to the minimum required to define this subjectively good posture. Also specific proceedures were defined to address the fit of a wheelchair. For example, if the clinician desired a specific clearance behind the knee, and or clearance at the greater trochanters, the position of the user and the MPD would be adjusted to the specified clearances. Also, if a users would be using a seat bottom cushion separate from the wheelchair (or other required support or padding), the proceedure required for these to be used with the MPD (or appropriate allowances made in adjustment or measurement). Once the desired position has been obtained measurements can be recorded from the MPD in a few minutes.

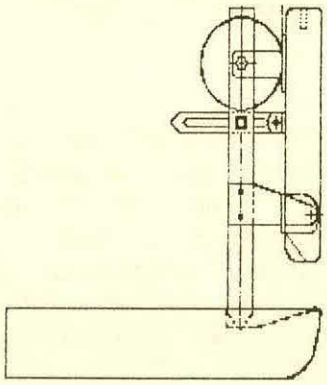
The resultant linear and angular measurements have a obvious relationship to posture, and thus a clinican can get a feel for size and posture from a tabular or graphical presentation of this information.

SELECTION OF AN APPROPRIATE WHEELCHAIR

Comparison of the measurements obtained from the MPD with those published by manufacturers, in a standard format, then rapidly enables the prescriber/user to select the appropriate make and model of wheelchair.

WHERE DO THE WHEELCHAIR MEASUREMENTS COME FROM ?

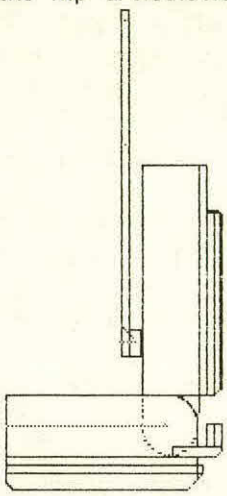
This information is obtained from the measurement of wheelchairs using two different, but interchangeable, loader gauges as reference. One is the European Loader Gauge[6] with a simplified approach to locating the footplate.



The European Loader Gauge

The second was the ISO Standard Dummy[7], minus foot weight unit. This was simply modified to

approximate the contact contours of the European loader gauge, and to provide a similar angular adjustment in the "hip" articulation.



The ISO Dummy

For wheelchairs with adjustable features, measurements proceedures are specified that either record the range of adjustment, or deviation from some normative position. The proceedure specified was determined based on the clinical significance of the measurement in question, and a desire to minimize the work required to obtain meaningful measurements, (approximately 30 minutes per chair).

The raw data from the wheelchair measurement proceedure is converted in to MPD equivalent measurements through the application of relatively straight forward algebraic formulas (developed by Brubaker and Ferguson-Pell), so that it is expressed in the most clinically significan terms. The presentation of information will be in the form of scaled and labeled bar graphs that provide for ready visual comparisons when seeking the users/wheelchair match.

DISCUSSION

The methods briefly outlined here have been developed as a result of a National and International effort to develop Standards for Wheelchairs. The purpose of these efforts **is not** to produce a standard wheelchair, but to produce standard methods of testing wheelchairs with an accompanying **standard method of disclosure of information** about the wheelchair. This standard method of disclosure will enable the meaningful comparison of many different makes and models of wheelchair. Thus potentially widening the choice of User and Prescriber.

It will not be necessary for users and prescribers to use an MPD equivalent device to benifit from the presentation of wheelchair measurements in a consistent fashion. However, the use of such devices or equivalent methods that quantitatively define subjectively good posture can have a significantly positive downstream impact on the increased availability of "better fitting" wheelchairs. Consider that each time a subjectively good posture is defined, and is used in a wheelchair purchase process, any compromise between the desired and the available is quantified. If such information is recorded over time, either by individual manufacturers or on some broader scale, then analysis of such information could be used for redesign of existing products based on statistical analysis of what the buyers want to purchase.

This paper has centered on the activities of one ad-hoc committee with a specific responsibility in the area of Seating, Positioning and Ergonomics. This subcommittee reports to the RESNA/ANSI Wheelchair Standards Committee chaired by Colin McLaurin.

ACKNOWLEDGEMENTS

The activity of producing Wheelchair Standards has been going on for about 10 years, because of this it is impossible, in the space allowed to acknowledge all the individuals who have contributed to this work. The work on Seating , Positioning and Ergonomics has been sponsored by the RESNA Technical Advisory Group to ANSI which is funded by a grant from the Veteran's Administration and by donations from wheelchair manufacturers.

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BEAD MATRIX INSERT SYSTEM - A FOLLOW-UP CLINICAL REPORT

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ABSTRACT

Custom contoured seating technology has made significant progress in the last decade. The vacuum consolidation method offers the possibility of producing customized seating components rapidly and economically. The paper reports on a follow-up clinical study involving 94 subjects who received the Bead Matrix Insert (BMI).

BACKGROUND

Clinical investigators have long been fascinated with the vacuum consolidation approach to replication of body shapes, or for use in directly providing supportive structures (1, 2). The vacuum consolidation approach uses a air-tight flexible bag filled with light weight particles (ex. 1mm diameter polystyrene beads) to which a vacuum source is attached. Once shaped to the surface of the body, evacuation of the air from within the bag causes the particles to mechanically interlock and hold their shape. Traditionally this process has been used to obtain and transfer body shapes so that durable plaster molds (replicas of the body) can be produced. The molds are then used as the substructure over which a custom contoured plastic seating component is produced (3).

Other investigators have attempted to bypass the need to produce a plaster mold by using vacuum consolidation to directly produce a seating component (4, 5). This entails mixing an adhesive resin with the bead like particles (Bead Matrix) in the air-tight bag before final shaping and evacuation of the air. The Bead Matrix Insert System (BMI) (Figures 1-3) incorporates this later concept, in addition to the following design criteria.

* A series of adjustable and independent seat, back and headrest components that can be readily interfaced into at least three sizes of standard wheelchairs.
* A resiliency of the bead/resin matrix substructure to enhance impact strength and provide a degree of cushioning.
* Permits the use of a simulator for the evaluation of patients prior to financial commitment.
* Allows rapid hand contouring to accommodate severe orthopaedic deformity.
* Allows evaluation, fabrication and final fitting of a complete seating system in less than eight working hours.
* A design philosophy that permits commercial production and marketing at retail prices that are within accepted cost ranges for custom contoured seating inserts.

The early developments of the BMI system have been previously reported (4). This paper

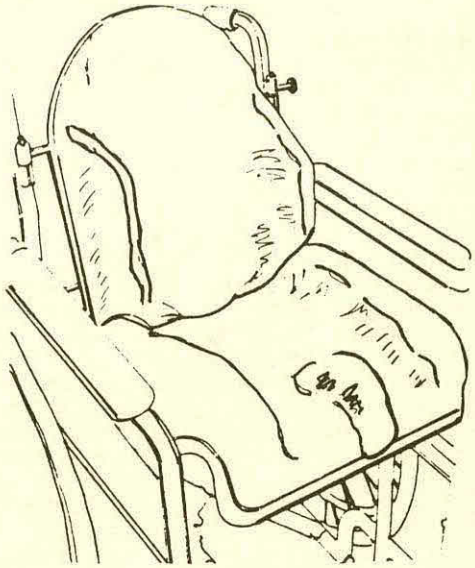

Figure 1 - BMI System Interfaced Into Standard Wheelchair

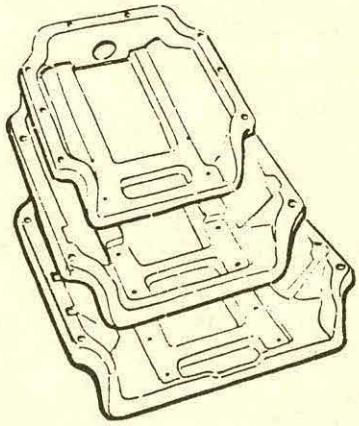

Figure 2 - Independent Seat and Back Plastic Trays Facilitate Adjustment and Wheelchair Attachment

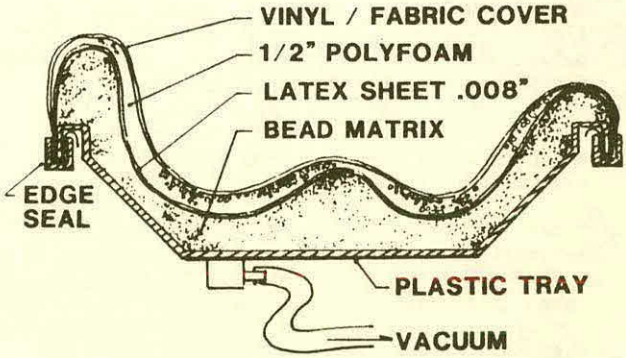

Figure 3 - Cross Section of a BMI Seat Component

reports on a three year clinical follow-up, as well as provides information as to its current development and availability status.

CLINICAL EVALUATION

Between the dates of June, 1982 and December 31, 1985, 123 clients have received BMI components; 6 received seats only, 23 received backs only, and 94 people received complete BMI systems. Clients were selected from the regular client flow through the UTREP Clinic, which evaluates approximately 250 persons annually. Subject selection was based on an independent therapy assessment of the client's seating needs, which were then compared with the proposed capabilities of the BMI system.

It is important to attempt to describe the characteristics of the population that participated in the study. The majority have a clinical diagnosis of cerebral palsy (98%). The description of this population using traditional terminology has limited value as it relates to seating decision-making. Therefore, a simplified classification scheme has been developed. This has been reported upon in detail previously (6, 7). The essence of this classification scheme will be described briefly. The two key inherent variables that primarily affect seating decisions have been identified as; a) abnormal tone (lack of neuromotor control) and b) skeletal deformity (either existing or pending). The classification scheme focuses on these two variables. Definitions are as follows:

Neuromotor Control

Good - indicates an ability to independently attain and sustain functional voluntary control in the seated posture

Fair - indicates an ability to independently attain and sustain functional control for only short periods of time

Poor - indicates an inability to independently attain functional control in the seated posture

Deformity

Mild - indicates minimal deformities, symmetrical posturing readily attainable with no functional limitations to range of motion (ROM).

Moderate - deformities may exist to the extent of: flexible scoliosis of not more than 50° or flexible kyphosis and/or a dislocated hip with pelvic obliquity and/or a leg length discrepancy not exceeding two inches, and/or flexion contractures of the hip and/or knee not to exceed 110° and/or foot and angle deformity which prevents the foot from assuming the a plantargrade position even if strapped.

Severe - defined as an flexible or rigid deformity to the spine, hips, pelvis, knees or feet in access of those previously defined as being in the moderate category which prevents individuals from assuming a symmetrical seated posture.

RESULTS

Unfortunately, research tools do not readily exist for objectively assessing the positive or negative affects of seating systems on a person's function, deformity, comfort, mobility, educational performance, independence or ease of management. Attempts have been made at establishing such evaluation methodology, particularly for modular seating systems (8, 9). However, the wide variances that exist within the specialized seating population, the innate characteristics of the disabilities themselves, the inability to establish test and control groups, and the presence of other contributing factors such as motivation, environmental influences, make the determination of objective assessment data time consuming, costly, and usually with considerable inconvenience to the population being studied. Therefore, the following results are acknowledged to be largely descriptive and subjective at best. However, within these limitations it is hoped that the information presented will be of value to clinicians contemplating the use of the BMI system, or similar types of seating technology. The results presented are only for those cases which received complete BMI systems (94 people).

All subjects except two had a primary diagnosis of cerebral palsy. One person had a spinal cord injury and the other Duchenne Muscular Dystrophy. The average age of the study population was 14.8 + 6.9 years (standard deviation assumes that the population follows a normal distribution, and therefore 68% falls within the age range of 7.9 to 21.7 years, with 14.8 being the average age). Each BMI system was used an average of 10.8 + 7.7 months and required an average of 1.28 days to fabricate.

Using the rating system described above the distribution on the neuromotor control scale was (good, fair, poor). The distribution on the skeletal deformity scale was (mild, moderate, severe). If we then combine individuals with common characteristics of neuromotor control and deformity we obtain groupings that are useful in decision making related to selection of specialized seating.

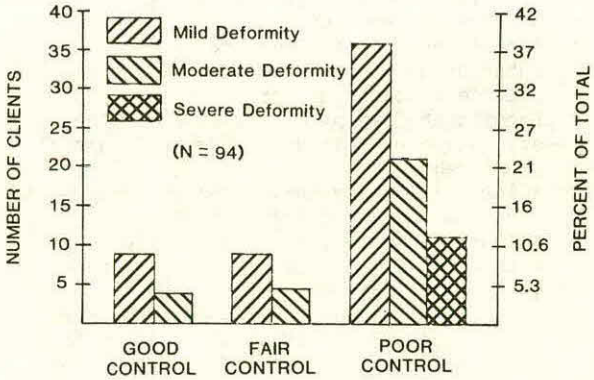

Figure 4 shows that the study group consisted mainly of people that had "poor" neuromotor

involvement with "mild" (36 cases) or "moderate" (22 cases) or "severe" (11 cases) deformities. There were no cases reported with good or fair neuromotor control with severe deformities. Average fabrication time for the poor/severe group was 1.45 days which was above the total population average (1.28 days).

PROBLEMS ENCOUNTERED

During the 3 1/2 year course of the clinical evaluation there was a number of rejections and revisions done to the seating system. In general, these were relatively low considering the severity of the seating needs. The Spinal Cord Injured client was removed from the BMI system due to pressure problems that were considered above safe levels. Seven components (2 seats and 5 backs) had to be redone mainly due to a lack of agressiveness in providing the needed control or for reasons of growth. Seven components had to be revised, most often to relieve pressure areas. Other problems encountered were due to difficulties in accommodating a four inch long leg length discrepancy and several cases in which dynamic spinal curves were not being adequately controlled.

CONCLUSIONS

Durability of the BMI system has been excellent in that there were no reports of failures of the seating components or wheelchair interface hardware. Furthermore, there were no reports or revisions due to heat related problems. Changes for growth, even for the earliest particpants, have been low possibly due to the older average age.

For the population described the BMI system offers an economic option for providing a durable custom contoured seating system. The system can be fabricated and provided in an average of 1.25 days (10 working hours). The system readily interfaces with most standard wheelchairs and will accommodate individuals from approximately 3 years old to adulthood.

FUTURE PLANS

The BMI System is now being used in six centers in addition to the UTREP. Commercialization of the system is now underway which includes design refinements to reduce cost and ease of fabrication. General availability is anticipated for Spring, 1986. Periodic educational seminars will be continued at the UTREP to facilitate training of clinicians in the appropriate use of BMI system.

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EFFECT OF ADAPTIVE SEATING ON PULMONARY FUNCTION IN CHILDREN WITH CEREBRAL PALSY

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ABSTRACT

This study was designed to measure the vital capacity (VC), forced expiratory volume in 1 second (FEV_1), and expiratory time (ET) of children with cerebral palsy when seated in a regular sling-type wheelchair and in an adaptive seating system. Eight children between the ages of 5 and 12 years who were diagnosed as having cerebral palsy were studied. A spirometer was used to obtain VC, FEV_1, and ET. The results showed a 57.7% increase in VC, 51.6% increase in FEV_1 as a percentage of VC, and 55.0% increase in ET in the adaptive seating system compared with the standard wheelchair. The results of this study show that pulmonary function increases when children with cerebral palsy are properly seated and positioned. These results have important implications for speech, sitting for prolonged periods, and prevention of hypoxia and pulmonary hypertension.

INTRODUCTION

Children with cerebral palsy exhibit varying levels of diminished respiratory function (1, 2, 3). Increases in vital capacity after a program of breathing exercises (4) suggest that poor voluntary control of the muscles of respiration is one explanation for diminished respiratory function. Another possible factor is the presence of spinal deformities that, whether functional and flexible, or structural and fixed, tend to alter the shape and volume of the chest and abdomen and consequently restrict their expansion.

Clinical observations suggest that seating position affects breathing among children with cerebral palsy (5). It is important that the best environment for communication whether with the use of hands or vocalization be provided to the child. Adaptive seating ensures that the child is seated upright with the head, neck, and trunk maintained in an upright position. Contact surfaces on the backrest and seat surface provide forces that tend to oppose spinal curvatures and support the chest and abdomen in as normal a position as possible. This is particularly true when the curves are flexible. In view of this effect of adaptive seating on the chest and abdomen, it is reasonable to expect better pulmonary function when the child is placed in an adaptive seat.

This study was designed to measure the vital capacity (VC), forced expiratory volume in 1 second (FEV_1), and expiratory time (ET) of children with cerebral palsy when seated in a regular sling-type wheelchair and in a wheelchair with modular inserts. Our purpose was to compare the effects of adaptive and nonadaptive seating on pulmonary function.

METHODS

Subjects

Eight children between the ages of 5 and 12 years, diagnosed as having spastic cerebral palsy, were studied. They were able to follow all the instructions given for correctly using the spirometer. All the subjects were non-ambulatory and none had any apparent evidence of intrinsic lung disease.

Materials and Equipment

Vital capacity, forced expiratory volume, and expiratory time were measured by a Breon Spirometer, Model 2400, with the mouth piece slightly modified for pediatric use. The Breon Spirometer was chosen for use in this population because, unlike other more sophisticated devices currently in use for pulmonary studies, it provided a visual feedback of performance to the subject and it was easy for them to use. A regular growing size Everest & Jennings wheelchair with sling seat and back was used as the nonadaptive seating device. A seating simulator designed and built by the University of Tennessee Rehabilitation Engineering Program for use with Modular Plastic Inserts (MPI) (6) was used as the adaptive seating system. Incorporated in the design of the simulator are independent adjustments for orientation in space, hip flexion angle, height of the foot support, length of the seat surface, height of the backrest, and the headrest. The backrest and seat surface modules provided lateral supports for the trunk and thigh, respectively. A chest panel and lap belt were also used to provide additional support.

Data Collection

The sequence of testing in both seating systems was randomized for all subjects. For the adaptive seating system, modules of suitable size were selected for each child. With the child in the seating system, adjustments were made to ensure that the head, neck, and trunk were supported in the upright position. The pelvis was placed in a midline position. The position of the hip joint was monitored with an electric goniometer and was maintained at 90° ($\pm$ 2°) of flexion. The knee and ankle joints were supported in 90° of flexion. In the nonadaptive seating position, each child was seated in an upright position. However, unlike the adaptive seating system, no supports were provided and the child's movements or positions were unrestricted. A 10-minute period was allowed after a change in seating system before data collection.

The children were given thorough instructions to perform the tasks and were allowed several practice runs. They were asked to take a deep breath and blow into the tube as fast as possible. They were allowed to perform four

tries in each seating system. The highest VC, the FEV_1, and the longest ET were obtained from the recorded plots. The FEV_1 was also calculated as a percentage of the VC (FEV_1 % VC). A t-test was performed on the difference between the means of VC, FEV_1 % VC, and ET in the two seating systems.

RESULTS

Mean VC obtained in the standard wheelchair was 0.97 liters and 1.53 liters in the adaptive seating system. This represents an increase in VC of 57.7% (Table I). Mean FEV_1 % VC increased from 51.7 in the standard wheelchair to 78.4 in the adaptive seating, a difference of 26.7 and an increase of 51.6%. Expiratory time also increased from a mean of 2.0 seconds in the standard wheelchair to 3.1 seconds in the adaptive seating system, a 55.0% increase. The high standard deviations obtained for ET is attributable to the relatively high ET values obtained on one subject in both seating systems. The t-tests produced statistically significant differences ($P<0.05$) between the means of VC, FEV_1 % VC, and ET obtained in both seating systems.

DISCUSSION

The results of this study show that pulmonary function, as measured by VC, FEV_1 % VC, and ET was higher when a child with cerebral palsy was seated in an adaptive seating system compared with a nonadaptive seating system. The increase in pulmonary function in the adaptive seat is perhaps attributable to changes in the physical shape, structure, and capacities of both the thorax and abdomen and better control of the muscles of respiration. In addition to reversing flexible spinal deformities, adaptive seating also places the head and neck in proper alignment with the thorax and, consequently, minimizes obstruction of the airways. Lower FEV_1 % VC obtained in the nonadaptive seating position may be due, at least in part, to some degree of obstruction of the upper airways. It is our intention to examine this more closely by monitoring other parameters of airway obstruction such as residual lung volume and flow-volume curves. Also of study interest is the effect of adaptive seating on functional residual capacity as it relates to preventing alveolar hypoxia and subsequent pulmonary hypertension.

The increase in ET for children in the adaptive seating system is due to a corresponding increase in VC. This is probably a consequence of an unobstructed airway in the children while seated in the adaptive seating system. This has important implications for speech and communication through vocalization. Vocalization relates to both voice and use of connected speech. Because of better VC and ET, it is reasonable to state that for improved breath support for more normal speech production the adaptive seating position is the position of choice in this population. Clinical observations have also indicated that the adaptive seating position enhances feeding and the function of the alimentary system. Also, given the relative poor physical fitness (7) and the predisposition to respiratory infections (8) of children with cerebral palsy, it is important that pulmonary function be maximized, especially during prolonged seating.

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	VC (Liters)		FEV_1 % VC		ET (sec)	
	WC	AS	WC	AS	WC	AS
N	8	8	6	6	8	8
Mean	0.97	1.53	51.7	78.4	2.0	3.1
SD	.29	.24	16.93	17.23	2.30	3.58
% Change	57.7		51.6		55.0	

Table I - Mean VC, FEV_1 % VC, and ET in adaptive seating (AS) and standard wheelchair (WC).

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Weightbearing Tissue Contour and Deformation By Magnetic Resonance Imaging

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ABSTRACT

Preliminary results were obtained from magnetic resonance images of a normal and a paraplegic supine subject to measure in vivo soft tissue strains on various support surfaces. Crossectional views of the tissue anatomy could be observed conveniently using this safe and non-invasive technique. Tissue deformation was quantitated in the 0 to 60 mm Hg pressure load range with an approximately 1 mm^2 spatial resolution. From this data the compressive stiffness was calculated for gluteus muscles, skin and subcutaneous tissues and compared to the stiffness of foam wheelchair cushions. The results will be used to aid in the design of new cushions for wheelchair and seating use.

INTRODUCTION

Compression of soft tissues and their deformation under externally applied forces are the limiting factors in the fitting of body support systems in orthotics and prosthetics and wheelchair seating. Tissue pressure and strain are also important determinants of comfort on beds and mattresses. Measurement of muscle cross-sectional area can also be used to estimate contraction strength and function in ergonomics. Through the use of Magnetic Resonance Imaging (MRI), a safe and non-invasive method for measuring in vivo tissue shapes has become available. By observing the response of atomic nuclei in magnetic fields this technique can reconstruct longitudinal and transverse views of internal tissue structures from skin surface to bony prominences.(1) Without known harmful effects, magnetic resonance techniques can provide clear views of anatomy and pathology of internal body structures in the gravity environment of recumbent human subjects.

The objective of this work was to determine the application of MRI techniques for measurement of in vivo soft tissue strains on various support surfaces to aid in the design of cushions for wheelchair and seating use.

METHODS

MRI signals were generated, processed, stored and displayed by the Siemens 0.3 Tesla Magnetom Imaging System.

The preliminary results reported here were obtained from two 23-year old male subjects. One was a paraplegic of 4 years duration weighing 123 lbs. and the other was a normal control of slightly larger body size, weighing 140 lbs. The subjects were supine in the magnetic field of the imaging system supported by 3 inch thick polyurethane foam cushions (E&J Durafoam™ or Scimedics Laminair™). Measurements were made with and without a 10 kg mass sandbag placed on the right iliac crest to increase skin and gluteus muscle tissue loading by the wheelchair cushions at the trochanter and ischial tuberosities. Measurements were also taken with the subjects recumbent on cushions but with the buttocks unsupported and the tissues hanging free without compressive loading. The external pressure was measured under the bony prominences before and after each imaging run using the Scimedic pressure evaluating system with a transducer bag of 65 cm^2 area. The applied surface load was then calculated from the measured pressure and the transducer surface area. To improve the signal to noise ratio, an imaging run required the averaging of four images and was completed in less than 10 minutes. Between runs the position of the subjects was maintained constant by registering the location of surface anatomic landmarks on the support table.

Images were displayed on the CRT and hard copies made on photo negative films. Cursor control on the CRT display allowed measurement of tissue dimensions, cross-sectional areas, image slice volumes and signal intensity.

RESULTS

Clear images of the cross-sectional anatomy were obtained through the tip of the ischial tuberosity and the greater trochanter of the femur. Muscle thickness and the thickness of the skin and subcutaneous tissues were measured vertically from the bony prominence to the outside skin surface posteriorly. The total tissue thickness from skin to bone was then plotted as a function of the mean pressure measured at the skin-support interface before and after the imaging run. For the two subjects, the results are shown on Figure 1 for each bony prominence. Rapid reduction in tissue thickness is evident with increasing surface pressure. Paraplegic tissues were maximally indented by their own bony prominences without additional weight.

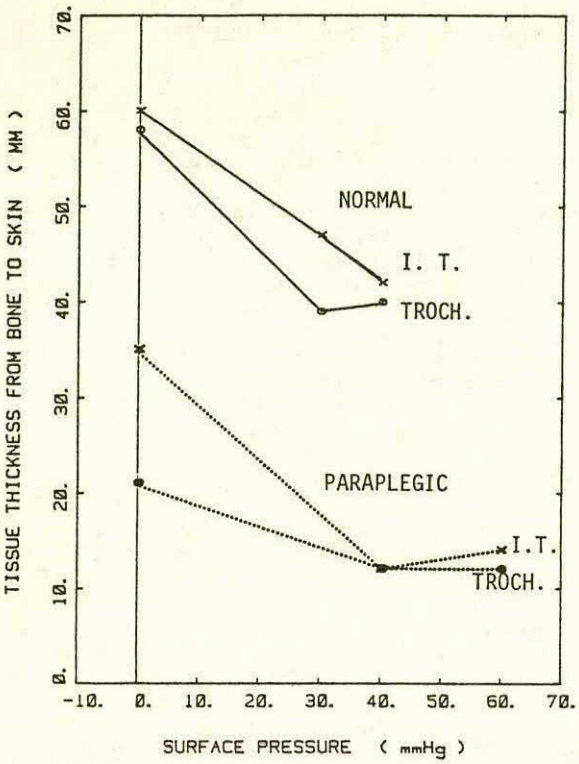

Figure 1. Change in Tissue Thickness with Increasing Pressure Under Boney Prominences of Supine Subjects.

The additional 10 kg weight caused no further reduction in tissue thickness; it only increased tissue pressures. On the normal tissue, however, the additional weight on the ischial tuberosity could cause further indentation.

Tissue deformation was also calculated as the difference between tissue thickness at the free hanging condition minus that thickness at the applied load. This deformation, the mean of two observations, is plotted in Figure 2 for each type of tissue against the applied load, which was calculated from the measured pressure. The reduction in thickness with increasing loads is shown here as positive deformation for all tissues except normal skin, which showed anomalously increasing thickness with increasing surface loading. These load-deformation plots were used to calculate the initial slopes of the curves to obtain an estimate for the stiffness of each tissue in the load bearing regions of the buttocks. The calculated stiffness values are shown in Table 1.For comparison, note that the support cushion stiffness in the 0 to 25% deflection range for a 7 cm diameter indentor is approximately 3.6 N/mm for Dura foam TM and 3.7 N/mm for Laminair TM. These values are a reasonable match for muscle but too low for the skin. For deflection larger than 25% the cushion stiffness will further increase making the discrepency smaller at the skin but larger for the muscle.

Table 1. Compressive Stiffness of Tissues from MRI and Surface Pressure

	Stiffness (N/mm)	
Subject	Muscle	Skin
Normal	1.4	-8.7
Paraplegic	3.4	6.8

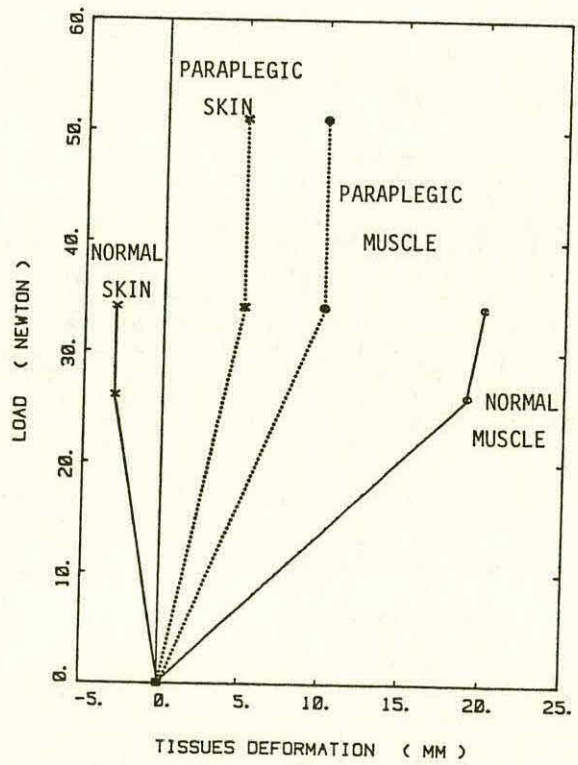

Figure 2. Load-Deformation Curves for Individual Soft Tissues. Each Value is the Mean of Ischial and Trochanteric Observations.

The changing cross-sectional area of the muscles near the ischial tuberosity and trochanter was also plotted as a function of applied pressure and shown in Figure 3. The normal muscles showed rapid decline of cross-sectional area with increasing pressure. The area of the paraplegic muscle, however, was nearly constant as pressure increased, but lateral shifting of the muscle mass with increased enveloping of the bony prominences was observed.

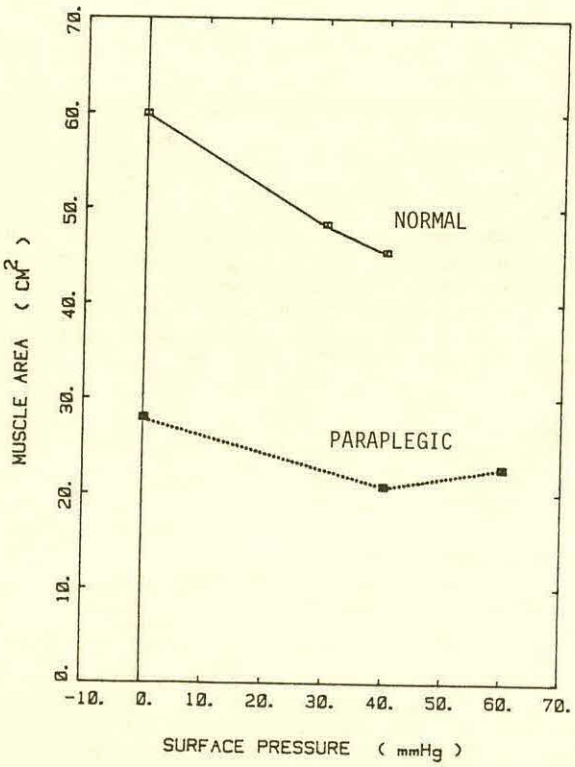

Figure 3. Change in Cross-Sectional Area of Gluteous Muscles Resulting from Increased External Pressure.

CONCLUSIONS

In vivo soft tissue strains could be measured at high resolutions using the convenient, safe and non-invasive method of magnetic resonance imaging. The results showed "bottoming out" of the paraplegic tissues with less pressure than the normal controls under the compressive action of the boney prominences. Increased stiffness and lateral shifting of the paraplegic muscle mass under compressive load indicates the need for precise matching of the shape and material of the support surface to the tissue contour to preserve as much of the soft tissue padding under the boney prominences as possible.

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PRESSURE MONITORING CHAIR

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ABSTRACT

An instrumented adjustable chair has been developed for the investigation of the effect of seat shape and seat orientation on pressure distribution. The chair contains 204 linear force transducers to measure shear and normal forces independently over the seat bottom and seat back. The force transducers are monitored by a personal computer. Data samples may be collected from a 3 second interval up to extended sampling rates. Output is in the form of pressure readings for each transducer and pressure contour maps. Preliminary testing indicates that the system is reproducible. Pressure patterns are dependent on shape and posture, but repositioning of subjects greatly effects pressure values.

INTRODUCTION

Adaptive seating for individuals with neuromuscular involvement has undergone rapid advancements in the past five years. The theoretical basis for selection of seating configurations has been more clearly delineated (1,2) and the range of commercial and custom options for seat selection has been greatly expanded. At the same time advancements in orthopedic surgery have made upright sitting feasible for many children and young adults who otherwise would be confined to horizontal positioning.

The physical goals of adaptive seating for individuals with neuromuscular involvement include:

1. inhibition of abnormal reflex patterns
2. facilitation of voluntary control
3. correction of functional postural deformities
4. accomodation of structural postural deformities

Achievement of these goals frequently requires intimately fitted components, especially for those clients with reflex dominance, severe imbalance of muscle tone and resultant deformities. The considerable forces produced by such external supports and their potential effects on body tissues require evaluation.

Several centers have reported results of pressure evaluation at the skin-cushion interface using matrices of air-filled transducers under the buttocks of seated subjects (3,4,5,6). The objective of these investigations is reduction of the incidence of decubitus ulcers in patients with spinal cord lesions. Drummond, Breed and others at the University of Wisconsin (7,8,9) developed a pressure-scanning seat with the objective of examining pressure distribution and the center of pressure in subjects with asymmetry due to spinal deformity. Following their study of subjects with cerebral palsy they noted the limited usefulness of the scanner for this particular patient group and suggested modifications such as the capacity to measure back, lateral and foot pressures (8).

The design of the prototype described below incorporates these modifications and has the additional capacity to measure shear forces as well as pressure under various conditions of contour and orientation of the support surfaces. The objective is to evaluate pressure distributions and the center of pressure in seating configurations commonly prescribed for the severely involved neurological client.

DESIGN

The total system consists of three components: the chair; the transducers and associated electronics for communicating with the computer; and the computer with software for data processing and output.

Chair

The chair portion of the system is a position and orientation simulator (figure 1). It is adjustable in sitting depth, seat bottom inclination, seat back

angle and axial seat back angle. Footrest and headrest positions and orientations are also adjustable.

The chair is mounted on a wheeled base for transportability.

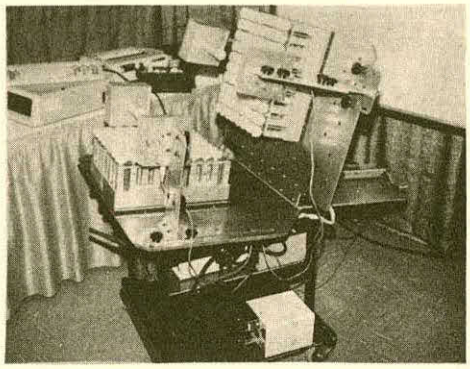

Figure 1: The chair with transducers mounted on the seat bottom and seat back. The power supply and data acquisition system are mounted underneath.

Transducers

The criteria established for this development project required that there be a large number of force transducers and that shear and normal forces be discernable. Air bladder transducers measure peak forces regardless of the orientation of the force, thus they were not suitable for our application. Existing strain gauge transducers were too expensive since many were required. Variable resistance elastomers are commonly used for pressure measurement in rehabilitative settings, but we found them to be of insufficient resolution and reproducibility for our application.

After considerable investigation we designed a transducer specific to our application, (figure 2). The sensing element was a Hall Effect transducer within a small magnetic field. The Hall Effect transducer produces a signal dependent on the orientation and strength of the magnetic field. Two magnets were mounted on a spring and shaft assembly such that they pass by the transducer when the shaft is depressed. This configuration produced a linear signal over a translation of approximately 2mm.

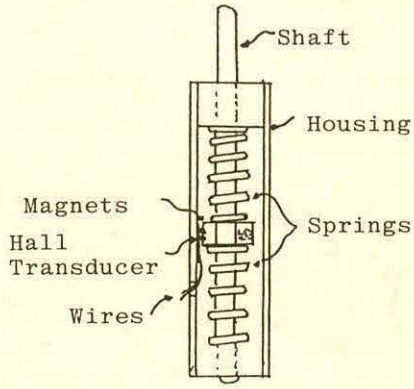

Figure 2: Schematic of the transducer.

A power supply was built to provide 6 volts to each transducer. Output was analog, and ranged between 3 and 9 volts. The Keithley DAS (data acquisition system) Series 500, was purchased for analog to digital conversion and multiplexing. The Keithley was found to be extremely good for this application.

The seat bottom and seat back each had 80 transducers mounted perpendicular to their surface to measure normal forces. They were arranged in 8 by 10 arrays with 38mm between centres. The 80 transducers were split into four, 4 by 5 quadrants. Four transducers were mounted under each quadrant to monitor shear movements of the quadrant. We would rather have had greater resolution of shear, but size restrictions were prohibitive. Twelve transducers were made available to be located laterally or in other positions.

Each normal transducer presented a shaft on the outer surface. Each shaft inserted into a 32mm diameter cap. Caps were made with top surfaces varying between flat and 45 degrees. By changing caps, seat bottom and seat back surfaces of different shapes could be generated.

Data Processing

Data collection and processing functions include the following:

a) Defining of data collection and processing functions. Parameters such as sampling rate, number of samples to be recorded and whether to record normal and/or shear forces are determined by the operator. Samples may be collected at rates of 3 seconds or greater. For processing the data to produce pressure maps (figure 3) the increment between pressure contours may be set. Also, procedures for calculating centers of pressure may be turned on or off.

b) Transducer offsets may be recorded at any time to provide a zeroing function.

c) Data may be collected in a continuous scanning mode or in a test mode as defined by the data collection parameters. During data collection pressures are displayed in mmHg units.

d) Data is processed to generate pressure maps after data collection is finished. Each data sample may be processed any number of times according to different processing parameters.

e) Processed and unprocessed data may be reviewed or printed. Also, pressure maps may be plotted.

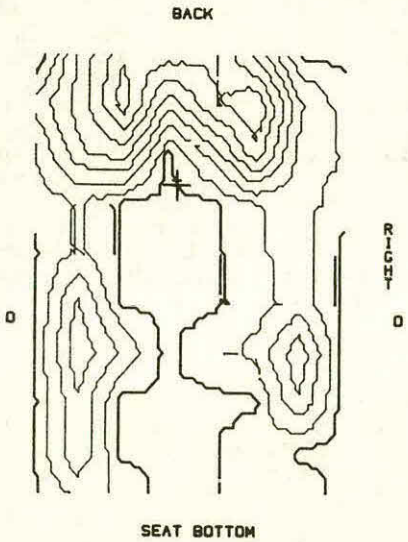

Figure 3: Seat bottom pressure map of a normal child on a flat surface. Pressure contour increment is 3 mmHg. The feet are supported. Note there is very little pressure under the thighs. The cross indicates the center of pressure.

The software is coded in Pascal and runs on an IBM PC/XT computer with 256K RAM and a color graphics adapter with color monitor. The graphics is low resolution, which is sufficient for this application.

RESULTS

To date, the pressure monitoring chair has just been operational for a few months. With few exceptions time has been spent on familiarization and testing of the device. Ninety percent of the transducers are functioning to our expectation. Their absolute error is within 5% and they return to within 1 mmHg of zero after removal of the load. The other 10% are sticking. This is due to friction on their shafts, a problem that can be resolved. Absolute error ranges up to 20% and all return to within 4 mmHg of zero. The transducers have only required recalibration once since completion of the chair.

During our familiarization with the chair it has become obvious that pressure distribution is very sensitive to seat shape and orientation. Small adjustments in footrest height make large differences to sharing of pressure between the thighs and buttocks. Pre-ischial bars create large pressure peaks.

CONCLUSIONS

The pressure monitoring chair is a prototype. In many respects it is quite crude but it will help us define better design parameters for future versions.

Preliminary results indicate that support surface shape does have a substantial effect on the distribution of pressure as does the orientation of the seat and its components. It is hoped that our new development will help define these relationships and contribute to the design and delivery of seating systems.

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ADDRESS

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A Computer Keyboard Designed for People Who Use a Mouthstick

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Abstract
The requirements for a "keyboard" that is designed for use by people who do not eight finger entry capability are discussed and form the basis for a new keyboard layout that can be used effectively by computer operators who use a mouthstick or similar single digit means of entry. The results of testing the efficiency of a number of keyboard designs are presented and recommendations for the most efficient keyboard are given.

While personal computers hold great promise as a means of providing severely handicapped people with a means of effectively interacting with and controlling their environment, current keyboards limit the fulfillment of the promise. Present generation keyboards have been designed for operators who have ten fingers to use and are capable of inputting simultaneous multiple keystrokes. These input devices are often clumsy or ineffective tools for people who are limited to stroke inputs with one activator (e.g. a mouthstick or single finger). This study has been conducted to develop a keyboard-type control interface that is specifically configured for such people.

A QWERTY keyboard (standard typewriter keyboard) has been more designed by accident than by logic. The original design was to help prevent any two adjacent keys from being used consecutively because they were likely to collide and stick on the early manual typewriters. Several superior keyboards have been promoted for electric typewriters but these designs are for use with eight fingers. The placement of the keys is to minimize consecutive letters being struck by the same finger and do not consider the "single digit" problem where the keys are struck by a single pointer and the key placement should minimize pointer motion as measured in units of keys traveled across.

To study this problem, software was developed so that the keyboard exists only in computer memory and is accessed by means of a mouse type control interface. The selected key is continuously displayed on the screen and is "typed" by sip or puff. This keyboard is referred to as a "virtual" keyboard because it has no tactile existance. There are several advantages to a virtual keyboard:

(A) The keyboard can be instantly modified or replaced,
(B) Permanent keyboard storage is simply a disc data record,
(C) The keyboard "wraps" around itself (when the pointer moves off any side of the keyboard, it comes back on the other side)
(D) Horizontal and vertical keyboard sensitivities are adjustable,
(E) Various "function" keys can be defined,
(F) The system is "Interrupt driven" such that both the keyboard and the appliance work simultaneously,
(G) Software advantages such as invisibility.

Thirteen keyboard designs using single and double letter frequency tables were developed. Space/return was chosen to be the origin for the keyboard descriptions because it is the most frequently used "character". This was then surrounded by the most frequently used keys according to the single letter frequency table. These were placed by their usage according to the double letter frequency tables. The lesser used letters were placed in the corners, and then separate rows of special characters and numbers were then added.

A 22000 character excerpt from a story by Steinbeck was analyzed for single and double letter frequencies to verify average letter distribution. The results of this analysis are shown in Tables 1 and 2. A computer analysis of the excerpt was then made for each of the thirteen keyboard designs to determine the number of letter positions a pointer would have to move to type the excerpt. These results are shown in Table 3.

A QWERTY keyboard was input as a control. Special characters and numbers were standardized and used in all other keyboards without change. These other keyboards can be classified by row and column count where letters are positioned. Letters were placed in rows/columns: 1 by 26, 2 by 13, 3 by 9, 4 by 7, and 5 by 5. There were two types of keyboards for each shape, an alphabetical ordered set of keys and at least one frequency ordered set of keys. An analysis of the motion required to type the excerpt with the different keyboards is shown in Graph 1. It is inferred that a square keyboard would be best in the absence of special characters and numbers. As the graph shown; however, a 4 by 7 keyboard (with special character and number rows added) minimized pointer motion for both types of keyboards. The lower curve (minimum movement) is of frequency ordered keyboards. The order used by the Apple King keyboard (a commercially available tactile keyboard) has the minimum pointer movement required of any of the keyboards tested; although, all of the frequency based 4 by 7 keyboards gave similar results.

The conclusion is that a 4 by 7 frequency based keyboard minimizes the pointer movement required to type a paper of approximately average letter distributions.

Several quadriplegic patients are currently using the system to determine desired changes to the current hardware and software. This usage will allow parameters to be set for a more intensive

training to follow. A commercially available program "Typing Tutor" has been altered for the new keyboard system and is being used to both train and test the patients. Typing Tutor is geared to the touch typist and is not entirely suitable to the single pointer system but its advantages of clarity, ease of use, and immediate availability far outweigh its disadvantage of being geared to the eight-fingered typist.

As patients are successfully trained to interface to the computer, work will progress on the environmental control aspects of the project. An interface between the computer and any electrical appliance is already available. This would allow a handicapped person to control the home environment from his bed or chair.

Table 1

Single Letter Frequencies
(Steinbeck Exerpt)

Blank - 3930

A	- 1283	N	- 1055
B	- 299	O	- 1260
C	- 388	P	- 276
D	- 1022	Q	- 12
E	- 2105	R	- 936
F	- 343	S	- 929
G	- 405	T	- 1582
H	- 1284	U	- 519
I	- 1074	V	- 99
J	- 80	W	- 440
K	- 212	X	- 9
L	- 784	Y	- 401
M	- 337	Z	- 0

All others - 742 (includes numbers and special characters)

Table 2

Double Letter Frequencies
(Steinbeck Exerpt)

last letter:	A	B	C	D	E	F	G	H	I	J	K	L	M	N	O	P	Q	R	S	T	U	V	W	X	Y	Z
first letter: A	0	52	132	89	88	66	67	146	88	0	6	122	60	294	24	44	0	201	167	230	10	19	128	0	62	0
B		6	0	0	92	2	0	1	45	0	0	33	5	0	31	0	0	19	1	0	49	0	0	0	16	0
C			1	1	74	0	0	73	51	0	68	24	0	32	51	0	0	33	6	27	32	0	0	3	4	0
D				15	336	20	12	0	111	0	0	73	0	246	164	0	0	60	13	0	17	0	1	0	69	0
E					63	39	64	694	41	2	70	200	99	203	17	70	0	420	174	189	12	102	76	7	63	0
F						23	0	0	41	0	3	37	0	0	115	0	0	44	5	22	17	0	0	0	0	0
G							16	41	55	0	0	5	0	131	53	0	0	54	16	0	39	0	0	0	3	0
H								0	193	2	4	0	0	1	87	0	0	15	67	603	26	0	74	0	6	0
I									0	0	28	160	86	305	35	27	0	108	171	222	11	33	52	1	2	0
J										0	0	0	0	1	65	0	0	0	0	0	13	0	0	0	0	0
K											0	19	0	31	32	0	0	18	17	0	0	0	0	0	1	0
L												128	3	7	99	28	0	26	35	30	71	2	9	0	86	0
M													7	6	102	10	0	10	15	1	31	0	0	0	4	0
N														8	202	0	0	26	46	80	70	1	33	1	9	0
O															65	56	0	235	50	215	212	26	133	2	61	0
P																23	0	19	7	5	62	0	0	3	5	0
Q																	0	0	6	0	12	0	0	0	0	0
R																		27	25	67	71	2	2	0	23	0
S																			33	174	91	0	6	0	16	0
T																				53	94	0	10	0	17	0
U																					0	0	0	0	0	0
V																						0	1	0	0	0
W																							0	0	2	0
X																								0	0	0
Y																									0	0
Z																										0

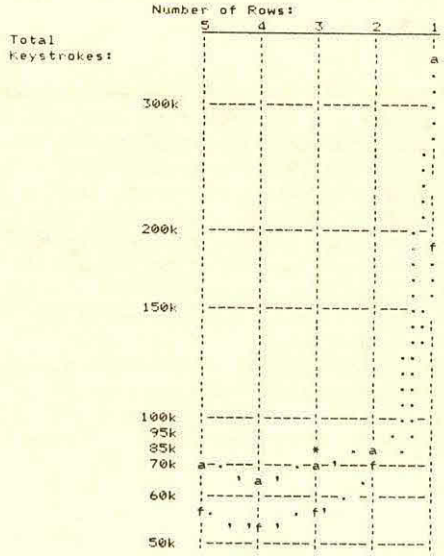
Table 3

Virtual Keyboard Statistics
(Steinbeck Exerpt of 22269 Characters)

Keyboard Designation	Right	Left	Up	Down	Total
Standard Keyboard:					
QWERTY (3X09)	26151	29940	11224	11154	78469
Alphabetically ordered keyboards:					
A1X26	162735	162027	776	776	326314
A2X13	36922	33352	6149	6149	82572
A3X09	21737	27524	10840	10840	70941
A4X07	16425	20870	13080	13089	63464
A5X05	14090	16178	18922	18932	68122
Frequency ordered keyboards:					
F1X26	159323	130704	776	776	291579
F2X13	31545	26668	5429	5429	69071
F3X09	20137	20524	9105	9105	58871
F4X07 (Apple King)	13522	13865	12542	12515	52444
F4X07 (v1)	15526	16226	10777	10777	53306
F4X07 (v2)	14889	15554	11645	11636	53724
F4X07 (v3)	15314	14005	13067	13049	55435
F5X05	14501	15142	13427	13407	56477

Graph 1

Log Plot of Total Keystrokes vrs. Number of Rows in Keyboard
(Steinbeck Exerpt of 22269 Characters)



f = frequecy ordered keyboard
a = alphabetic keyboard
* = QWERTY keyboard

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A SYSTEMATIC APPROACH TO DESIGN OF A MINIMUM DISTANCE ALPHABETICAL KEYBOARD

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ABSTRACT

A deterministic strategy for the reorganization of a compact alphabetical keyboard is discussed as a method of improving interfaces for computers and communication devices. The algorithm uses statistical information about the English language to eliminate impractical arrangements and concentrate its evaluation procedure on those which appear more useful. A good approximation to the optimal solution is generated by exactly solving a series of reduced assignment problems. The improvement in predicted rate to be obtained by this optimization scheme is significant.

INTRODUCTION

Individuals who must rely upon assistive devices for communication find themselves in the frustrating situation of being limited to message output rates which fall far below that of normal speech. One of the most significant factors which determines a person's communication rate is the interface between the device and its operator.

The most basic act in the operation of a communication device is the selection of items from a menu, whether the selection strategy is scanning, encoding, or direct selection. One method of improving the person/machine interface without altering the physical construction of the device facilitates this act by reorganizing the menu on the device's "keyboard" in such a manner as to reduce, on average, the time required to make a selection. It has been hypothesized (1,2) that, in the operation of a device by an "expert" (i.e., practiced) user, it is mainly a person's motor ability which determines message output rate. Redesign of the interface to make it "optimal" for the expert user must therefore be based primarily upon movement-time considerations. Two datasets are required. The first of these consists of a description of the amount of time required for the person to make the transition from selecting one location on the keyboard to selecting another location, either in the form of a simple lookup table or as a mathematical model of the person's motion. The second input to the optimization scheme is the set of transition frequencies between menu items. The optimization algorithm attempts to place in closest proximity on the keyboard those menu items which occur most frequently together in speech.

The systematic assignment of menu items to positions on a keyboard often involves an unmanageable number of possibilities. Such problems may be approached completely deterministically when the search space is not too large or the response surface is of a known shape. The other extreme approach is random search, which may be resorted to in the case of a hopelessly large search space and/or an extremely convoluted response surface. The current problem may appear to fall more into the latter category, but it is nonetheless possible to exploit information about the problem to focus upon likely areas. Since the overwhelming size of the problem dictates that we settle for a good suboptimal solution, any algorithm which adopts an entirely deterministic strategy had better be motivated by solid criteria for reducing the search space to a workable size. The method proposed here takes advantage of the statistics of the English language to generate a reduced set of possible keyboard arrangements for which the assignment problem may be solved exactly. The heuristic and deterministic components of the algorithm are easily separable and are applied in sequence.

THE PROCEDURE

The subject of this discussion is a variant of a simple assignment procedure called a "greedy" algorithm. Such a procedure generally makes use of two rank-order lists: one of positions and one of objects to be assigned to those positions. The best object is assigned to the best position, the second-best object is assigned to the second-best position, and so forth until the worst object is banished to the worst position and the rank lists are exhausted. The greedy's simplicity is both its greatest advantage and its greatest weakness; it is fast, yet it ignores many arrangements which could be substantially better than those which it considers. The results yielded by a greedy algorithm are strongly dependent upon the criteria used to generate the rank lists. But despite its obvious simple-mindedness, the greedy could be quite useful for applications like determining the optimal configuration of items on a word or picture board for which each item constitutes a stand-alone utterance (i.e., items are not usually selected in sequence to compose an utterance). Since in this case transition times between items are not important, frequently-used items would be placed where they could be most easily accessed from a neutral or rest position, and their placement with respect to one another would not be a design consideration.

The scope of the search may be greatly expanded by combining the iterative framework of the greedy with some other technique for determining exact solutions of assignment

problems, such as exhaustive search, and partitioning the same rank lists into subset blocks. In this hybrid method, which we call "block-greedy", the items in the first block are assigned simultaneously to the best positions, the items in the next block get placed in the next-best positions, and so forth until the lists are exhausted. The assignment problem for each stage is solved exactly, generating the best possible solution given the constraint that items assigned in previous stages are not moved from their positions. The optimization criterion employed in the exact solution phase is that the product of the interkey distance and the frequency of transition, summed over all possible transitions between two keys, is minimized. We have applied this program to the design of a compact layout of the 28-element character set consisting of the alphabet plus period and space. The stages of development of this layout are shown in figure 1.

```
    * * *                 * * *
  * * * * *             * U S . *
  * O T H *             * O T H *
  * I _ E *             G I _ E W
  * N A R *             * N A R *
  * * * * *             * D L Y *

   Stage 1               Stage 2


    * * *                 Q Z X
  F U S . P             F U S . P
  G O T H M             C O T H M
  G I _ E W             G I _ E W
  K N A R B             K N A R B
  J D L Y V             J D L Y V

   Stage 3               Stage 4
```

Figure 1: Development stages of minimum-distance compact alphabetical layout.

The assumptions contributing to this particular design are that the user makes all of his selections on the keyboard with one finger or the equivalent (headstick, mouthstick, nose, etc.) and that all movement times involved in selection are directly proportional to the distance between letters being selected. Note that the proportionality of movement time to distance cannot in practice be assumed, since movement time often depends in complex ways on such parameters as target size, target location, and direction of movement (3). The assumption that minimum distance implies minimum time is made here for the dual purpose of simplifying the discussion and generalizing the result. Since no other assumptions are made concerning the user's movement capabilities, the geometry of the layout was chosen to be as compact as possible, yielding a 5 by 5 grid capped by a line of the remaining three positions.

At present, the "exact solution" portion of the algorithm employs the most inefficient technique, namely that of exhaustive search. There exist other methods of finding optimal solutions to this type of assignment problem, such as branch-and-bound techniques (4). These are currently being investigated and will eventually be incorporated into the current program, thereby improving its efficiency and allowing for the exact optimization of 12 to 15 elements within a reasonable time frame. The alphabetical layout shown here took approximately one hour to generate on an IBM PC-AT using a program written in C and affords an improvement in predicted communication rate (for the movement time assumptions mentioned above) of 26% over an array in alphabetical order with the space key centrally located.

DISCUSSION

The block-greedy procedure owes its reliability mainly to the fact that the frequency distribution of letters in the English language is highly skewed. The nine most frequently used letters alone acount for over 40% of the digrams (letter pairs) encountered in speech (calculated from (5)). This indicates that in a compact array, these are the letters which will be most frequently accessed from any position and therefore ought to be assigned to the central positions. Furthermore, optimizing the assignments of these nine only with respect to one another is sufficient to account for a significant improvement in message output rate. To support this rash assertion, let us consider for a moment how it is possible to degrade the solution so far presented. If we attempt to find the worst solution possible for this 28-element array (by reversing the evaluation criterion in the exact solution phase) and evaluate it, we find it to be 30% slower than our "optimal" layout. Now if we generate the worst solution we can within the limits of the constraint that the first nine characters are in their best positions, the result is only 13% worse, but still represents a gain of 16% over the layout in alphabetical order. As we consider blocks of lower rank, the quality of the layout becomes increasingly insensitive to degradation. Thus the partitioning of the rank lists into seemingly arbitrary blocks does not contribute to much inaccuracy as long as the initial blocks containing the high-frequency items are chosen large enough.

The block-greedy algorithm's usefulness lies in its ability to generate good-quality solutions to the assignment problem within a reasonable time span. Inclusion of the added dimension of code assignment or selection of alternative geometries are problems which do not lend themselves as well to such a sequential, deterministic procedure. To optimize with respect to additional dimensions requires that another procedure be invoked previously, additional a priori information be

applied, or that a different strategy be adopted. Ultimately, a hybrid algorithm consisting of one component which is capable of considering many dimensions (such as the genetic-based adaptive technique described in (6)) applied within a block-greedy framework should provide the capacity for generating good arrangements of large menus while taking into account as many optimization criteria and dimensions as present themselves.

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ADAPTIVE TECHNIQUE FOR CUSTOMIZED INTERFACE DESIGN WITH APPLICATION TO NONVOCAL COMMUNICATION.

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ABSTRACT

An optimization algorithm based on a genetic model is applied in the design of individually customized interfaces for computers and communication devices. The user is represented by a model of movement time dependence on two device parameters: distance and direction of movement. The genetic algorithm systematically generates keyboards and evaluates their performance (in terms of predicted communication rate) according to each user's model. Each generation's best performers are selected as 'parents' of the next generation of keyboards, until the procedure is halted.

The extent of improvement in predicted rate obtained by individually customized versus standard key arrangements and minimum distance key arrangements is considerable in some cases and is shown to be highly sensitive to each user's model coefficients.

INTRODUCTION

The accepted interface for most computer users is the QWERTY keyboard. There has been little interest in changing this keyboard, since the gains in speed are seen as not being worth the retraining and expense involved. Improvement in rate becomes much more important when we consider users with neuromotor speech impairment, particularly when a device is required to provide basic communication. The great variability of motor impairments seems to preclude the existence of any universal "best" design. There appears to be a need for customized interfaces where the design procedure incorporates the potential user's capabilities and restrictions. While customized hardware is often prohibitively expensive, computers offer the possibility of customizing through software. This paper considers the problem of customized design as one of adapting the interface to the user's needs and capabilities. The design technique operates on a model of the user's motor abilities and evaluates each potential keyboard in terms of the communication rate it predicts for that user-keyboard pair. For a discussion of the motor assessment technique on which modelling is based, see (1)(2) & (3).

Many parameters are required to specify an interface, including menu, arrangement, dimensions, key size and code, all of which can be treated by this optimization technique. This paper will discuss key arrangement optimization in order to demonstrate the design algorithm's capabilities. The assumption is that the user strikes keys in sequence, using a finger on depressable keys, a joystick moving a cursor on a CRT screen, or some other variant of this basic scheme. The goal is to arrange and locate the 'keys' to increase communication rate. A 'menu' of the twenty-six letters plus the space and period is considered. Given the vast number (28! or roughly 3×10^{29})of possible key arrangements, an exceptionally efficient technique is required for identifying good designs. 'A priori' knowledge can eliminate unworkable or very poor designs to arrive at a set worthy of further investigation. Searching among all or a representative sub-group of this set leads to selection of the design that proves best by some evaluation criteria. The specific search algorithm presented in this paper utilizes a technique known as a reproductive plan. It is based on simulated biological evolution, and employs mutation and natural selection to evolve improved designs.

DESIGN BY SEARCH

The search phase of the design process can take on many forms. One approach involves calculating an optimization function for one or more designs arrived at in the a priori knowledge phase, making improvements based on this evaluation, and continuing in this vein until a satisfactory design is reached, or no further feasible improvements can be envisioned. This is a deterministic search, often described by a hill-climbing analogy. Its weakness is that when the 'terrain' is complex and unknown it is possible to arrive at an inadequate design 'peak,' unaware that higher 'peaks' may be lurking beyond the neighboring valleys.

Another option is random search. Designs are generated at random and evaluated. This continues until a satisfactory design is reached, time or money runs out, or the designer is convinced that no satisfactory solution exists. In purely random search no information gained by past evaluations is used in generating new designs.

Exhaustive search i.e., evaluation of all possible designs, is feasible only where the number of possibilities is not too large.

Simulated biological evolution represents a mixed strategy. It contains several random components, chief among them being simulated mutation, that is, random change. The hill-climbing component is provided by simulated natural selection; it involves utilizing information on past good designs to influence the development of future designs (5)(6).

REPRODUCTIVE PLANS AND EVOLUTIONARY DESIGN

A reproductive plan is a simulated genetic-based adaptive stragegy; it mimics the mechanisms of biological evolution. The process begins with generation of a set of 24 keyboards (key arrangements). Each of the keyboards is evaluated and the better designs are selected to contribute offspring to the next generation.

Each offspring may be identical to its parent, or differ instead by one or more mutations. The number and character of these mutations are determined by chance; they do not necessarily produce improvements. The interaction of selection and mutation provides the technique with attributes of both deterministic and random search routines, effectively probing the entire search space without becoming stuck on the top of a low peak.

APPLYING EVOLUTIONARY DESIGN TO ACHIEVE CUSTOMIZED INTERFACES

Our design problem here consists of arranging a twenty-eight character set, the letters plus space and period, on a rectangular grid. Prediction of the user's average movement time in communication is based on digram frequencies i.e., frequencies of transitions from one menu item to the next in the production of a sample of text; and a model of the user's movement time as a function of device parameters such as inter-key distances, key widths, and directions of movement (1)(2)(3). Mutations are of the simplest type; two keys are chosen at random and their positions are interchanged.

Experience to date with assessment and modelling of the motor abilities of people with neuromotor deficits has shown that different individuals are represented by a wide range of values of the coefficients for each of the parameters in the model. In other words, the strength of the dependence of movement time on each parameter differs considerably from person to person (7). For example, Mr. A may take twice as long to move his hand from the bottom to the top of the test panel as he does to go from the top to the bottom; for Ms. B the situation may be reversed. The optimization technique presented here allows us to select the best keyboard for each of these individuals.

In addition, this procedure lets us compare (using a technique similar to that described in (8)), for example, how Mr. A would do using his own custom design with how he would do with Ms. B's, or with a standard design and thus lets us ask how much is to be gained by individually custom designing. If we simply took a "QWERTY" keyboard, or a board laid out in alphabetical order, how much worse would predicted rate be than on an individually customized layout? Alternately, what about a keyboard which is laid out to minimize average movement distance i.e., without regard to direction dependence of movement speed. The answers to these questions can only be empirical, and require evaluation of designs which take into account not only the numerous parameters mentioned above, but also such considerations as mental load for learning and experienced use, as well as individual differences in cognitive abilities.

The example below shows how much improvement customization brings to hypothetical users whose movement times depend on distance in the following manner: movement in the horizontal direction is either one, three or five times as rapid as movement in the vertical (toward and away from body). These values, which are within the range of values observed in assessing motor behavior of disabled device users(7), are chosen to illustrate how the extent of benefit of customization can depend on individual motor characteristics. Note that actual disabled individuals' models have usually been more complex and less symmetrical than these three hypothetical examples.

We assume the keys are located on a grid with adjacent keys one unit of distance apart, and that for the three users we are considering, movement time is proportional to distance with velocity varying elliptically with direction, according to the ratios shown in Table 1.

Ratio	Velocities	
	x axis	y axis
1:1	$v_x = 1$	$v_y = 1$
3:1	$v_x = 1\ 1/2$	$v_y = 1/2$
5:1	$v_x = 1\ 2/3$	$v_y = 1/3$

Table 1. Velocities in the x and y direction for 3 hypothetical individuals.

Interkey transition time is represented by

$$\sqrt{\frac{(d_x)^2}{(V_x)^2} + \frac{(d_y)^2}{(V_y)^2}}$$

where $d_x \equiv$ distance in x direction
$d_y \equiv$ distance in y direction

Performance in terms of communication rate for a sample of English (9) was calculated for each of these three hypothetical individuals for the following four layouts of the 28-item menu:

- an alphabetically ordered layout on a 7 (horizontal) by 4 (vertical)grid,
- a QWERTY layout within a 10 x 3 grid,
- a minimum distance layout within a 6 x 6 grid and
- a minimum time layout (i.e., a layout customized for each individual's motor abilities)

Note that for the case of $v_x = v_y = 1$, the minimum time layout is the minimum distance

layout. For minimum time layouts, grid size was chosen a priori. For the 1:1 ratio of movement times, a square grid was used; for the 3:1 ratio, dimensions of 10 (horizontal) by 3 (vertical) units were chosen; and for the 5:1 ratio, a 13 by 3 grid was used.

Table 2. presents the percent improvement in rate obtained by using a customized (minimum time) layout (resulting from 300 generations of the algorithm) over each of the alphabetic, QWERTY and minimum distance layouts, for each hypothetical individual.

	% Improvement Using Minimum Time Layout Over		
Ratio	Alpha	QWERTY	Minimum Distance
1:1	40%	42%	≡ 0%
1:3	39	34	17%
1:5	47	40	32%

Table 2. % Improvement Obtained by Using Minimum Time Key Arrangement over Alpha, QWERTY and Minimum Distance Key Arrangements.

In spite of the fact that the standard, 'horizontal' shape of the alpha and QWERTY key arrangements stands them in good stead for the individual with greater facility moving in the horizontal rather than vertical direction, customization nevertheless brings such users a large (34 to 47%) increase in rate in comparison. The customized board also results in higher predicted rate than the minimum distance board, in particular for the user with greater bias favoring horizontal movements.

Encoding offers another dimension to consider. Directional dependencies may take on greater or less importance when a small set of keys is used. The interaction of user motor characteristics with keyboard, code and menu characteristics makes it impossible to estimate in advance the benefit to be derived from customization. It is necessary to quantitatively assess the dependence of the client's movement time on device parameters, and calculate the predicted rate for the device design selected.

Finally, the selection of menu items is possible by means of the optimization algorithm described above. By mutating not only physical layout parameters, but also code assignment and menu choice simultaneously, further, perhaps major, improvement in the benefit to each user may be achieved.

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Keyboard Optimization Technique to Improve Output Rate of Disabled Individuals

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ABSTRACT - A set of software tools have been developed to interactively generate keyboards optimized for single digit operation. Emphasis has been placed on the creation of a flexible, modular system that imposes few constraints when designing the keyboards. Any combination of language units can make up the keyboard and the type of layout desired may be tailored to an individual's particular disability. Keyboards have been generated that reduce average between-key movement distances by 50%, with predicted communication rate improvements of 35%. Clinical trials are being pursued to verify these predictions and to solicit suggestions for enhancements.

INTRODUCTION:

Improving the communication rate of non-vocal individuals has long been a goal of rehabilitation technology. Many disabled individuals using direct selection input methods - be they for a communication device, writing system, or computer - access a keyboard-like device with a single digit, usually a headstick or finger. Numerous techniques have been devised to improve the output rate of such individuals, generally concentrating on the language units offered or produced by a sequence of keystrokes [Foulds and Crochetire, 1975; Goodenough-Trepagnier, 1982; Vanderheiden, 1984].

The technique for communication rate improvement presented herein transcends the issue of the result of a motoric act and concentrates on the efficiency of the sequence of motor acts themselves [Rosen et al., 1984]. That is, it applies an algorithm to the layout of units on the keyboard which minimizes a user-specific scoring function for the arrangement of keys on the keyboard. A concrete example of this would be minimizing the distance that an individual's fingertip would travel in the process of typing this paper using only alphabetic units and punctuation. The results may also be utilized to aid in the selection of an optimal set of language units by permitting comparision between candidate language unit menus.

There are several aspects of this problem that merit discussion:

- Determining digram frequencies for an arbitrary set of language (or output) units
- The input and evaluation of potential key layouts (i.e., what should be optimized?)
- The algorithm for creating optimum keyboards

These will be addressed in turn and some example data will be presented.

Obtaining Digram Frequency Data - Many existing communication devices and boards have a mixture of alphabetic, word fragment, word, sentence fragment, sentence and phrase units. This combination of language units renders most of the existing, letter-based, digram data [Carterette and Jones, 1974; Solso, 1979] and word counts [Kucera and Francis, 1967; Dahl, 1979] useless for evaluating actual communication device keyboard layouts . A program called ANALYZE was written to generate digram frequency data for an aribtrary set of language units.

ANALYZE takes as input a file containing the units to be present on the keyboard and a text file of typical output that will be produced by the keyboard. ANALYZE then simulates producing the output sample using the given menu of keyboard units by using a technique of matching the longest keyboard unit with the initial letters of the string being produced [Goodenough-Trepagnier et al., 1982]. This process, referred to hereafter as text decomposition, is shown in Figure 1. for a simple, 9 unit keyboard. The result of this process is a digram table, data describing the frequency that each language unit is used after every other language unit [Solso et al.,1979]. This data is a complete description of an efficient way the user could produce the target text using the available language units.

Sentence:	Candidate Units:		
THIS IS THE END			
HIS IS THE END	R	S	E_
IS IS THE END			
IS THE END	END	IS_	_ (space)
THE END			
HE END	T	H	I
E END			
END			

Figure 1. Decomposing text into keyboard units.

Evaluation of Potential Keyboard Layouts - Given a digram frequency table and an arrangement of a keyboard offering those language units, it is possible to model the process required to produce the original text on that keyboard. For instance, to produce the sentence in the above example, the digram frequency table would show that it was necessary to go from the **T** key to the **H** key two times, from the **H** to the **IS_** key once, and so on. Knowing how far those keys are from one another on a keyboard we wish to evaluate, as well as how long it takes an individual to move that distance, we can predict how long it will take that individual to produce the example sentence with that keyboard. Furthermore, if the text that was decomposed to create the digram tables was sufficiently representative and large, the prediction would be indicative of that individual's communication rate with the device [Rosen et al., 1985].

EVALUATE is the program written to evaluate potential keyboard layouts. It operates on a digram frequency file output from ANALYZE and a keyboard description file consisting of language units and their locations. The program then evaluates the keyboard by calculating the following

double sum:

$$\sum_{i=1}^{n}\sum_{j=1}^{n} frequency_{ij} \cdot evaluation function_{ij}$$

where: — n is the number of language units
— frequency data output from ANALYZE

The evaluation function may be as simple as the distance between the subsequent keys, however, a more informative measure about the keyboard would be a modelled time required to produce the digram. This time may be calculated for a user assumed to be obeying Fitts law [Bennett et al., 1954], or by using a more sophisticated user-specific modelled time per character from motor-testing data analysis. It has been demonstrated that production rates predicted using the latter correlate well with actual production times [Rosen et al., 1985].

Optimizing the Keyboard Layout - The final program attempts to find an arrangement of the language units on a keyboard that will minimize the score described above. It is impractical to simply evaluate all of the possible keyboard layouts and pick the one having the best score due to the astronomical number of permutations of keyboard layouts. An alphabetic, 28 key layout has 28! or more than 100,000,000,000,000,000,000,000,000,000 different potential arrangements. Exhaustive searching is not a viable means of determining the optimum keyboard [Getschow, 1983].

The author has devised an algorithm for finding a solution that is at least close to the exact or exhaustive solution. The situation described above makes it impossible to actually determine the correctness of the solution, however the technique results in solutions equivalent or superior to those created by other proposed techniques [Getschow, 1983]. The algorithm converges quickly, a keyboard can typically be optimized in a half-hour using an IBM PC® and techniques for improving this further are being investigated.

The algorithm starts with the language units randomly arranged on an oversized keyboard. The uppermost, leftmost keyboard location is examined. If exchanging the contents of that location with the contents of any of its nearest neighbors produces an improvement in the score for the keyboard, then the contents are interchanged. Next, the neighboring location is examined. This is done until all the locations on the oversize keyboard have been evaluated and exchanged when appropriate. This process is then repeated over the entire keyboard until exchanges are no longer occurring. The algorithm has the effect of "walking" the keys into optimum adjacencies.

The user has the option of viewing the intermediate keyboards or simply getting the final result. The result is output in a form ready to be run through EVALUATE so that its performance can be evaluated reletive to potential keyboards having other language units or on bodies of text other than those from which it was generated.

RESULTS & DISCUSSION:

To reiterate, the process that may be clinically used to construct a keyboard layout involves this sequence of steps:

A user inputs a text file of language units that is believed to be appropriate for a client's language abilities. This is done using any text editor that produces ASCII text files. The language units are input to the file one per line, any sequence of up to 255 characters per language unit is valid.

ANALYZE uses these language units to decompose a large file of target text that should resemble, as closely as possible, the output the client will produce with the keyboard. This process will create a digram frequency matrix for the client's language units producing the particular text file chosen.

This digram frequency matrix is then used by OPTIMIZE to generate a keyboard for the client. The function that the keyboard will be optimized for is specified by the clinician. It would be best at this point to have access to data that predicts the user's key-to-key movement times [Rosen et al., 1985]. In the absence of this data a Fitt's law person model or shortest distance optimization may be used. This process will produce a near optimum keyboard for the digram data / user model combination chosen.

This keyboard will be output to the screen, printer, and/or a data file that may be run through EVALUATE. This will be most useful when several language unit sets are being considered and need to be compared. The clinician choses the keyboard with the best score and can make a communication board based upon it.

Table 1. presents some example data about existing keyboards from ANALYZE. The time data is from a Fitt's law person model. Three alphanumeric keyboards are presented as well as the Write-100 units [Goodenough-Trepagnier et al., 1982] and an alphanumeric/word layout that includes the 15 most commonly occurring words over 2 letters in length [Dahl, 1979]. These simulated keyboards were 25.4 mm keys on 25.4 mm centers, it was assumed further assumed that activation time was negligible, i.e. when the modelled individual touched the key it was activated.

Name of Keyboard	mm/stroke	mm/char	sec/stroke	sec/char
QWERTY	84.4	84.4	.140	.140
DVORAK	102.2	102.2	.155	.155
ALPHA	90.9	90.9	.145	.145
WR_100	145.4	102.7	.184	.130
HYPO_1	82.3	73.5	.135	.121

Table 1. Results from ANALYZE.

Table 2. shows the effects of keyboard optimization on the language units analyzed above. Significant reductions in time/character have been realized in every case. The three alphanumeric keyboards all

converged to the optimized alpha layout. This argues strongly for the correctness of the algorithm.

Language Units	dist/char (mm)	reduced (%)	time/char (sec)	reduced (%)
ALPHA	45.3	50.5	.094	35.2
WR_100	51.9	49.5	.089	31.5
HYPO_1	47.8	35.0	.093	23.1

Table 2. Results from OPTIMIZE.

The optimization tools described are powerful, yet improvements may contribute to their being more functional in a clinical setting. It may be desirable to be able to specify constraints during the optimization process. For instance, we may want to require that the alphabet be a contiguous block of keys and have any words or digits that may be included arrayed around this block. The need for these features will be assessed during clinical evaluation.

Another potential problem in the operation of these programs involve the nature of the text used to generate the digram frequency data. If a keyboard having only single-letter units is being generated, it is not critical that the text file closely resemble the desired output. The variation, on the letter frequency level, from one source of text to another, is not significant. As the keyboard units get longer, however, the variations in the soure text may have a major effect on the generated layout. A sufficiently large corpus of text produced by non-vocal communicators is not readily available for this purpose. Corpora of spoken English may be sufficiently similar for the generation of keyboards, but few are available in raw form. Written english corpora are definitely inadequate for the generation or optimization of keyboards with large language units.

CONCLUSION:

The optimization tools presented have the potential to dramatically increase the communication rates of disabled individuals. The tools are independent of language unit selection and may be customized to a particular user's disability, especially if analytical motor data has been obtained. Clinical trials are now necessary to determine if the predicted rate improvements will be realized in practice.

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Human Quality Synthetic Speech Based Upon Concatenated Diphones

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ABSTRACT - This paper discusses the development of a speech synthesizer for use with communication aids for the nonvocal. The synthesizer employs concatenated diphones in order to provide natural sounding, unlimited speech. An inventory of approximately 1300 diphones allows the synthesizer to use commercially available speech technology that is compatible with popular personal computers.

INTRODUCTION:

Synthetic speech in nonvocal communication systems has limitations imposed by the quality and naturalness of the speech. The primary means of accessing spoken output is the concatenation of segments of speech into meaningful messages. While the use of phonemic units allows for the generation of unlimited vocabulary, it fails to preserve the naturally occuring transitions that exist between phonemes in natural speech. Conversely, the concatenation of stored words and phrases provides very natural sounding speech but is limited by availability of items in the vocabulary.

This project has derived an inventory of concatenatable elements referred to as diphones. These are drawn from human speech and include the naturally occuring transition between two phonemes. It is these transitions which play a significant role in the naturalness and intelligibility of speech. Since diphones can be joined to produce any word in the English language with a reasonable approximation of the natural transitions between phonemes, it is possible to have a speech synthesizer that provides both unlimited vocabulary and high intelligibility.

This synthesis technique has been implemented using the speech coding technique of Linear Predictive Coding (LPC). This has allowed the project to make use of existing LPC synthesis devices that are available for personal computers. Thus eliminating the need for the development of additional hardware.

TRADITIONAL SPEECH SYNTHESIS:

Recent years have witnessed a dramatic growth in the availabilty of speech devices whose outputs range from very robotic to very human-like. This difference in quality is due largely to the degree to which the interphoneme transitions match those of human speech (Olive and Spickenagel, 1976). Traditional phoneme synthesis, as used in low cost synthesizers based upon the Votrax SC01, form words from a set of 40 to 60 phonemes or allophones. In most instances, this is done with only minor attention given to the naturalness of the transition between the elements. Foulds and Bronk (1981) have shown that many transitions are, in fact, quite unnatural and contribute to the robotic character of the speech. Higher quality synthetic speech is available from devices such as the Infovox SA101 (Magnusson et al 1984) and the Digital Equipment Corp. DECTalk (Bruckert 1984). These provide dynamic models of the phoneme transitions and produce considerably higher quality output.

THE DIPHONE CONCEPT:

The concept of creating a synthesizer based on diphones was first proposed by Peterson, Wang and Silvertsen (1958). They observed that the frequency spectrum of a phoneme near the boundaries with preceding and succeeding phonemes was substantially altered by the transitions between the individual phonemes. This variation was shown to be contextually determined. Thus, the change in the spectrum of a phoneme depends upon the particular phonemes which surround it. It was also observed that the spectrum of the phoneme was least altered near its mid-point. Peterson et al suggested that the phoneme was not an appropriate unit for concatenation since it was not possible to provide smooth spectral transitions between all possible combinations. They proposed that diphones, which begin at the steady-state (from the frequency domain perspective) mid-point of one phoneme, include the natural transition, and end at the steady-state mid-point of the succeeding phoneme, would be a more appropriate unit. This would require up to 1600 diphones (all possible combinations of 40 phonemes) rather than simply 40 phonemes. It would, however, avoid the necessity of accomodating the spectral alterations in the process of concatenating the units into a speech message.

Diphone speech is nearly as intelligible as synthesis natural speech. Dixon and Maxey (1968), Olive and Spickenagel (1976), Schwartz et al (1979), Elsendoorn (1984), and Varga (1985) have demonstrated the production of intelligible speech using diphone inventories on large computers.

SELECTION OF A DIPHONE INVENTORY:

The selection of a diphone inventory for this project has taken place over several years. Thibault and Foulds (1982) describe the process whereby the possible number of diphones was reduced to approximately 1000. This eliminates combinations of diphones that do not occur in spoken English. That work also introduced the use of longer polyphones that include one or more complete phonemes between the boundaries of a pseudo-diphone.

During this process, the identified diphones were each recorded in a carrier word by a trained speaker. This provided a natural environment for each diphone from which it could be extracted.

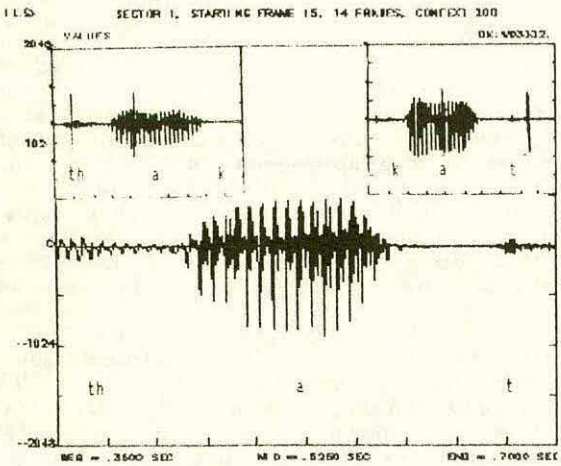
LABORATORY ENVIRONMENT:

The development of the diphone inventory was accomplished at the speech research laboratory of the Rehabilitation Engineering Center. This facility is equipped with audio recording equipment and a PDP11/34 minicomputer for signal processing and analysis. Primary software used on the PDP11 is the Interactive Laboratory System (ILS) which allows complete manipulation and editing of a speech signal as well as LPC analysis and synthesis.

DIPHONE INVENTORY IMPLEMENTATION:

The recordings of the carrier words containing the imbedded diphones were digitized onto the PDP11, and each diphone was extracted using ILS. In order to determine the rules for diphone extraction, a "first pass" diphone inventory was created by fixing the endpoints of a diphone at zero-crossings corresponding with the central points (in time) of the two consecutive sounds. The diphone transition was placed in the middle.

Next, work was done to refine and complete the inventory. Several guidelines have become apparent from the evaluation of the synthetic speech obtained through the concatenation of analyzed diphones and others are being developed.



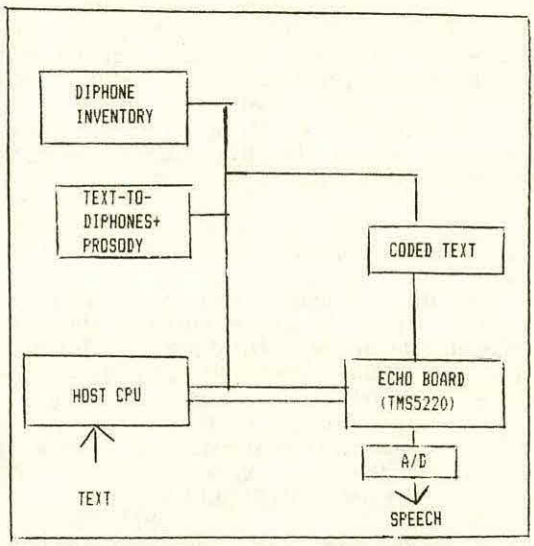
Diphone concatenation

IMPLEMENTATION ON A MICROCOMPUTER:

The ECHO board (a Streets Electronics voice synthesizer containing the Texas Instruments signal processing chip, TMS5220) requires LPC parameters for the production of synthesized speech. The LPC encoding allows the 1300 diphones to be stored in a small amount of memory. In addition, the naturalness and intelligibility of the synthetic speech improves because LPC coding preserves the short-time spectral characteristics of the original speech and smooths the spectral parameters at the junctions of the concatenated diphones.

The analysis of the diphones to obtain the LPC parameters (reflection coefficients, energy, and pitch) was performed on the PDP11/34 using the ILS software package. The voice synthesizer containing the Texas Instruments TMS5220 chip is interfaced to an IBM-PC. Coding of the ILS LPC parameters to obtain the 5220 LPC parameters is required because of the non-linear relationship between the two sets of parameters. Further packing of the coded parameters was needed to meet the format requirements of the voice synthesizer.

Continuation of this work will include the adaptation of a text to speech algorithm provided by Dr. Peter Maggs of the University of Illinois. Since this code only transforms typed text into phonemes and not into diphones, an algorithm to transform phomemes into diphones is being developed. This text to speech algorithm will also be altered to allow for elementary prosodic features, including stress, rhythm, and intonation.

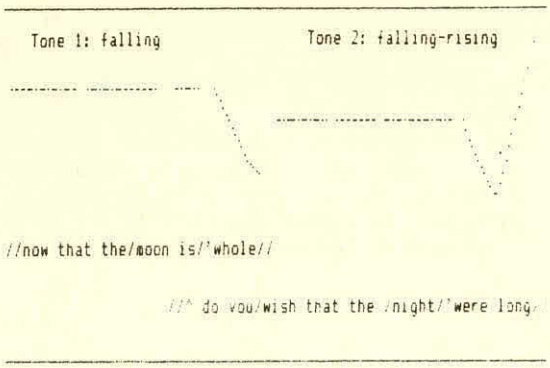


Text-to-speech using diphones

EXPERIMENTS IN PROSODIC CONTROL:

Currently, the project work on intonation rule development is focusing on applying a variety of intonation contours by hand in order to determine the basic contours themselves. Using M.A.K. Halliday's (1970) five intonation groups as models, diphone sentences are given the contour (or contours) that best approximates that used when a live subject speaks the sentence.

The diphone sentence "Now that the moon is whole, do you wish that the night were long?" contains two tone groups separated by the comma. Variations of Halliday's intonation groups 1 and 2 were hand superimposed on both tone groups, respectively.



Intonation applied to a diphone sentence

Tone 1 (falling) has a variable pretonic (the part before the stressed foot) with a basic form of mid to mid-high and level. The major sentence stress of the tonic starts where the pretonic left off and falls to a low pitch. In "now that the moon is whole" the pretonic is high and meets the beginning of the tonic. Tone 2 (falling rising (pointed) or high rising) has a high level pretonic which meets a tonic that falls then rises or jumps to a tonic that rises sharply. For "do you wish that the night were long" the tonic falls then rises to far above its original level.

Once the best fit forms have been determined by hand, the next step is to standardize them into several basic forms and a small, finite number of variations. Given a text to speech algorithm that lets the user specify the basic form s/he wants (using puncuation), the program must find the tonic and decide which, if any, variations apply.

CONCLUSION:

The diphone speech synthesized on the PDP11 closely resembles natural speech. Listeners have no difficulty in understanding spoken messages, and are capable of recognizing the original speaker who produced the carrier words. While more work is required to add sentence prosody in order to improve naturalness, the diphone sentences have been found to be nearly indistinguishable from sentences made from stored words. Thus, the project has achieved its goal of providing the intelligibility of stored speech while maintaining the flexibility of unlimited speech.

The synthetic speech produced by the ECHO board is highly intelligible although the speech quality is reduced. The loss of quality is due to the fixed bit rate and frame duration required by the TMS5220. The use of a new synthesis component, however, the Texas Instruments TMS320 signal processor, will highly improve the intelligibity of the speech since there are no limitations on the bit rate or frame duration. This synthetic speech could then rival that of the PDP11.

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EFFECT OF A LUMBAR SUPPORT ON SEATED BUTTOCK PRESSURE

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The purpose was to compare ischial tuberosity pressure when sitting with and without a lumbar support. An adapted pedobarograph (ischiobarograph) was incorporated into an adjustable chair. A TV camera detected changes in light levels from the underside of a plexiglass sheet. The signal was passed to a video analyzer board and converted into 3 colors. Each color was calibrated to represent the following pressure intervals: white-80 mm of Hg and greater, red-40 to 80 mm of Hg, and blue-20 to 40 mm of Hg. Subjects (N=10) were seated on the ischiobarograph and recordings of the pressure contours were taken before and after the placement of a lumbar support. Two trials were completed for each subject. An intra-class correlation coefficient between trials was .98. A paired t-test demonstrated a significant decrease in the white isobar in both trials (p=.0008, p=.0007) following lumbar support placement. The use of a lumbar support may facilitate a more ideal sitting posture for high risk pressure sore patients.

INTRODUCTION AND BACKGROUND

Pressure sores represent one of the major complications facing disabled individuals, especially those with spinal cord injuries. Zacharkow (20) suggests that sitting posture may be a major contributing factor leading to the development of pressure sores.

The incidence of pressure sores as reported by Clark (5) is 21.6 percent for paraplegia and 23.1 percent for quadriplegia. Geisler (9) reports over 4 percent of deaths among the spinal cord injured have been directly attributed to pressure sores. Amyloidosis resulting from chronic skin ulceration (7), can lead to renal failure (11) which is the leading cause of death in the chronic spinal cord injured population. The estimated medical cost associated with the healing of a pressure sore ranges from $10,000 to $46,000. Young and Burns (19) determined that during the first four years following injury the spinal cord injured patient with pressure sores averaged $15,000 a year more in medical costs than those without sores.

The ischial pressure sore has been found to be the most prevalent (6,8), and it has been determined that more pressure sores are due to sitting than to the recumbant position.

Kyphotic sitting can eventually lead to asymmetrical sitting postures characterized by spinal scoliosis and pelvic obliquity, factors which appear to be closely associated with uneven sitting pressure distribution and the eventual production of a pressure sore (14,15). If the normal standing posture (lordosis) is maintained in sitting, advantages may include decreased disc pressure (2), relaxation of the lumbar paraspinal muscles (1), prevention of kyphotic and scoliotic spinal deformities with pelvic obliqueties (12), improved diaphragmatic breathing (3), and less fatigability with increased comfort (10,19).

The inherent wheelchair problems, namely the built-in seat to backrest angle, does not allow proper use of a lumbar support (20). Yet, the seat to backrest angles that would provide appropriate use of a lumbar support must be designed so as not to hinder a spinal cord injured individual from becoming functionally independent (transfers, chair wheeling, etc.) The 95 degree seat to backrest angle used clinically by this investigator and recommended by Zacharkow (20) appears to allow maximal use of the lumbar support without altering the patient's functional potential. At this angle, if a lumbar support does not significantly increase the buttock (ischial tuberosity) pressure (isobars), then it may represent a more suitable sitting posture for the spinal cord injured population. Because of the serious consequences resulting from a pressure sore in the spinal cord injured patient, a sitting posture that potentially enhances the development of an ischial sore could never be justified regardless of other conceivable benefits of this posture.

The purpose of this study was to compare the effect of a lumbar support on the area of highest buttock pressure (ischial tuberosity) at a 95 degree seat to backrest angle in normal subjects when sitting with and without a lumbar support.

MATERIALS AND METHODS

Subjects

Four males and six females were included in this study. Subjects were volunteers meeting the following criteria: 1) able bodied; 2) 20 to 50 years old; 3) overall good health; 4) no history of low back pain in the last 6 months; 5) no evidence of scoliosis; 6) no noticeable limitation in lumbar extension when back bending in the standing position.

Instrumentation

The barograph used for measuring seat pressure in this study is an adaptation of a pedobarograph designed by Chodera and Lord (4) for displaying pressures under the foot. A television camera detects changes in light levels from the underside of a mat covered plexiglass sheet illuminated at its edges. These light levels are converted to different colors, each representing contours of constant pressure (isobars) under the foot.

The ischiobarograph (17) for measuring sitting pressure consists of an 18 by 18 inch plexiglass sheet (1/2 inch thick) incorporated into a seating device that can be adjusted to different thigh trunk angles. The device is a plywood box

with an adjustable seat and backrest with the seat surface being the plexiglass sheet. Two domestic 13 W fluorescent tubes 21 inches long are located at the optically polished edges and covered with a dense cloth to prevent stray lighting.

An angled mirror (45 degrees) within the box deflects the optical path of the detecting surface to a horizontally mounted television camera. The output from the camera passes to a video analyzer board residing in one of the expansion slots of an IBM personal computer equipped with a dual disc drive, and color monitor. This analyzer contains a microcomputer which converts the video signal into four colors, each corresponding to different grey scale light levels, and is controlled by menu driven software. Samples of buttock pressures were taken at a rate of three pictures per second. The seated pressure pattern is plotted on the IBM color monitor and can be stored on floppy diskette. Utilizing an IBM printer, the pressure pattern was converted to hard copy. The pressure areas were then traced with a planimeter (Apple Versawriter) to quantify the area corresponding to each color.

Calibration of the barograph was achieved by using a penumatic pressure cuff incorporated into a wooden box. When placed on the transducer seat and inflated, equal pressure was applied throughout the pressure cuff bladder and the pebbled baromat was compressed into the plexiglass sheet. A manometer attached to the pressure cuff simultaneously displayed the amount of pressure in mm of Hg. The pressure intervals consisted of the following: 1) White-80 millimeters of mercury and greater; 2) Red-40 to 80 millimeters of mercury; 3) Blue-20 to 40 millimeters of mercury. The highest intervals, white and red, were the primary focus of this study. A second white pressure interval was adjusted to represent 30 mm of Hg and greater. The purpose for this adjustment is discussed under procedure.

The lumbar support consisted of 1.75 inches of firm polyurethane foam (durafloat II-10 pound density and 140 pound indentation load deflection), and a less dense 1.5 inch polyethyelene foam (4 pound density and 70 pound indention load deflection). The support was 4 inches in height, 15 inches long, and 3.25 inches thick uncompressed.

Procedure

The investigative process involved three tirals and three positions:

Position 1: Sitting without a lumbar support and recording that pressure pattern at the end of 45 seconds.

Position 2: Sitting with a lumbar support and recording that pressure pattern at the end of 45 seconds.

Position 3: Sitting without a lumbar support and recording that pressure pattern at the end of 45 seconds (same as position 1).

Trial 1 involved positions 1 and 2, while trial 2 involved positions 1, 2, and 3. Following the completion of trial 1, the subject was asked to move from the barograph for 1 minute, then return and be repositioned for the commencement of trial 2.

Trial 1 began when the subject assumed a seated posture (position 1) with the backrest reclined 15 degrees and seat inclined 10 degrees, producing a 95 degree seat to backrest angle. The feet were supported by an adjustable footrest so the knees were at an 80 degree angle and the thighs (femur) parallel with the seat surface. The ankles were maintained at 100 degrees using the tibia and fifth metatarsal as landmarks for measurement. All angles were determined by goniometric measurement. The posture of the subject (position 1) was the relaxed sitting posture, defined as the subject having his/her sacrum positioned at the back of the seat and in contact with the chair back, eyes fixed horizontally, and hands and forearms resting on the thighs. This pressure contour (isobar) was recorded and stored at the end of a 45 second period.

The subject was then tilted forward just enough to allow placement of the lumbar pad (position 2) so that the lowest edge was approximately one half inch below the posterior iliac crests. This placement was performed by palpating the posterior superior iliac spines to attain the adequate position of the support. This placement duplicates the clinical procedure utilized by the investigator when modifying wheelchairs with lumbar support. This sitting pressure pattern was recorded and stored following a 45 second period.

The subject was asked to move from the ischiobarograph, then return and be repositioned for trial 2. Trial 2 exactly duplicates the trial 1 procedure with one exception. Following the last recording with the lumbar support (position 2) the support was removed, and another recording taken at the end of a 45 second period. This served to test any ordering effect regarding the lumbar support placement.

Trial 3 involved positions 1 and 2, with the barograph adjusted so that the white isobar represented a pressure of 300 mm of Hg and greater. Each subject was seated in position 1, as described earlier, to determine if the ischial tuberosity pressure equaled or exceeded the 300 mm of Hg pressure when sitting without a lumbar support. Maintaining the 300 mm of Hg setting, each subject assumed the sitting posture with the lumbar support (position 2) to determine if the ischial tuberosity pressure was greater or less than 300 mm of Hg.

Methods of Analysis

The reliability of the ischiobarograph, planimetry tracing of subjects, and testing procedure was determined by an intraclass correlation between trial 1 and trial 2 differences. Also included in this correlation was position 1 for the two trials. The intraclass correlation coefficients were tested at a .05 level of significance.

A two sided paired t-test (16) was done to compare the white isobars of position 1 to the white isobars of position 2 for both trial 1 and trial 2. This was also completed for the red isobars in each trial. A two sided paired t-test compared position 3 to position 1 (trial 1 and 2) to determine any ordering effect of the lumbar support.

RESULTS

An intra-class correlation coefficient between trial 1 and trial 2 white area differences was .87, and the red area differences was .95. The intraclass correlation between the two trials for the white area before lumbar support placement was .93, and the red area before lumbar support placement was .99. All of these intraclass correlations were significant to the .05 level indicating the instrumentation, procedure, and planimeter tracing was highly reproducible between trials.

Eight paired t-tests were completed simultaneously for the white and red pressure areas by the Bonferroni method (16). This adjustment was used because of the inability to assume equal variances among the white and red area measurements. The white isobar was significantly decreased ($p < .05$) after lumbar support placement (position 2) as compared to before placement (position 1) in both trial 1 and trial 2. The red isobar showed no significant differences for similar comparisons in both trial 1 (.41) and trial 2 (.42). The white and red isobar demonstrated the same levels of significance when the order of lumbar support placement was changed.

DISCUSSION

The significant decrease in the white isobar when sitting with a lumbar support demonstrated that the area of highest pressure was consistently reduced when the lumbar support was used. Because all subjects' initial ischial pressures were above 300 mm of Hg in trial 3, the results indicate a major reduction in the white isobar following placement of the lumbar support. Minns (13) reported average maximal peak pressures under the ischial tuberosities of normals seated on a rigid surface to be 489 mm of Hg. This study concurs with Minns (13) in that peak pressures on a rigid surface was found to be greater than 300 mm of Hg (the highest level tested). Because the white isobar (80 mm of Hg and greater) was significantly reduced with a lumbar support, it would indicate that ischial pressures over 300 mm of Hg were reduced down to and below the 80 mm of Hg level. If the spinal cord injured population demonstrate the same effects of a lumbar support on seated pressure as the able bodied population, the benefits could decrease human suffering and reduce the economic strain involved in the treatment of the ischial pressure sore. Further studies using the spinal cord injured population have been instituted by this investigator and could prove informative as to the application of lumbar supports for this patient population.

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Automated Workstations to Teach Prevocational and Vocational Skills to the Profoundly Retarded

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ABSTRACT

This poster session presents an automated system to be used for teaching basic environmental awareness and pre-vocational skills to severely and profoundly retarded individuals. The paper describes a graduated task system that progresses from simple contingency awareness to simulated workshop tasks. The hardware consists of an Apple IIe, industrial 96 I/O interface, and interchangeable work boards. The software uses a multitasking format that controls 6 work stations simultaneously. The program sets up the tasks on the work boards, monitors responses, operates various devices for reinforcement, has intelligent functions to alter response requirements on line, and prints out data and progress notes. The poster session presents and overview of thi system, a working model for examination and examples of student progress.

PAPER

Until recently little attention has been given to teaching cognitive and prevocational skills to severely and profoundly retarded individuals. This has been especially true for individuals who are also physically handicapped. This began to change as Sidman & Stoddard (1967) demonstrated that these individuals could learn when provided with consistant teaching experiences. These findings have been expanded to show that cognitive (Butterfield & Belmont, 1977) and prevocational skills (Gold, 1972; Bellamy, Horner & Inman, 1979) can also be mastered but for many of these individuals this learning process can be very slow and must include intense, consistant, and frequent teaching sessions which require almost superhuman patience. Recently Hanline, Hanson, Veltman, & Spaeth (1985) have shown that this training can be better provided by using micro switches to mediate responses and environmental events. While this has worked well in some limited areas the lack of a sophisticated and intelligent interface has greatly limited the possibilities of this strategy (Chubon, 1985).

For the past 8 years Rast and his associates have been developing programs and hardware to provide cognitive and prevocational training for these individuals (Rast & Williams, 1982; Rast & Williams, 1983; Willams & Rast, 1985; and Rast, 1985). During the past 2 years, the controlling devices have shifted from electromechanical relay equipment to an Apple IIe with an industrial I/O interface. By using multitasking software and a 96 I/O interface, one computer, in the present pilot systems can control six simultaneous classroom workstations or 10 residential environmental awareness units. In the most basic case simple contingency awareness is learned when a single switch, adapted to the physical capabilities of the resident, operates an event in the resident's environment. For example, a kick pad might activate a vibrating pad in a daybed. As this cognitive skill is mastered two switches may be introduced to teach choice and differentiation. At higher levels students are required to match colors, symbols, and sounds. As the student progresses the tasks become: following instructions by sorting objects and assembling simple objects to simulate workshop tasks. Since all of the problems and responses in these sessions are mediated by the computer the student receives the kind of consistant contingent feedback necessary to master the skills. Automated switch systems offer a number of advantages over using a trainer to do the same kinds of things. First, because the closure of a microswitch is an absolute event, and, because the switch does not have anything more interesting to attend to than whether or not it is closed, the immediacy and accuracy of the feedback given to the learner is much better from the switch than from the trainer. This is especially true when teaching extremely handicapped individuals who may wait long periods of time between responses. Second, because switches are less expensive than the extra staff necessary to monitor large numbers of sessions, it is possible to provide more cost effective training while also giving the student more opportunities to learn in a mechanized environment. Third, the measurement and programming capabilities of the control equipment allows the teacher more time with the students and the more accurate data facilitates more sensitive training resulting in faster learning by the student. Finally, the fact that the handicapped individual directly controls their environment adds dignity to their lives.

One teachers aide with one Apple IIe computer can run simultaneous sessions for a class of 6 students. The students sit at simulated work stations working on automated task boards. The boards in each booth are easily removable so different tasks can be used in the same booth on the same day. The teacher stays between the booths where he or she can

see the students through one-way mirrors while watching the students and monitoring the equipment and assists the students in the work booth or waiting area. When the sessions are completed the computer signals completion and store the data on the data bank disk. Based on this data the teacher selects goals for the student according to the IDT recommendations.

The training tasks for the prevocational and cognitive training areas are designed to provide a slow but steady increase in the difficulty of response requirements. The initial training sessions involve simple contingency shaping in responding to the keys and then to the stimuli that are intended to provide discriminative control. The initial task is simply to press a 2 inch lit key. From that point the task will branch according to the eventual goals of the training.

For assembly goals the students progresses through simple chained responses, to second order schedules, to conditionally linked chains to simple assembly tasks. This entire progression can be controlled by the computer through expert functions in the software. This includes on line decision branches for response patterns, one to one attention for all training trials, complete documentation of training, and periodic progress notes and charts. The teacher can modify these decision branches to control the direction of the training by using present menus.

For sorting goals the progressions of training tasks go from simple discriminations, to matching to sample, to more difficult matching (e.g., delayed, conditional, and/or increased sample sizes), to second order schedules, and finally to simple sorting tasks. This includes fading of the programmed edible reinforcers, fading to a workshop environment, and increased task requirements to simulate actual work.

Experience with over 100 residents has shown this to be an excellent way to establish motivation to work at a task and to stay on the task for long periods of time without external prompting. These same tasks can also be used as the baselines or response modalities for testing. Using visual or auditory cues can serve as a baseline for vision and hearing examinations, and the repeated acquisition and delayed matching to sample conditions are excellent baselines to assess the effects of drug dosages on learning and performance.

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A Multidisciplinary Approach to Feeding Assessment at Gillette Children's Hospital

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ABSTRACT

The goal of the physician, occupational therapist, speech/language pathologist, and nutritionist working directly together in the Feeding Clinic at Gillette Children's Hospital is to maximize an individual's potential to ingest food or self-feed and to maximize an individual's potential in growth. After identifying the individual's needs in the areas of medical, motor performance, positioning, adaptive equipment, oral motor functioning, and diet, it is then necessary to make specific recommendations to further these goals. It is often necessary in order to achieve the team's goals to involve the orthotics department in problem solving or fabricating specialized adaptive equipment. Because of the many causes of feeding problems, the expertise of professionals trained in many different areas is needed to facilitate comprehensive assessment and effective treatment.

INTRODUCTION

The Feeding Clinic specializes in servicing any person who has motor problems which interfere with eating or drinking. This may be a result of cerebral palsy, brain damage, muscle problems or a variety of related conditions. Since these individuals often have unique difficulties in many areas, it is beneficial to have a team who looks at various aspects of the feeding process and then presents a cohesive, rational approach taking into consideration each other's aspects.

BACKGROUND

This clinic was established in 1984 since a need was identified to meet the demands in these areas for this population. The usual interdisciplinary process of the feeding evaluation includes a medical assessment by a developmental pediatrician and a comprehensive feeding assessment and intervention by a team consisting of an occupational therapist, speech/language pathologist and a nutritionist.

METHODS OF EVALUATION

The physician involved in a feeding program needs to be aware of conditions likely to lead to feeding difficulties. In the person with neurologically-based problems, these may include neurological/structural abnormalities of the mouth, throat, or esophagus. Allergies or food intolerances may result in vomiting of food which has already been taken and will need to be distinguished from gastroesophageal reflux and other causes of vomiting.

The role of the occupational therapist in the feeding clinic is to observe, evaluate and make recommendations in the areas of motor performance, positioning, and adaptive equipment. Motor performance issues range from head and trunk control to upper extremity usage. Areas of evaluation include muscle tone, muscle control and mobility, hand to mouth pattern, positioning of utensils/food and dominance. Positioning concerns involve the necessity for proper alignment for more efficient swallowing. A break down of this includes physical support, symmetry, midline orientation, general mobility and a need for seating devices. Objectives for good positioning include 1) Protection of the airway, 2) greater control of bolus (masticated food), 3) greater control of the bolus location and 4) reduction of tone/reflex abnormalities.

The speech/language pathologist's primary responsibilities on the feeding team are to assess the clients' oral function skills, determine feeding techniques which will facilitate improvement in areas of deficit, and then teach these techniques to the primary caregiver. Specific areas of oral function assessed include; oral reflexes, lip mobility, tongue mobility, chewing, sucking, swallowing, and drinking.

The nutrition assessment includes diet analysis, anthropometric (height and weight) measurements, clinical assessment, and laboratory assessment.

The orthotics department is called upon to help evaluate the child's needs in the areas of specialized equipment for positioning and adaptive equipment for eating, assisting the caretaker and/or the individual in more independent feeding.

Additional considerations are made by all the team members and these include evaluation of the following: gradation of stimulation, environmental factors, awareness of fears, cognitive ability and the ability to cooperate or contribute to the feeding process, comforts and precautions to be taken, and ability of caretaker(s) to carry through.

EVALUATION PROCESS

When a child is referred for a Feeding and Nutrition Evaluation, it is first determined by team members whether the referral is appropriate and what services will be needed. In addition to the actual feeding and nutrition session, many patients are also seen

by such professionals as a developmental pediatrician, pediatric gastroenterologist, and orthotists. Swallowing studies are performed when indicated.

The patient is initially observed by the feeding team either being fed or self-feeding a meal brought from home. The occupational therapist and speech/language pathologist then assess the various positioning, equipment and feeding technique needs by working directly with the patient. At this time varied food textures may be introduced and orthotists may be called in to consult. Once optimal techniques and positioning determinations have been made, the caretaker is taught how to employ them. The nutritionist then takes a comprehensive nutrition survey and makes dietary recommendations.

All Feeding and Nutrition Evaluations are followed with a written report detailing both observations and recommendations. Follow-up consultation and evaluations are scheduled as appropriate.

RESULTS

After evaluating the individual in all these areas each discipline in conjunction with the team makes their recommendations.

In the medical areas, abnormal structure, muscle control, or function of the extremities may require specialized techniques to accomplish feeding ranging from medication to surgery to the possibility of discontinuing oral feeding due to aspiration.

The occupational therapist's goals can be achieved by special seating devices, handling techniques and special considerations which require precautionary measures. Adaptive equipment needs are considered to help achieve independence, improve stability and control, and aid the caretaker in maintaining positioning considerations. These range from a lap tray, foot supports, special utensils and selective cups, bottles, or dishes which are usually commercially available.

The speech/language pathologist's intervention techniques may include changing food and/or fluid consistency, utilizing oral facilitation techniques, and restructuring mealtime.

With information gathered during the feeding evaluation, the nutritionist may make diet modifications in the areas of route of feeding (gastrostomy and/or oral feeding), food texture or consistency, specific food restriction(s), caloric intake and nutrient supplements which are formulated to maximize an individual's growth potential.

To help complement and further the individual's feeding goals, the orthotist is called upon to devise and fabricate specialized equipment specific for an individual. This is especially true in instances when ordinary measures or commercially available products do not meet the patient's needs and more technical expertise is required.

Typical concerns and possible solutions often presented at the time of the feeding evaluation include the following examples:

1. Individual with poor head control. These children often require added measures of support for their head such as a head rest and/or seating device which has a slight recline. This aids the child in keeping his head upright and in midline to facilitate proper swallowing. Added difficulties which accompany this problem are difficulty in chewing, swallowing and lip closure. Oral facilitation techniques are demonstrated and instruction is given to the caretaker to carry out these methods to be used during feeding. Additional calories are often added to the diet to supplement any loss which may occur due possibly to tongue thrust or lack of coordinated swallowing, etc.
2. Difficulties in trunk control or extensor thrusting often require specialized seating devices which have support pads, trunk straps and/or pommels to maintain proper orientation and inhibit automatic reactions or primitive reflexes.
3. Vomiting or reflux problems may require special fabrication of positioning devices such as a prone stander to decrease intra-abdominal pressure during and after feeding. Surgical procedures such as a Nissen fundoplication may be necessary to construct a mechanical barrrier to the reflux of gastric contents into the esophagus. A combination of tube feedings and oral feedings is determined to prevent weight loss, encourage weight gain, and to continue oral feeding so that these skills are maintained.
4. For individuals with motor incoordination but a hand to mouth pattern, specialized utensils, bowls, dishes, or cups are issued. It is necessary at times to involve orthotics in creating a mechanical device to assist and accomplish a more efficient pattern of hand to mouth movement.
5. Modified equipment can prevent hyperextension of the head as in a cut out cup and allow for better swallowing techniques.

CONCLUSION

A team approach is a logical resolution to complex feeding problems. Interdependency of services is necessary to meet individual needs. The procedure utilized by the Feeding and Nutrition Team at Gillette Children's Hospital has proven to be an efficient method of conducting a comprehensive evaluation during a single clinic visit.

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Communication System for a Severely Physically Handicapped Individual with Visual Impairment.

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ABSTRACT

The progression of three communication systems developed to meet the changing needs of a severely physically handicapped and visually impaired individual is presented. The systems utilized Morse code, visual scanning, and audio scanning techniques. The capabilities and limitations of each system and the problems encountered in implementing them are discussed.

INTRODUCTION

Providing a nonspeaking, physically handicapped, visually impaired person with a means of communicating is a difficult task. For the recently injured individual whose functional capabilities are affected by fluctuating medical conditions, the task is even more challenging. This paper examines a unique case at the University of Michigan Medical Center where a series of communication systems were developed for a high level spinal cord injured patient with very limited motor function.

PATIENT BACKGROUND

The patient, R.M., is an 18-year old male who sustained a C1 level spinal cord injury extending to the fifth cranial nerve due to a motor vehicle accident. He is a ventilator-dependent quadriplegic with no voluntary motor control except for eye movement, eyeblink, and some chin movement.

R.M. was assessed for possible use of a computer-based communication system while on a Roto-bed in a neurological intensive care unit. Cranial nerve damage resulted in the loss of lateral eye movement, loss of tongue and forehead movement, minimal movement of the lower left facial muscles, very minimal and slow jaw movement, blepharospasms, and rapid eyelid fatigue. It appeared that R.M.'s hearing was within normal limits. No cognitive deficiencies were noted, although formal cognitive testing was not performed.

R.M. demonstrated good control with eyeblink. He communicated "yes" by blinking once and "no" by blinking twice. He had vertical eye movement but loss of lateral eye movement resulted in a limited field of vision. His visual acuity could not be accurately assessed due to lack of communication capabilities.

After the initial evaluation, Speech Pathology and Rehabilitation Engineering collaborated to develop a communication system that R.M. could use effectively. Auditory feedback in the system was essential due to R.M.'s poor vision. Since eyeblink was R.M.'s only controllable movement, a control interface activated by eyeblink was necessary.

METHODS

Control Interface

An infrared switch (1) was used in conjunction with R.M.'s eyeblink as a control interface. The single-turn potentiometer on the switch box was replaced with a ten-turn potentiometer allowing finer sensitivity adjustment of the switch. The switch sensor was mounted on a pair of eyeglass frames so that an eyeblink activated the switch (figure 1).

R.M. was able to activate the switch on cue. At times, however, extraneous eye movements and blinks would unintentionally activate it. To alleviate this problem, a variable delay (0 - 0.5 seconds) was built into the switch. With the delay, the eye had to remain closed for a certain length of time before the switch was activated. The delay time was adjustable by setting a second potentiometer on the delay box. This delay allowed some eye movement and quick eyeblinks without producing undesired switch activation.

Figure 1. R.M. using infrared switch

The infrared switch with an added delay proved to be an effective interface that R.M. could activate consistently with eyeblinks. Therefore, the infrared switch was the control interface used throughout R.M.'s progression of communication systems.

Morse Code System

The first communication system provided was set up in R.M.'s room in the intensive care unit. It consisted of an Apple IIc computer, a monitor, a Votrax Personal Speech System, an infrared switch, and software that allowed alternate in-

puts to run a computer (2). Morse code, one of the specialized input routines available, was incorporated as the input method. Software was developed to provide editing capabilities and voice synthesis.

Text was entered to the computer by inputting Morse code via eyeblinks to the infrared switch. A quick blink sent a "dot" to the computer while a longer blink sent a "dash". Auditory feedback in the form of differing tones distinguished between the two signals. Each character was spoken by the Votrax and printed on the screen after it was input. Inputting the Morse code for a SPACE combined the previous letters entered to form a word, while the code for ESCAPE combined the previous words entered to form a sentence. The BACKSPACE code served to clear the last character entered and the BACKSLASH code cleared all the entries stored in memory.

Speech Pathology and Rehabilitation Engineering worked with R.M. for several weeks, but it soon became apparent that a communication system using Morse code was not the most appropriate for him. Rapid eyelid fatigue associated with R.M.'s cranial nerve damage greatly affected his ability to enter Morse code using eyeblinks. It was noted at this time, however, that R.M.'s vision seemed to be improving from his initial evaluation. Thus a communication system utilizing visual scanning was investigated.

Visual Scanning System

R.M. required a communication system that would utilize his improved but still poor vision and accommodate his loss of lateral eye movement. A one dimensional scanning program that seemed appropriate for R.M.'s needs had been written previously in the Rehabilitation Engineering division for a non-speaking, visually impaired individual (3). This software, along with the same hardware used for the Morse code system, comprised R.M.'s second communication system.

The program utilizes colorful graphics and oversized characters in a linear scanning array ranging from 1 to 9 rows. It has a menu-driven, branching structure in which character strings in each row of the array are created specifically for the user. String entries represent either text messages or menu names. A flashing arrow scans through the array until a selection is made. If a message is selected, it is sent to the speech synthesizer; if a menu is selected, that menu replaces the current one on the screen. The scanning rate is set at the beginning of the program.

The original program was modified to include four lines of text below the graphics array. Each selection was printed at the bottom of the screen as well as spoken to eliminate the need for constant attention by the person with whom R.M. was communicating.

R.M.'s main menu array contained groups of letters, the symbols "SP R < CL", and the message "WORDS". Selections were made using the infrared switch. The groups of letters represented other menu arrays containing each letter of the group as individual entries in separate rows. The symbols "SP R < CL" stood for a menu array containing the entries SPACE, RETURN, BACKSPACE, and CLEAR that worked in the same manner described earlier. The entry "WORDS" represented another menu array containing other menus for commonly used words (i.e. "NAMES" or "RECREATION").

The visual scanning system was used successfully by R.M. Although he could not see an entire message if it extended into the far right side of the screen, he was able to tell what message it was by the part he could see. His first message since his injury six weeks earlier was "THANK YOU. I LOVE YOU ALL". Additional arrays were added to the program as R.M. began to communicate more. As his proficiency increased, the scanning rate was increased. R.M. enjoyed using the system and communicated several hours a day with his family, friends, and nurses. One difficulty encountered in implementing this system was that R.M. had to be sitting up in bed in order to see the screen. His opportunities to communicate were thus limited since his medical condition required that his position be changed throughout the day.

Unfortunately, after effectively using the visual scanning system for about three months, R.M. developed severe corneal abrasions due to loss of natural tearing. These impaired his vision so greatly that he was no longer able to use the visual scanning system to communicate.

Audio Scan System

A new communication system using an Apple IIe computer was consequently developed that employed auditory feedback and did not rely on R.M.'s vision (figure 2). A flexible two-dimensional audio scan program was written. This program was interfaced to software employing a modified implementation of a "ten branch abbreviation expansion" technique (4). The abbreviation expansion software was written in the "C" programming language (and was originally developed at our center for another patient).

The customized software is fully programmable in both its audio scanning and abbreviation expansion aspects. The audio scan system is a two-dimensional scanning array where the first entry of each row is spoken as it is scanned. Activation of the control switch after an entry is spoken initiates an audio scan through the row. A second switch activation selects a specific row entry which is printed on the screen. The scanning rate is programmable. This audio scan system is similar to one described by Beukelman et al. (5). Control functions allowing words and sentences to be formed as in R.M.'s previous systems are included. Additional functions in this system include "SLOW", "FAST", and "STOP" for controlling scanning rate. Three eyeblinks start up the system again after "STOP" is selected.

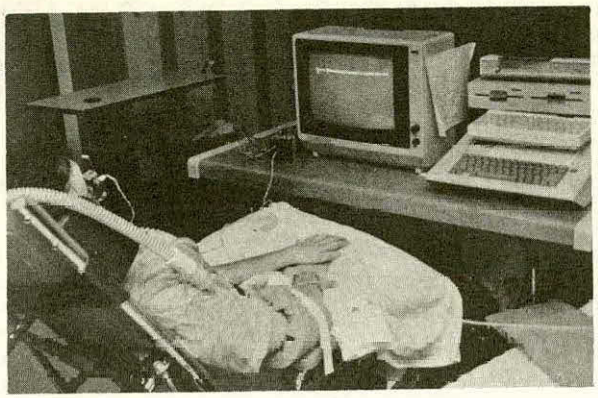

Figure 2. R.M. using the audio scan system.

The abbreviation expansion aspect of the system permits an increased communication rate by accessing a large stored vocabulary with a reduced number of key selections. Abbreviations are coded in the form of one or more letters followed by a number. As letters are typed, abbreviation menus are displayed in the top portion of the screen with the numbers 0-9. When the number code for an abbreviation is input, the code is automatically converted to the text stored under that abbreviation. The system features the ability to independently output text to a voice synthesizer, printer, or remote computer.

DISCUSSION

The audio scan system with abbreviation expansion has greater flexibility and more control functions than the visual scanning system. It can also be used from any position in a bed or wheelchair, as long as the voice synthesizer can be heard. R.M.'s use of the audio scan system, however, has been affected by several factors. He was slow to accept a new system after being so successful with the visual scanning system. and has difficulties hearing the voice synthesizer when many people are around or when nearby televisions and radios are on. Also, the audio scan system on the Apple IIe is large and cumbersome.

Speech Pathology is working closely with R.M. to develop an efficient coding system for storing words and phrases that will enable him to better recall abbreviations. R.M. is actively involved in choosing the words and phrases he wishes to have stored as abbreviations in the system. Presently, R.M. uses the audio scan system twice daily during sessions with Speech Pathology and Rehabilitation Engineering, but does not use the system on his own. He prefers to have a person verbally scan through an alphabet matrix and "blink out" the letters he wants one by one.

Additional applications of R.M.'s communication system are being explored. One such application is the use of the audio scan system to run standard word-processing software on a second computer. This would give him the ability to write, store, and print text.

Fortunately R.M. has shown some cranial nerve return over the past several months since his injury. He has regained some upper and lower facial movement and also ocular movement. This is promising as it greatly expands the communication options available to R.M. Rehabilitation Engineering and Speech Pathology are continuing to work together in hopes of providing R.M. with a flexible communication system that will provide him with the motivation necessary to use it effectively after his discharge.

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A COMPUTER INTERFACE FOR A SEATING PRESSURE EVALUATOR

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ABSTRACT

Functional electrical stimulation is under study as a method of preventing decubitus ulcers in immobilized individuals. This work requires the ability to measure pressure changes in real time. A microprocessor-based interface has been developed which allows a general purpose microcomputer to acquire and store data from a two-dimensional pressure sensor array.

INTRODUCTION

Pressure sores are a major problem for individuals who are wheelchair dependent, particularly those with sensory losses. Many researchers have investigated the causes of these pressure sores and have sought methods of preventing them. A project at the University of Michigan Medical Center (UMMC) will investigate the use of functional electrical stimulation (FES) as a means of preventing pressure sores (1). Part of this research involves measuring and recording the dynamic forces exerted at the seating interface.

The pressure measurement instrument used at UMMC is the the Texas Interface Pressure Evaluator (TIPE). It consists of a pressure monitor pad and a separate display unit (2). The pad is connected to the display unit by a ribbon cable. This system is capable of measuring surface pressure and presenting the experimenter with a transient visual picture of the conditions present at the TIPE pad. However, there is no provision for permanent recording of observed data. Many experiments require information about pressure behavior to be recorded over an extended period. This recorded data must also be converted to a form which prepares it for computer analysis.

An electronic interface has been constructed to replace the TIPE display unit and to act as a direct link between the TIPE sensor pad and a computer. The interface converts the state of pad switches to a digital format which can be read and recorded by any general purpose microcomputer. The digitization and transmission of all switches occurs 30 times each second.

BACKGROUND

The standard commercial version of the TIPE system consists of two parts:

Sensor pad. This is a square pad with 144 pressure sensitive switches arranged in a 12 x 12 matrix across its surface (Figure 1). Individual switches are separated by 1.25". The pad consists of separate top and bottom sheets of translucent vinyl plastic bonded together around the outside edge. This forms an air-tight cushion which can be inflated to a desired pressure through an attached hose. Electrically conductive strips bonded to the inside of the pad connect the switches to form rows and columns. The switch at each intersection consists of a small rectangle of resistive material. It is oriented so that an electrical connection with a resistance of 100K ohms is formed between the row and column whenever the top and bottom surfaces make physical contact. The switch closure occurs when the externally applied pressure exceeds the constant internal pressure of the pad. At any moment the open and closed switches define a two-dimensional pressure map of the pad surface.

Display unit. The electronics and the visual display are mounted in a metal cabinet of shoe box size. On the face of the display box is the gauge of a manometer. This is a purely mechanical pressure measurement system connected to the TIPE pad by flexible tubing. A hand-held squeeze bulb is used with this gauge to set the pad pressure. TIPE pad rows and column are connected to the display through a 24-wire cable. The visual display is a 12 x 12 matrix of light emitting diodes (LEDs) mounted on the front panel of the display unit. The LEDs reflect the state of corresponding pad switches: lighted for closed switches, dark for open switches.

MATERIALS

The TIPE interface is assembled on a single 5" x 6" circuit board and is installed in a box with a small DC power supply (Figure 1). This interface box replaces the TIPE display unit and is connected to the TIPE pad with the original cable. A separate squeeze bulb and pressure gauge from a sphygmomanometer replace the same parts of the standard display unit. Information passes from the interface box to the computer over a two wire cable. This cable link follows the RS-232 standard for serial data communication. The communication protocol used by the channel is: 9600 baud, 8 data bits, no parity bit, one stop bit.

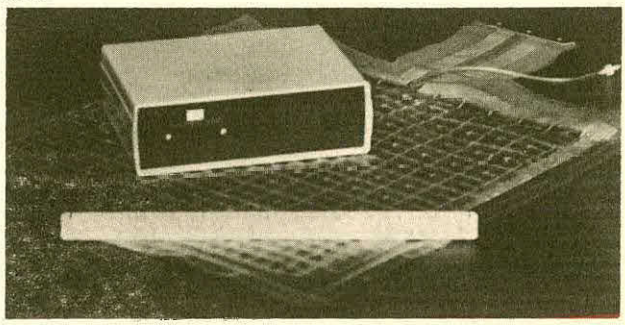

Figure 1

The interface is based on a Hitachi HD6303 Microcomputer Unit (MCU). This is a single chip microcomputer which includes 128 bytes of random access memory (RAM), a programmable timer, a serial communication interface, and 13 programmable input/output lines in a 40-pin package (Figure 2). The HD6303 MCU utilizes the same machine language instruction set as the Motorola 6800 series microprocessors (3). It is controlled by a program stored in a standard 2048 byte erasable programmable read only memory (EPROM). Timing for the serial interface and the internal sampling rate timer is derived from a single crystal-controlled oscillator running at 4.9152 MHz. The MCU addresses individual switches of the TIPE pad through a circuit composed of digital latches and analog switches. The analog response from the pad is converted to a digital signal by a voltage comparator.

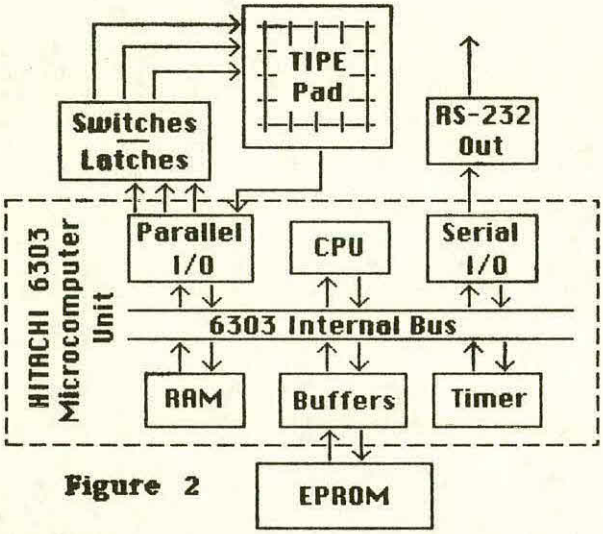

Figure 2

THEORY OF OPERATION

When power is first applied to the TIPE interface automatic reset circuitry causes the MCU to execute a series of initialization instructions. After initialization the system needs no further external intervention. It will send TIPE pad information continuously, one complete set every 33 msec. This is a sampling rate of 30 Hz.

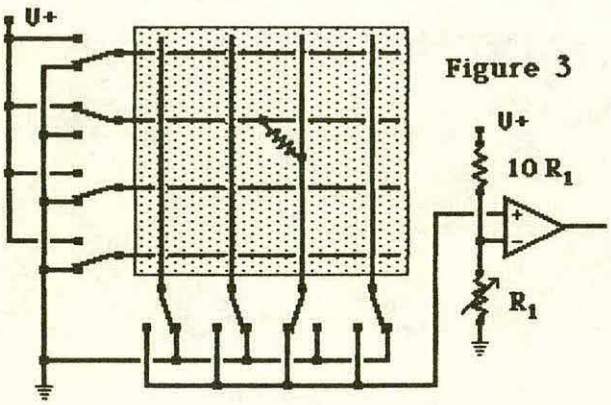

Figure 3

The conversion routine checks each one of the 144 pad switches to determine whether it is open or closed. This procedure utilizes three circuits. The first is a set of three 8-bit latches (Figure 2). The MCU stores 24 bits of information in these latches to set the conditions necessary to read one pad switch. The output of the latches is connected to the second circuit. This consists of 24 single pole, double throw (SPDT) analog electronic switches (Figure 3). Each row and column is connected to the common terminal of one of these switches. Each latch output controls one analog switch. Using this setup the MCU can connect individual pad rows either to the positive supply voltage (V+) or to ground (GND). Pad columns can be connected either to the special pressure switch read circuit or to GND. In the read circuit the voltage from the column being tested is compared to a constant reference voltage. The output of this stage is a digital signal indicating that the pad switch is closed (high) or open (low).

The MCU begins the process of reading a specific pad switch by setting the row and column latches. The row to be tested is connected to V+ (Figure 3). The column to be tested is connected to the voltage comparator read circuit. All other rows and columns are connected to GND. If the pad switch at the intersection of the active row and column is closed the row voltage (V+) will cause the column voltage to rise. If the pad switch is open the column voltage will remain at or near GND. The MCU reads the state of the pad switch from the output of the voltage comparator. This state is stored and transmitted as a binary 1 (switch closed) or 0 (switch open).

The microprocessor reads pad switches in groups of 6. The results of these 6 reads are assembled into a 6-bit binary number in which each bit represents one switch. The serial transmitter which sends data to the remote computer does so 8 bits (one byte) at a time. To convert from 6 bits to 8 bits the MCU adds two bits to the top of the number. The MCU prepends the bit sequence "01" to the first transmitted byte of each sample set, and the sequence "10" to every other byte. The remote computer uses this difference to synchronize itself with the TIPE interface.

The two processes of reading switches and sending data occur simultaneously. Transmitting 24 bytes of formatted information is slower than reading the same amount of raw data. The fastest transmission rate commonly available on computers which support the RS-232 standard is 9600 baud. At this speed data transfer takes 25 msec. A choice of 33 msec per sample accommodates all necessary processing and leaves a cushion of 5 msec before the next sampling process begins.

SWITCH MATRIX DECODING

The control circuit of the TIPE interface is a straightforward application of the HD6303 microproccessor chip. In the past the same basic circuit has been used as the main intelligence in a variety of devices, including a stand-alone Morse code to RS-232 interface similar to the TIPE

interface. The novel aspect of the current application is the need for a method to test the state of individual switches in a large X-Y switch matrix. The TIPE pad is designed so that a closed switch forms a 100K ohm connection between a row and a column. The simplest way to check for a switch closure is to measure the resistance between each row and column. Any measurable resistance (less than infinity) means that the switch is closed. This method works for single switches but when many are closed the scheme is unable to distinguish between various patterns. Figure 4a shows a portion of the switch matrix with one switch closed. The resistance between the row marked R and the column marked C is 100K ohms. The circuit in 4b will show the same resistance between R and C even though the connecting switch is now open. There are three distincts electrical paths from R to C, each having a resistance of 300K ohms. The total equivalent resistance of the three parallel paths is 100K ohms. The only way that the control circuit can choose between these two variations (and numerous others) is to record the resistance between every pair of rows and columns and then solve a large system of equations for the state of each switch.

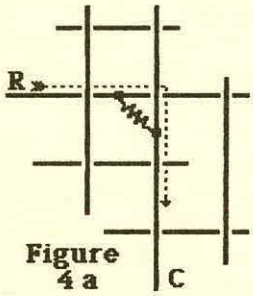

Figure 4a

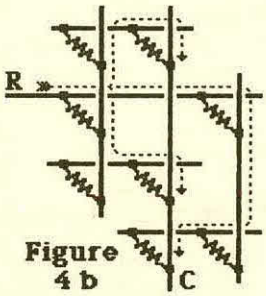

Figure 4b

This difficulty of multiple electrical paths is avoided in the TIPE interface by adding the SPDT analog switches previously described. Before reading a switch, the microprocessor connects all unused rows and columns to GND. This eliminates all paths except the one that leads directly from the active row to the active column. The voltage at the comparator is dependent upon the number of closed switches in the active column. If the switch being tested is open the column voltage will be at or near GND. If only that switch is closed the column voltage will be V+. However, each additional closed switch in the column provides a parallel path to ground, thereby lowering the column voltage. In the worst case, with all 12 column switches closed the column voltage will be V+ divided by 12. The reference voltage used by the comparator to test the column must be set above GND but still below this worst case voltage.

IMPLEMENTATION

This system is currently being used in the laboratory to study the characteristics of seating surfaces. The TIPE pad and interface are connected to the main serial port of a Zenith Z-150 computer (device COM1:). The Zenith is both software and hardware compatible with the IBM-PC. Data aquisition software written for the Zenith receives switch matrix sample sets from the interface and stores them for later analysis. Information can be recorded at rates from .01 to 30 samples per second. Lower sampling rates are achieved by discarding a portion of incoming switch data. At lower rates the recording time of an experiment may be extended to several hours. The parameters such as sampling rate and duration are defined by the experimenter using a configuration file. This is a text file which can be edited to accommodate changes in experimental design.

During sampling the computer continuously updates a console screen display. This display includes time information in numeric form and a 12 x 12 array of 1/4" squares which mimic the standard LED display unit. An individual square shows bright red when the pad switch it represents is closed. An open switch is represented by a faint blue square. This feature makes it possible to test the pad and interface during preliminary setup and to monitor the system during actual sampling.

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MAN-LIFT ATTACHMENT FOR PARAPLEGIC TRACTOR OPERATORS

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ABSTRACT

The design and development of a tractor man-lift attachment for handicapped operators, which uses an electrically-driven lift screw, is described. Plans for local construction are being prepared.

INTRODUCTION

In 1979 a project was started at Purdue University to assist farmers with serious physical handicaps who wish to continue active involvement in their farm operations. One of the most frequent requests is for information on access to the operator station on tractors and self-propelled machines. Many farmers with arm or leg amputations, paraplegia, back injuries, or arthritis can operate a tractor or combine if they can be lifted up to the operator's seat and provided with suitable control modifications.

Several disabled farmers have had man-lifts built in local shops and are using them. Plans for a simple, safe, widely adaptable man-lift which could be fabricated in a local machine shop to fit a specific tractor or combine model would meet a long felt need for many disabled farmers.

REQUIREMENTS

1. Provide a seat to lift the operator from the ground to the cab door in a position which allows him to move into the tractor seat using hand grips.
2. Provide a seat stowed position, preferably inside the tractor outline, with minimum interference to the operator's vision.
3. Design the lift to be powered by electricity from the battery, so the lift can be operated without starting the tractor engine.
4. Provide a lift speed of about 25 mm/s (5 fpm) and lift capacity of 136 kg (300 lb).
5. Provide fail-safe lift operation, so that power or drive mechanism failure will not cause the lift to fall.

It is desirable that the design be adaptable to various tractor models and easily attached. Modifications may be required according to the individual operator's mobility, strength and physical disability.

EXISTING DESIGNS

Most of the man-lift designs which have been built and used fall into the following types:

1. Vertical-lift platform for operators who can stand but can't climb.
2. Vertical lift with slide-through seat or sling.
3. Pivoting seat on inclined longitudinal rails in front of the rear tire.
4. Parallelogram arms with a pivot seat.
5. Swingable jack-knifing boom mounted inside the cab which hoists the operator by means of a cable and sling. The boom folds into the cab to place the operator in the tractor seat.

FEATURES OF THE PURDUE-DESIGNED MAN-LIFT

The Purdue man-lift was fitted to and installed on a Case 1290 tractor. It is shown in Fig. 1. A vertical lift design witn a horizontal seat arm pivoting at a point between a vertical post and the seat was selected. As shown in Fig. 2, a seat arm length and pivot location can be selected which will allow the seat to be swung out and lowered for operator access from his wheelchair, to be swung in after lifting to place his feet inside the cab, and finally to be swung forward and partially lowered to a stowed position above the tractor front wheel. The vertical post can be located in a position that best supports its lower end and clears the side of the tractor.

This configuration is obviously unsuitable for a tractor cab having the door hinged at the front, because the opened door interferes with access from the front.

Fig. 1 Vertical screw chair-lift mounted on a Case 1290 tractor.

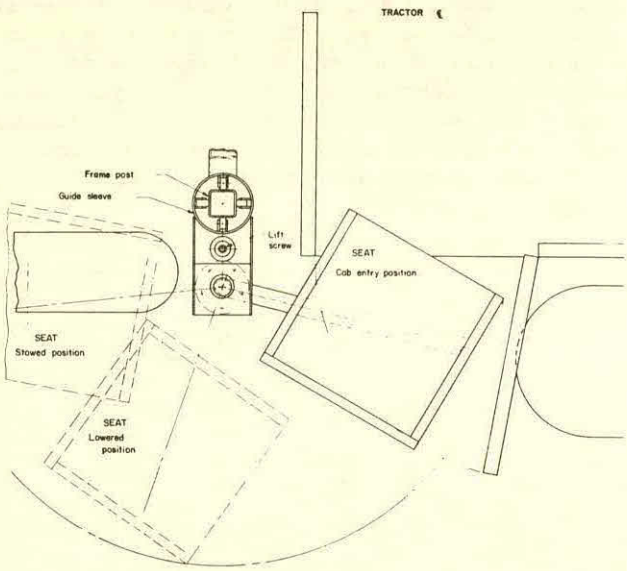

Fig. 2. Diagram of frame post, lift screw and seat pivot for partial cab entry.

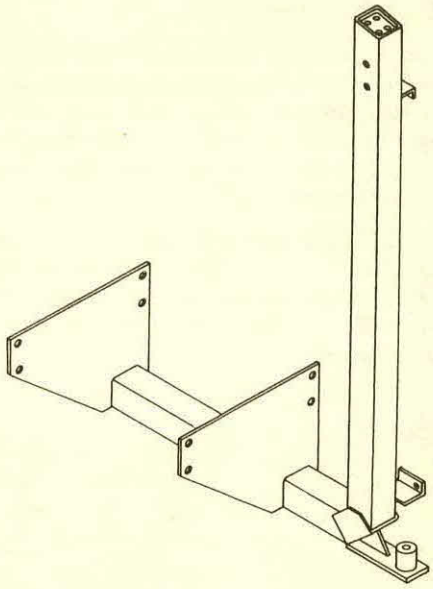

Fig. 3. Frame weldment.

LIFT TYPE

Most lifts have used small electrically powered cable winches. These are reasonable in cost, but the manufacturer usually warns that they are not to be used as hoists for people, presumably because of the possibility of cable breakage or malfunction of the automatic brake.

Some have used hydraulic cylinders or motors powered by an electric motor and a separate hydraulic pump. These, however, are rather expensive.

An electrically powered vertical screw was selected, because it can automatically hold a load from dropping when power or a driving element fails.

FRAME

The frame is a 76 mm (3 in) square, 6.35 mm (1/4 in) wall steel tube fabricated into an L-shape, as shown in Fig. 3, with the short leg passing across under the tractor engine and welded to 4.76 mm (3/16 in) thick mounting plates, which bolt to the tractor frame on each side. A stop bar and a lift screw bearing plate are welded to the base of the vertical leg. A bolting plate with four tapped holes is welded inside the top of the vertical tube to support the lift screw and motor bar.

Calculations for a 136 kg (300 lb.) load in the chair give a maximum tube stress of 34.47 MPa (5000 psi) in bending and 23.44 MPa (3400) psi in torsion, indicating a very adequate factor of safety. Deflection was minimal when using the lift.

GUIDE SLEEVE

The guide sleeve around the vertical tube is 153 mm (6 in) Std. pipe x 203 mm (8 in) long. Ears are welded inside the sleeve to carry eight straddle-mounted, 9.53 mm (3/8 in) bore, needle cam follower bearings, which guide the sleeve up and down the square tube and prevent rotation about the tube.

A 4.76 mm (3/16 in) thick U-bracket is welded to the sleeve, as shown in Fig. 4. It supports a wheel hub with tapered roller bearings to provide the seat arm pivot. It also retains the nut for the lift screw, which passes through the U-bracket at a point between the sleeve and the pivot hub.

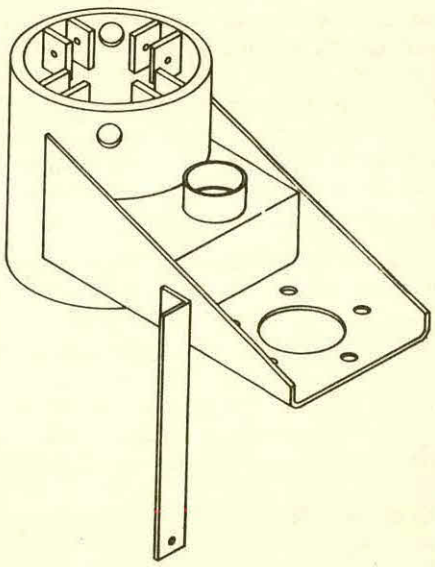

Fig. 4. Guide sleeve weldment.

SEAT AND SWING ARM

The seat is made of 1.52 mm (16 Ga.) steel with a 203 mm (8 in) high back. The seat arms are made of 25.4 mm (1 in) square x 3.05 mm (11 Ga.) wall tubing; and the right arm pivots at the top rear to allow it to be swung up and back, so that the operator can slide over into the seat from his wheelchair.

A 51 mm (2 in) square x 4.76 mm (3/16 in) wall tube is welded to a 41.3 mm (1 5/8 in) dia. wheel spindle at the inner end and carries the welded-on seat and gussets, as shown in Fig. 5. The wheel spindle and hub comprise the swinging seat pivot.

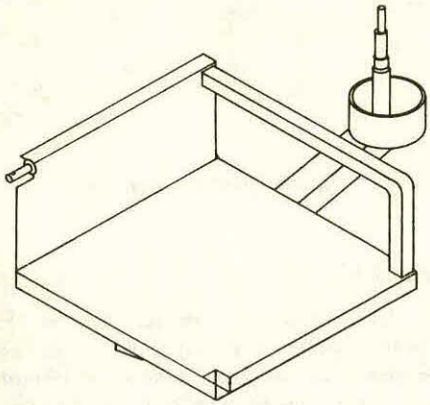

Fig. 5. Seat and arm weldment.

LIFT ACTUATION

The lift is powered by a permanent magnet 12V DC electric motor, developing 0.129 kW (1/6 hp) at 1800 r/min (rpm). It drives a 15.88 mm dia x 1524 mm (60 in) Acme thread lift screw at 500 r/min (rpm) by means of a 3L V-belt. A 9.5 x 88.9 mm (3/8 x 3 1/2 in) support bar is bolted across the top of the vertical frame tube and carries the motor on one side and the lift screw on the other. A sealed self-aligning, flange mounted ball bearing supports the lift screw. The lower end of the lift screw is stabilized by a bearing block welded to the lower frame. The lift screw nut is tapered to fit into a countersink with a fabric washer in order to inhibit vibration and allow self-alignment.

The lift screw is lubricated by high viscosity grease. Dust is excluded by a pair of bellows which can retract to 10% of their extended length. Their 51 mm (2 in) I.D. ends are clamped to mounting rings at the top, on each side of the U-bracket weldment and to the bearing block at the bottom. The bellows are interconnected by drilled holes to allow air interchange during raising and lowering, thus avoiding introduction of dust.

LIFT CONTROLS

One double throw, double pole, self-centering toggle switch is provided on the seat and another on the tractor dash, either of which may be used to raise or lower the lift. Self-centering switches are necessary to prevent the possibility of both switches being on at once and blowing the protective fuse. Momentary toggle switches act as limit switches for upward travel, for downward travel for operator access and for downward travel to the stowed position above the front wheel. A wiring diagram is shown in Fig. 6.

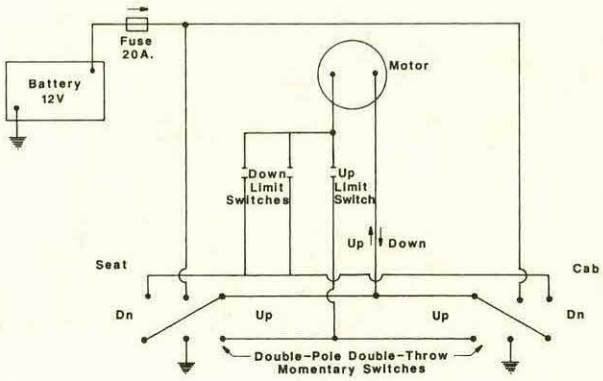

Fig. 6. Wiring diagram.

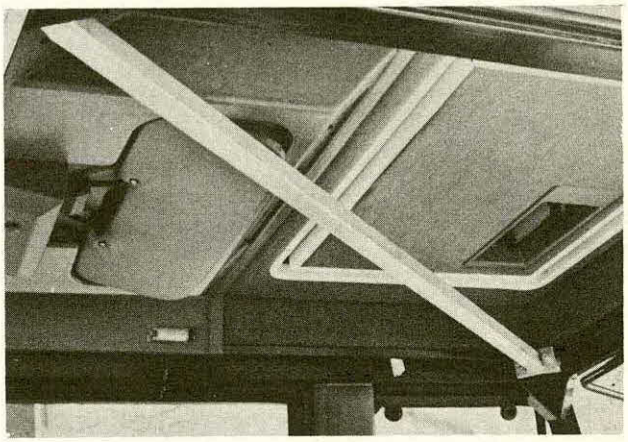

Fig. 7. Overhead handgrip bar.

MOUNTING AND DISMOUNTING PROCEDURE

To mount the tractor, the operator positions his wheelchair beside the lowered lift seat. After removing the left wheelchair arm and swinging back the right lift seat arm, he can slide over to the lift seat. He then actuates the seat switch to lift to the level of the tractor seat, opening the door on the way. He controls the swing position of the seat by grasping the door; and after lifting, he pulls the seat to the latched position, partially through the door. He can then grasp the diagonal overhead hand-grip bar (Fig. 7) to move into the tractor seat. Pulling on a webbing control strap attached to the seat arm near the pivot (Fig. 8) releases the latch,

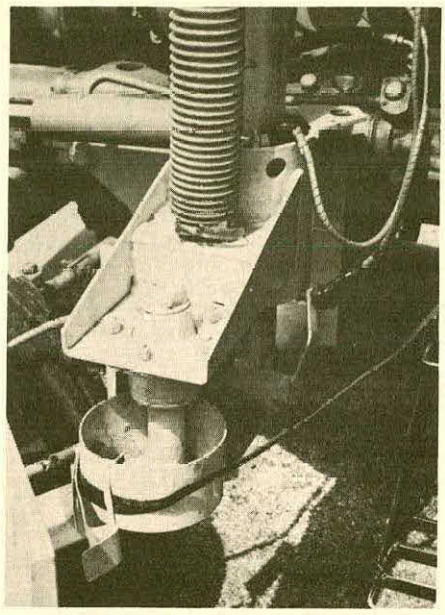

Fig. 8. Seat latch and control strap.

and a return spring swings the seat to its forward position. The operator uses the dash-mounted switch to lower the seat to stowed position above the tractor front wheel and then shuts the cab door.

To leave the cab, the operator opens the door, raises the lift seat and pulls the control strap, which latches the seat in position at the door. He then moves into the lift seat, using the overhead hand-grip bar, releases the latch and lowers to the ground, where his wheelchair is within reach.

HAND ACTUATION OF CLUTCH AND BRAKES

Separate hand levers were devised for the two brake pedals. The lever foot plates hook over the tops of the brake pedals and are held in place by a small bolt with a thumb nut for quick detachment. It is desirable to retain the lock-together feature of the brake pedals for road travel and also desirable to retain the option of actuating the brake pedals by foot when necessary, even though the manual levers are in place.

The clutch hand lever must be quickly detachable to enable the operator to enter and exit. It was found possible to design a foot plate which hooks over the top of the clutch pedal and shifts out, so that it is stabilized by the foot clutch shank. The clutch should be capable of locking in the disengaged position to free the operator's left hand for other purposes when simultaneous clutch release and brake actuation are required, as when stopping on an incline. A simple latch utilizing a hole in the vertical front wall of the seat pedestal is shown in Fig. 9.

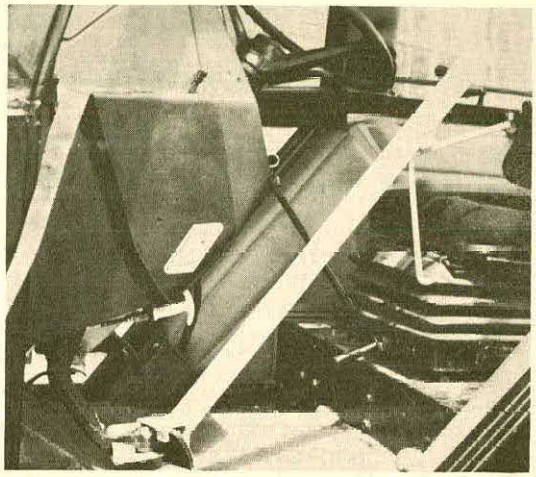

Fig. 9. Clutch hand lever with latch.

OTHER ACCESSORIES

A full view tractor mirror can eliminate the need for turning the upper torso to monitor trailing implements and is especially beneficial to farmers with back injuries. Other types of mirrors can be used to aid hitching and monitoring of various field operations with minimal body movement.

Steering devices such as spinner knobs allow for more positive, one-handed control of the steering wheel.

Automatic hitching devices allow hitching and unhitching of drawbar implements from the driver's seat with just the pull of a rope or cable attached to a spring-loaded pin. A universal spring-loaded tongue lift assembly keeps the implement hitch elevated to a level at which it can be funneled into the drawbar. Some automatic hitches use a hydraulic cylinder.

PLANS

Man-lift plans with detail drawings and photographs are being made available at a nominal cost. A man-lift for a combine can be built using the same basic principles.

ACKNOWLEDGEMENTS:

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The J.I. Case Co. donated the 1290 tractor on which the lift was mounted and tested.

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THE DEVELOPMENT OF A PRISM COMMUNICATION DEVICE

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ABSTRACT

This paper describes the identification of a clinical problem, a technical solution, and clinical application of a multi-faceted communication display designed to increase independence of use and vocabulary space on a wheelchair lap tray when it is used as an augmentative communication display. Severely physically disabled, non speaking users with good voluntary control of a single switch can independently access the three sides of the prism communication device in order to access up to 540 one inch graphic items.

INTRODUCTION

Nonspeaking physically disabled individuals who use graphic systems of communication are usually severely restricted in the number of vocabulary items which they can independently access due to space restrictions.

Conventional communication displays on wheelchair lap trays can typically accomodate 300-400 items if the symbols, pictures or words are 3/4 inch to 1 inch size and closely spaced. Solutions to date to increase space for additional vocabulary have been auxiliary displays. These are frequently constructed in the form of books or pages to allow more vocabulary to be used in addition to a main display (1). Although more vocabulary is provided, the augmentative communication user is dependent on the communication partner to provide the display. The augmentative communication user is thus no longer independent to initiate and develop communication (2). The non-speaking user is usually dependent upon his able-bodied partner to make the addditional displays available and turn to the appropriate page as the user requests. These auxiliary displays are often lost or misplaced thus suddenly restricting the augmentative communication user's vocabulary. This clinical problem led to the development of a communication device which would provide greater vocabulary area and be independently accessed by the user thus decreasing dependency on the communication partner.

BACKGROUND

In 1982 the Rehabilitation Engineering Department and the Augmentative Communication Service established a team at the Hugh MacMillan Medical Centre (operated by the Ontario Crippled Children's Centre) to bridge the gap between clinical, technical and research issues in the field of augmentative communication, to assure practical solutions to clinical problems and to develop and evaluate prototype devices. Clinicians and engineers embarked on a collaborative project to develop and evaluate a multi-faceted communication display to allow independent access and increased vocabulary space within the wheelchair lap tray in response to the problem identified above (see Figure 1).

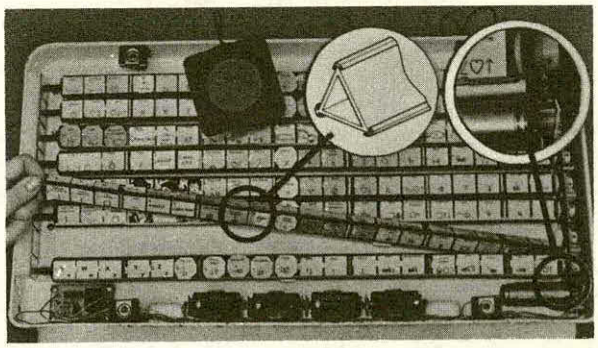

Figure 1. Prototype Prism Tray with Lexan Cover Removed

DEVICE DESCRIPTION

The principle design requirements presented were as follows: maximum overall display area for vocabulary items, independent access to the device via a single switch, easy access to the display surface to allow modification of the vocabulary as required, easy mounting to any wheelchair and adequate durability and reliability of the device.

The display portion on the prism communication device consists of a series of triangular prisms on which 480 one inch words, pictures or symbols can be attached. Triangular prisms were chosen since they maximized the number of graphic items that could be implemented in a fixed area. The prisms are mounted side by side, are interconnected via a series of gears and idlers and are driven by a small D.C. motor/gearbox. This design concept is similar to that employed in commercial billboards. The sequencing of the prisms is provided by an electronic

control interfaced to the user's interface switch and controlled by a three lobed cam located next to the prism closest to the motor. Each time that the motor is activated by the user's switch, the prism set is sequenced 120 degrees which provides the user with a different display surface and hence a different symbol set. In this manner, the user can have independent access to approximately three times the number of symbols that would otherwise be available on a single flat surface.

The overall size of the prism communication device (12 inches long x 22 3/4 inches wide x 1 3/4 inches high) took into consideration the required mounting compatibility with existing wheelchair trays.

The main housing of the device is vacuum formed white plastic. The top cover consists of a clear plastic sheet which provides a clear, durable and rigid surface. It is fastened to the main housing by means of four quarter turn fasteners which can be turned by a coin to allow for easy access to the inside of the prism communication device and the prism surfaces. Thus the prisms can be easily removed and replaced as required.

The tray system is powered by four internally mounted and replaceable standard D size batteries.

DEVICE APPLICATION

Seven augmentative communication users were given the device for 15 weeks. They were between 5 and 27 years of age; lived at home or in institution; had no serious visual problems; were nonspeaking due to Cerebral Palsy; required auxiliary displays; had severe physical disabilities which limited their page turning abilities and thus limited their independent use of auxiliary displays; accessed their displays either by direct pointing, encoding or scanning techniques and were able to use a single switch.

From clinical observations and questionnaires completed by the seven augmentative communication users and their caregivers, the following advantages and disadvantages were noted: In general, the device was found to be useful as an aid to a very specific target population of augmentative communication users. This primarily includes those individuals whose hand function is severely restricted, but who have adequate voluntary movement and control for accessing a single switch and whose involuntary reflex patterns do not interfere with their use of the switch. These are individuals who require up to 480 one inch graphic items, are not easily able to turn pages attached to the main display or retrieve additional displays independently. There were a number of positive comments regarding physical attraction which heightened interest and increased communication interaction with a greater number of listeners. Increase in vocabulary, independence and speed were also noted as advantages.

Problems identified in the prototype prism communication devices included difficulties in mounting, a need for more than 480 graphic items, lack of independent control of the on/off switch and technical failures. These technical failures included: warping of the prisms; difficulty in affixing stamps to the plastic prisms; and the intermittent accidental rotation of the prism each time the display surface was struck

CONCLUSIONS

As a result of clinical feedback, the prism communication device will soon become commercially available with the following modifications: one additional prism will be added to provide a total of 540 one inch graphic items; a unique mounting system to address the mounting problems is currently being developed to allow tilting and swing-away features; (3) the black rigid PVC material will be replaced by white noryl to prevent warping; the prisms will be grooved (see Figure 1) to allow easier insertion of the graphic material; and the motor sequencing circuit has been upgraded to eliminate unintentional activation.

This project demonstrates a technical solution to a clinical problem through a collaborative effort. Continued development projects through clinical and technical collaboration are essential in the field of augmentative communication. Only by clinicians clearly posing problems and providing support to rehabilitation engineering specialists will new and viable technical solutions to the problems faced by augmentative communication system users be addressed.

ACKNOWLEDGEMENTS

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Janice Light.

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A MODULAR MULTIPLE DRIVE/LOAD ERGOMETER FOR THE DEVELOPMENT OF AN OPTIMAL BIOMECHANICAL MODEL OF THE UPPER LIMB

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ABSTRACT

The Human-Machine Integration Group of the Palo Alto VA Rehabilitation R&D Center is keenly interested in the development of simulation tools which will facilitate the design of devices based on optimal human-machine interaction. To this end, a study is underway to develop a tractable, computerized biomechanical model of the upper limb performing optimizable ergonometric tasks, such as arm cranking for maximum work output. Experimental results, obtained through use of the modular multi-faceted arm-powered ergometer discussed in this paper, will be used to develop this biomechanical model by the application of modern optimal control theory. The model will then be used to generate optimal configuration and dimensional parameters for any arm-powered drive system, based on specified performance criteria. This paper focusses on the design of the arm-powered ergometer to be used in this study.

INTRODUCTION

A multi-faceted arm-cranking ergometer apparatus had been designed for a current study to develop a theoretical biomechanical model of the upper limb (fig. 1). This computerized model will employ optimal control theory to optimize certain performance tasks. As a first task, subjects will be instructed to maximize power output during arm cranking routines, using hand cranks adjusted to varying radii of rotation. In order to develop a more universal mathematical model, several drive and load systems will be employed, each with such variable features as crank length for arm cranking experiments, or lever length and position for oscillating lever propulsion. In addition, the apparatus will be instrumented to acquire kinematic and dynamic data, which will be analyzed together with simultaneously collected electromyogram data. Experimental results will be used first to define the theoretical model and then to refine it, as other propulsion tasks are studied.

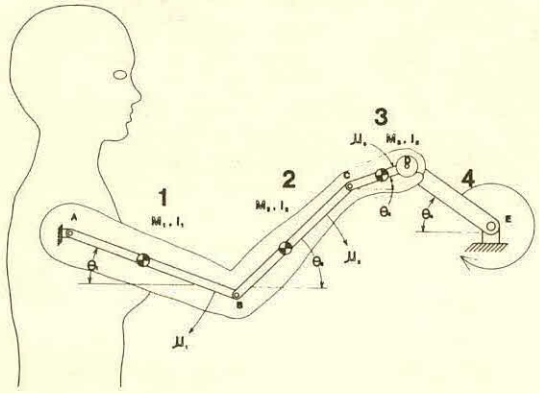

Figure 1. Biomechanical model of upper limb.

DESIGN METHODOLOGY

Most of the design of the arm-cranking apparatus has been performed using CADAM, a computer-aided design (CAD) software package developed by Cadam, Inc., and licensed by IBM. The physical CAD apparatus used consists of an IBM 5080 color graphics workstation, linked to an IBM 4341 computer. This system is a part of the CAD facility at Stanford University's Center for Design Research, a constituent part of the Stanford Institute for Manufacturing and Automation.

DESCRIPTION OF APPARATUS

Drive Systems

To date, two basic drive system designs have been implemented for initial experimentation: rotary crank (fig. 2); and oscillating lever (fig. 3). Linear and eccentric trajectory drive system designs are in progress, and will be used in subsequent experiments. In addition, the novel Kallander tapered belt drive will be combined with an oscillating lever or linear input, to achieve variable leverage throughout a power stroke without varying the nominal trajectory.

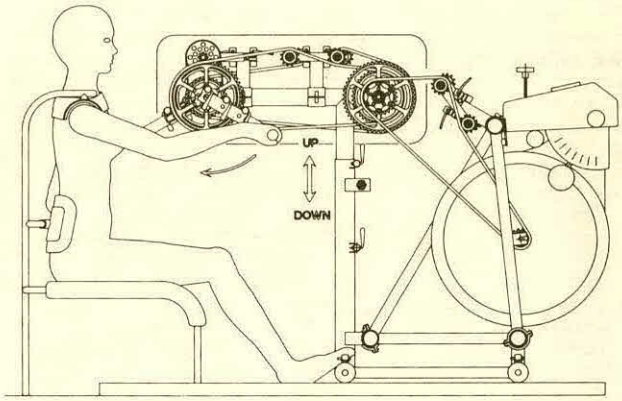

Figure 2. Rotary crank configuration, driving Monark 868.

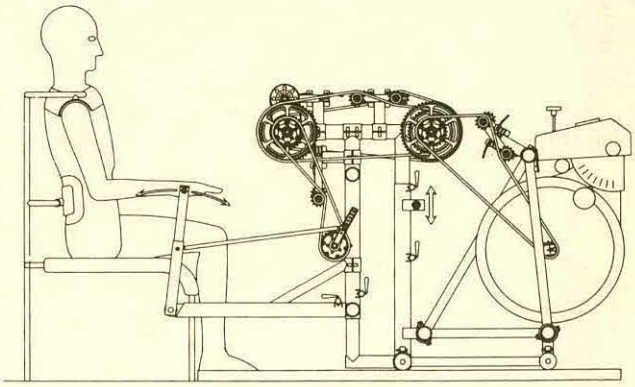

Figure 3. Oscillating lever configuration, driving Monark 868.

Load Devices

Power generated by the subject may be absorbed by three types of braking device: friction, viscous, and hysteresis (or eddy current). In the first case, the flywheel and controls from a pair of Monark 868 Ergometers have been adapted to a newly designed compact frame, which attaches to the main frame of the drive input aparatus. Frictional force generated by the belt on the flywheel rim is nearly independent of flywheel velocity; thus power output is directly proportional to the angular velocity of the cranks or flywheel.

For a viscous load, where resistance is a function of velocity squared, a Powercam Road Machine fan/flywheel ergometer will be adapted and made interchangeable with the Monark system on the apparatus.

A Dynavit Ergometronic 45 will provide the third type of load mechanism. The physics of this hysteresis brake yield a constant power output, independent of crank or flywheel velocity: as cranking speed increases, resistance is reduced, providing a smooth, even absorption of power.

Adjustable Parameters

Both the positioning of the subject relative to the drive input and the basic dimensions which define the trajectory of a particular drive mechanism may be varied to more closely fit the individual subject and to maximize the specified performance task. In addition, the response of the load resistance to power generation variations over time may have an effect on maximum power output, and will be ascertained by the comparative implementation of the three load systems. In general, the apparatus must incorporate the necessary flexibility in operator position, drive mechanism geometry and type of load system to fully challenge and develop the theoretical model.

Upper limb motion regimes may be established using one or more of several possible configurations. The gear ratio may be varied between left and right cranks, as may their relative phase angle. The split-spindle feature which allows the former, may also be employed to provide reverse cranking between the arms, and the provision of independent load devices for each arm allows the option of complete right-left mechanical decoupling. In addition, provision has been made for two different gripping positions. In the horizontal grip, the hand maintains a pronated position (fig.4), while in the angled grip, the hand is rotated seventy degrees away from pronation (fig. 5).

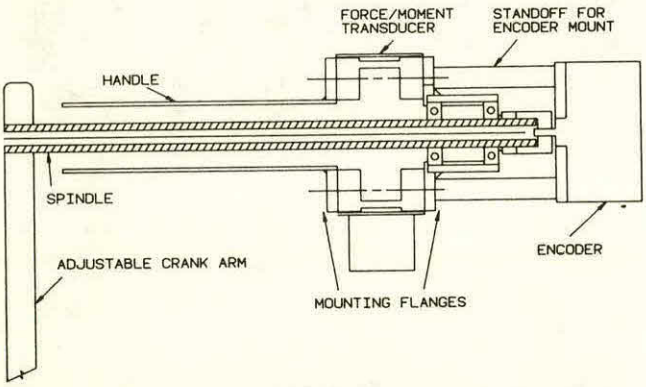

Figure 4. Section view of handle for pronated grip, with instrumentation.

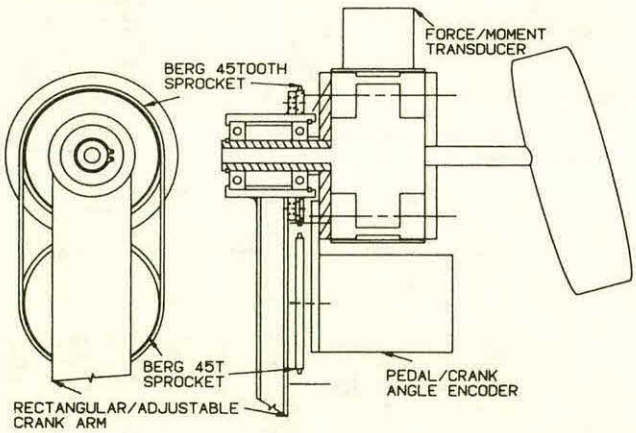

Figure 5. Section and end views of handle for rotated grip, with instrumentation.

Subject Positioning

Rather than have the subject posturally adapt to the apparatus, the subject is first seated in a comfortable position for the required task, and the apparatus is then brought to the desired operating position. Operator seating is thus fixed, while the ergometer is adjusted vertically in accordance with the subject's stature and then rolled on a track to within the subject's arm movement envelope. All adjustments, whether for positioning of the device or for adjusting a trajectory-parameter component, may be made with a minimum of simple hand tools, or in some cases with no tools at all. Once adjusted, settings are positively maintained with quick-release clamping devices.

Variable Drive System Geometry

The first experiments will examine the effects of crank length on maximal power generation. A variable crank arm has been designed, with an adjustable length of from two to twelve inches. It is important to adjust geometric parameters quickly between experimental runs: once again, no-tool quick-release fasteners have been employed.

As an alternative power input device, a modular lever drive system clamps onto the main arm-powered ergometer frame, linked by a chain run to the power take-up hub of the rotary crank mechanism. The lever-drive system permits variations in the length of the lever arm, and also allows changes in the relative lengths of two links in the four-bar linkage formed by the lever-to-crank mechanism, redefining the trajectory followed by the lever handle. In addition, the pivot point of the oscillating lever may be moved in the horizontal, vertical and lateral planes, to accomodate subject anthropometry and also to provide further variations in upper-limb motion regimes for additional experimental study.

Data Collection and Analysis

The arm-powered ergometer is equipped to simultaneously collect biomechanical and neuromuscular data. Forces and moments generated at crank or lever handles are registered by a six-degree-of-freedom force-torque transducer (JR3 Inc., Davis, CA), with angular correlation provided by Hewlett-Packard incremental optical shaft encoders on both the power input crank spindle and the handle itself (fig. 6). In addition, sixteen electromyograms will be recorded bilaterally from selected upper limb muscles, using surface electrodes. Both analog and digital data will be acquired by an IBM PC-AT computer, and information processed offline by a DEC Microvax II computer.

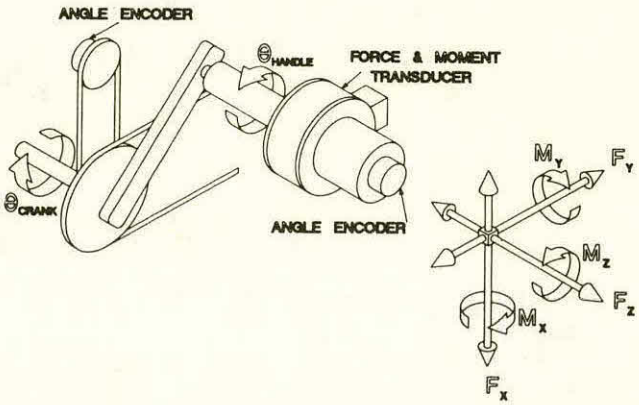

Figure 6. Schematic perspective of measured forces, moments and angles (pronated grip).

SUMMARY

Construction of the arm-powered ergometer is near completion, and first experiments using this modular multiple drive/load apparatus will soon begin.

ACKNOWLEDGEMENTS

Funding for this work was provided under a Veterans Administration project entitled "Optimal Biomechanical Design/Development of Arm-Powered Mobility Devices", M.R. Zomlefer, PhD., Principal Investigator. Construction and con-

sultation provided by Keith Bontrager, Santa Cruz, California. All design was performed at the Center for Design Research, Stanford University.

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P3.2

A MULTICHANNEL PORTABLE FUNCTIONAL ELECTRICAL STIMULATION SYSTEM

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ABSTRACT

A portable multichannel functional electrical stimulation (FES) system has been developed for real-time interactive control of upper and lower extremity motor function and rudimentary sensibility in spinal cord injured persons. The newly developed system utilizes two microprocessors and distributed processing to realize a versatile and flexible device. One microprocessor system functions as an input processor which transduces and processes various user generated commands. The second microprocessor is responsible for modulation of stimulus current amplitude, pulse width (PW) and interpulse interval (IPI) for multiple channels of stimulus output. Sixteen channels of constant current capacitively coupled biphasic stimulus output are provided for percutaneous stimulation. Alternatively, the FES system can control up to four eight channel implantable stimulators which have been developed by our

INTRODUCTION

Functional electrical stimulation (FES) has been used to provide control of hand function in quadriplegia [6,7] and control of locomotion in paraplegia [1,4,10]. Several research groups have designed and fabricated portable functional electrical stimulation systems to further investigate techniques developed in the laboratory. Early systems, including our own, were realized using discrete hardware solutions due primarily to power consumption constraints. These early systems were relatively simple and inflexible. With the advent of the low power, single chip microprocessor, more flexible portable FES systems have been developed.

This paper discusses the development of a portable microprocessor based multichannel FES system for real-time interactive control of movement and sensibility.

SYSTEM DESCRIPTION

A functional block diagram of the portable FES system is presented in Figure 1. The input(s) to the system are commands generated by the user from remaining voluntary function, such as shoulder position, which we use in our application to grade hand movement in upper extremity systems. The input commands are transduced by various transducer systems. They can be proportional and/or discrete in nature. For instance, shoulder position is measured relative to the sternum using a small two orthogonal axis proportional position transducer [2]. Discrete inputs such as switch closures may also be employed.

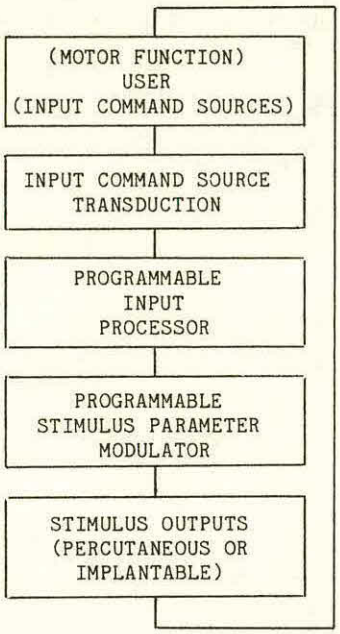

Figure 1. Functional block diagram of the portable FES system.

The programmable input processor is responsible for the processing of the transduced input commands. Signal conditioning, ranging from simple gain and filtering to realization of nonlinear transfer functions, is first carried out. Signal conditioning hardware consists of two channels of analog input with processor programmable gain and offset (used in conjunction with the shoulder position transducer). A 16 channel, eight bit analog to digital converter (ADC) is used to convert the transduced and partially processed user input commands, including discrete inputs such as switches.The conditioned input commands are then used as inputs to a control algorithm which provides the user with a simple means for controlling the FES system. The control algorithm in turn produces the control signal(s) necessary to modulate various parameters of the stimulus waveform. Audio cueing provides the user with feedback to the system operating state. We are currently investigating the use of sensibility for system state feedback.

The programmable stimulus parameter modulator is responsible for modulation of the cathodic phase pulse width (PW), current amplitude and the interpulse interval (IPI) timing of multiple stimulator channels as a function of the control signal(s). The input/output relationship of the

stimulus parameter modulator is accomplished through a technique of indexed lookup tables. The control signal(s) is used as an index or pointer into the various parameter lookup tables. The parameter lookup tables are determined on the computer based laboratory stimulation system [11]. These lookup tables are then transferred into the portable system. Cathodic pulse width timing can be varied from 0 to 32 msec with 0.5 usec resolution for each channel independently. It is possible to operate each of the output channels as an independent stimulator with its own interpulse interval timing. Most coordinated function however is accomplished as a result of several channels working together. The modulator allows the flexible formation of "groups" [11], i.e. one or more stimulus channels that operate at the same IPI. The IPI timing can be as short as 2 msec and has a 1 msec resolution. The following information is supplied to the modulator; (a) the number of groups being used, (b) the stimulus channels belonging exclusively to each stimulus group, (c) the IPI for each group, and (d) the stimulus PW for each individual stimulus channel, (e) the virtual to actual output channel mapping, (f) the current amplitude. In the case of the implant stimulator, the system is also used to format the serial data string needed to control channel addressing and stimulus timing.

An eight bit parallel interprocessor communications bus for communication between the input section and the modulator section is externally available for communication with other processor systems such as our laboratory system. The communications bus also allows for future expansion capabilities such as math coprocessors or other digital based subsystems.

The stimulus outputs can be either external, for stimulation through chronically indwelling percutaneous electrodes, or generated by a totally implantable stimulator system(s) which we have developed [9]. The stimulus output stage for both the percutaneous stimulator and the implantable stimulator generate sequential, constant current, capacitively coupled, biphasic outputs. The waveform consists of a rectangular cathodic current pulse followed by a current limited (typically 0.5 mA) capacitive discharged anodic pulse.The cathodic current pulse is controlled by the modulator with a typical range of 0 to 25 mA. Due to the topology of the actual circuitry, the pulse width for each channel is limited to a maximum of about 300 usec. A closed loop DC to DC converter provides 40 volts of stimulator output voltage for the percutaneous stimulator. For a constant current output of 20 mA, the output stage can support electrode impedances as high as 2 Kohms. When the portable FES system is used in conjunction with the implant stimulator, a circuit board with four transmitters is substituted for the percutaneous stimulator circuit board.

As can be seen from the block diagram in Figure 1, the user provides the only feedback regulation for the system. The FES system itself operates as an open loop controller at this time. Studies with closed loop regulation of position and force are presently being investigated in our laboratory [5].

The system electronics are housed in a 7 x 14 x 19 cm plastic enclosure with a weight of 1000 grams. A photograph of the system is presented in Figure 2. Power is supplied from up to six separate sets of four AA Nicad batteries connected in series. One set of four batteries is user accessible. An external input is also available for powering the system and recharging the batteries. The input processor handles all the power management.

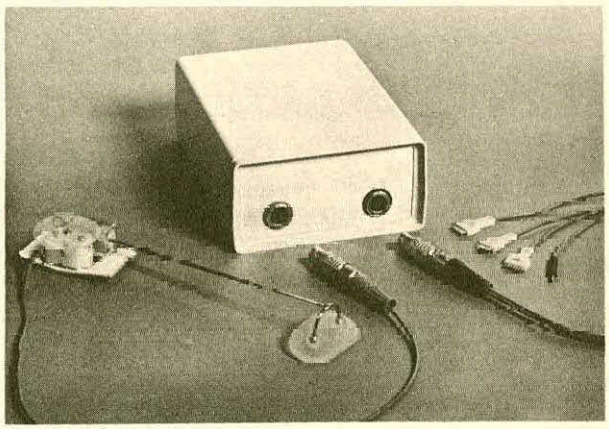

Figure 2. Portable FES system.

SYSTEM OPERATION

The primary application of the portable FES system by our center is for the control of hand function in persons with C5 or C6 spinal cord injury. The following example is for a percutaneous hand neuroprothesis system. The portable FES hand orthosis system consists of three basic components. They are the electronics package, input transducers and cable (usually a two axis shoulder position transducer and chest mounted switch) and electrode cable consisting of up to three percutaneous electrode site connectors and an anode.

The microprocessor based system serves as both an exercise system as well as a functional system. Electrically induced exercise enables the muscle strength and fatigue resistance to be increased [8]. When the stimulator cable is plugged in, without the transducer cable connected, the device operates in an exercise mode. If the transducer cable is connected before the electrode cable is connected, the system will operate in a functional interactive mode as described below. It is the cables alone that select the operating mode as well as turn the system on. No other external switches need to be set or other adjustments need to be made by the attendant. The exercise algorithm allows the grasp to be ramped open, closed and held at particular values. It can also cycle between two or more grasping patterns and turn on and off in preset time cycles. A typical exercise regime which is applied throughout the night while the user is sleeping provides a 50 minute period of alternating grasping modes (10 grasp cycles per mode) and a 10 minute resting period.

Typical functional interactive system operation might be as follows; first the electrode cable is connected to the electrode sites and the anode is applied to the upper arm. The transducer is attached to the chest and shoulder using double sided tape. These cables and transducers are worn beneath the user's clothing. The small electronics package is generally clasped to the underside of the individual's wheelchair seat. The transducer cable is connected to the stimulator unit first since it determines whether the unit is in a functional user interactive mode or it is in the exercise mode. The electrode cable is then connected, which powers the system up in an idle mode. The idle mode is a non stimulating low power consumption state. To functionally operate the system, the user depresses switch mounted adjacent to the shoulder position transducer. The system goes into a grasp mode selection scan in which audio cues of one and then two short tones indicate which grasp mode will be controlled. Releasing the chest switch will stop the scanning in the desired grasp mode. A short delay (approximately 3-5 seconds) allows the user to position his shoulder at the set point he desires. After the delay, the system will turn on in a functional mode with the defined set point representing a zero level of command. Shoulder movements from the set point will, in turn, proportionally control the selected grasp. Rapid movement along the axis orthogonal to the proportional axis (exceeding a velocity for a given period of time) will initiate a hold mode, which will maintain a constant stimulus output independent of shoulder position. To exit the hold mode, another rapid movement will allow the user to regain proportional control after realigning his shoulder to the position it was in along the proportional control axis when the hold was initiated. This provides a smooth transition from the hold mode to the proportional control mode without disrupting the grasp. Audio cues indicate the state of operation to the user. To place the system into the idle mode, the user needs only to depress and release the chest mounted switch. The system will remain powered until the electrode cable is disconnected.

This portable functional stimulation system has recently been introduced into clinical usage. An earlier single microprocessor based FES system has been utilized clinically for four years [3].

ACKNOWLEDGEMENTS

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QUANTIFICATION OF MOTOR CONTROL WITH TRACKING ACCURACY

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ABSTRACT

This paper describes a tracking system with which to quantify the accuracy of human motor performance. An Apple IIe computer was programmed to display a variety of patterns which serve as targets for both a handgrip tracking test and a joint-position tracking test. Accuracy of performance is determined by calculating the root-mean-square error between the target and the response; in addition, an accuracy index is derived from each response. These tests should be useful clinically in the early identification of motor control impairment in certain patients, as well as in the evaluation of the effectiveness of many clinical procedures designed to improve motor function.

INTRODUCTION

The scientific literature devoted to physical rehabilitation is replete with studies investigating muscle strength, endurance and joint mobility. Each of these elements is fundamental to motor performance and, therefore, deserves intensive investigation. However, a distinctly separate ingredient that is equally important to skilled movement, but remains alarmingly less prominent, is motor control, or attention to the accuracy of motor function.

Objective measurement of exacting motor performance can be obtained from pursuit tracking tests. In these experiments, individuals attempt to follow a displayed target with an object that is controlled by delicate adjustment of muscular activity. Since tracking skill is an inherent component of many functional activities, where there is an ongoing demand for precise modulation of the intensity and timing of muscle activity, tracking experiments should serve as a valid tool with which to quantify motor function in both normal and disabled individuals. Indeed, a limited number of tracking studies have been performed to objectively assess motor performance in individuals with multiple sclerosis (1,2), Parkinson's disease (3,4), cerebellar ataxia (5,6), and brain injuries (7,8). There still exists, however, a great need to validate the efficacy of various therapeutic procedures purported to improve motor function in individuals. The purpose of this paper is to describe a tracking system designed to provide this much-needed documentation.

MATERIALS

We have developed a system which involves two forms of pursuit tracking; force tracking and joint-position tracking. With both forms, an Apple IIe computer is programmed to display a series of previewed tracking patterns on the monitor. These patterns then serve as targets along which the individual attempts to superimpose the cursor by fine adjustment of either handgrip force or finger joint position.

For the handgrip tracking task the amplitude of each target pattern is based on the subject's predetermined maximum handgrip force. The force signal, representing the applied force on a handgrip dynamometer, is directed from an Interface load cell, through a Gould voltage amplifier and a 12-bit Interactive Structures A/D converter, to the computer. The computer is programmed to display the cursor at a calculated height on the monitor corresponding with the applied force. The subject is instructed to squeeze the handpiece with just the right intensity to adjust the vertical position of the cursor so as to superimpose it on the target as the cursor moves automatically from left to right across the screen with a sweep speed of 10 seconds.

For the joint-position tracking test an electrogoniometer is centered along the lateral side of the metacarpophalangeal joint of the index finger. The amplitude of the joint-position tracking pattern is based on the predetermined maximum range of flexion-extension at that joint. Similar to the former tracking test, the subject is instructed to trace the cursor as accurately as possible along the displayed target pattern by delicate adjustment of the finger joint position.

MEASUREMENT OF TRACKING PERFORMANCE

The computer is further programmed to quantify the accuracy of performance by calculating the root-mean-square (RMS) error between the target pattern and the actual response. In order to allow for meaningful comparisons of performance between individuals, a standardized score is also computed. This accuracy index (AI) is derived from the subject's RMS error, as well as the value of the subject's individualized pattern.

RESULTS

Illustration of a typical response of a normal subject to one of the handgrip tracking patterns is seen in Figure 1. Figure 2 shows the typical response of a normal subject to one of the joint-position patterns.

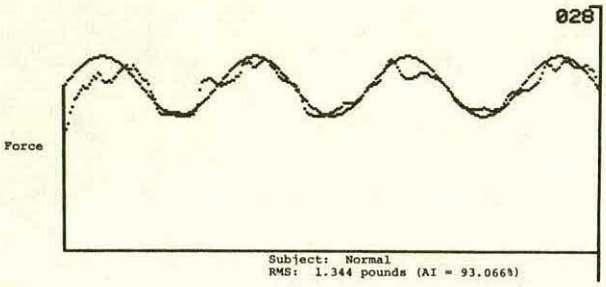

Fig. 1. Response of a normal subject to one of the handgrip tracking patterns. The number in the upper right corner indicates the 50% value in pounds of the subject's maximum handgrip force.

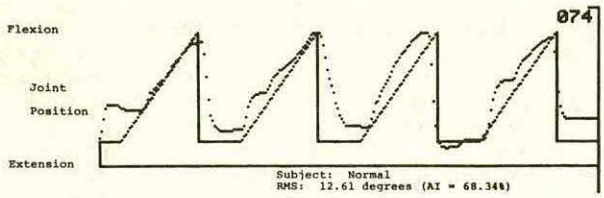

Fig. 2. Response of a normal subject to one of the joint-position tracking patterns. The number in the upper right corner indicates the full range of flexion-extension in degrees at the subject's metacarpophalangeal joint of the index finger.

Figure 3A shows the same joint-position pattern and the markedly reduced tracking accuracy, particularly in the extension range, of an individual with flexor spasticity. Furthermore, Figure 3B demonstrates the improved tracking performance of the same patient following five minutes of manual stretch to the subject's finger flexors and wrist flexors.

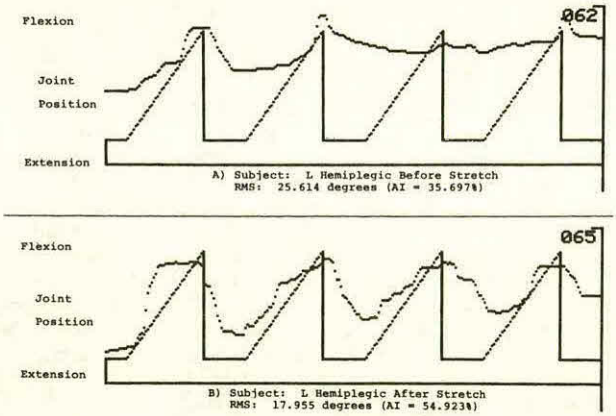

Fig. 3A-B. Response of a hemiplegic individual to one of the joint-position tracking patterns both before (A) and after (B) five minutes of manual stretching to the spastic finger and wrist flexors.

DISCUSSION

We believe the tracking system described above is a reliable and valid indicator of motor function in individuals. This relatively simplistic yet extremely informative apparatus should provide considerable value clinically in 1) early identification and continued monitoring of impaired motor control in certain neuromuscular disorders, and 2) evaluation of the efficacy of various surgeries, medications, exercises, and other modalities employed to improve motor control.

ACKNOWLEDGEMENTS

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JOINT ANGLE SENSORS FOR CLOSED-LOOP CONTROL

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ABSTRACT

In an automatic control system (closed-loop control system) the output of a sensing device is used to effect a control function without the aid of a human operator. The usual way to measure joint angles is with goniometers. However, these tools lack accuracy and reliability, are difficult to mount and they are uncosmetic. This paper presents a joint angle sensor design for use in closed-loop control of lower extremity movements in paraplegics. Current sensor evaluation shows good static performance and a frequency response from 0 Hz up to 10 Hz. Other advantages of this device include light weight, small size and low power consumption.

INTRODUCTION

Research in functional electrical stimulation has been undertaken to develop a closed-loop control system to help paraplegic people to walk normally as well as safely. Essential elements of this control system include sensors which measure parameters in order to control and analyze the movements of a walking subject. The sensor outputs, and analyzed results will be used to activate control functions semiautomatically or automatically without the conscious aid of a human operator. The present way to accomplish this is by using a mechanical brace with a built-in goniometer on the desired joints. The accuracy and reliability of current goniometers is not satisfactory. Problems include slippage during joint movement, short lifetime, electrical noise and a poor physical appearance. Goniometers which measure joint angles can only operate while attached to a brace. Therefore, it is desirable to develop an improved sensor for joint angle measurement.

METHOD

Figure 1 illustrates the conceptual design of the sensor in which the change in capacitance is proportional to the displacement. A spring-loaded device is used to sense the movement of the joint angle. The sensing element is encapsulated in medical grade silicone rubber tubing with an outer diameter of 1.125 mm. The device incoporates CMOS hybrid circuitry packaged in a 1" X 0.7" X 0.2" metal flatpack (1). This circuit converts the capacitance changes of the sensing element into a DC output voltage. The circuit draws less than 1 mA using a 8-9 volt battery supply.

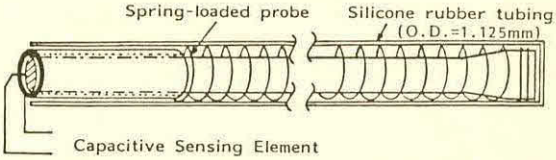

Figure 1. The Conceptual Design of the Joint Angle Sensor.

RESULTS

The measured static characteristics of the sensor are listed in Table 1. The sensitivity of the sensor is 0.95 mV/pf; the stability over 72 hours is 10mV drift at 5 volt full swing. Its temperature coefficient is -30 mV/°C. It has been tested throughout a temperature range of 4.5° C to 48.9° C (40° F to 120° F). Dynamic testing was performed by comparing this sensor, mounted near the knee, to a goniometer built in a brace worn by the same person. The frequency response from 0 Hz up to more than 10 Hz is flat within 0.7 dB. Comparison of the goniometer to the joint angle sensor and the frequency response data are illustrated in the Figures 2a and 2b.

Table 1. Specification of Joint Angle Sensor

Total Range	+0 to +180 degree
Linear Range	+0 to +160 degree
Accuracy	+22mV/degree
Threshold and Resolution	± 0.5 degree
Linearity	± 1 percent angle
Repeatability	± 0.5 degree
Electrical	
Voltage Supply	8 volt
Voltage Supply Range	6 to 15 volt
Current Draw	< 1mA
Voltage Output	0.95V/pf
Output Impedence	100 Kohm
Environment	
Temperature Range	+40° F to +120° F
Temperature Shift	+1.4° /degree

DISCUSSION

If desired, the small temperature shift of the sensor can be compensated by electronic circuitry. The sensor appears to have good performance characteristics for both surface mounting and implant applications. Nevertheless, for widespread applications, further studies to eliminate calibration and mounting errors, as well as to design a biocompatible implant package are necessary.

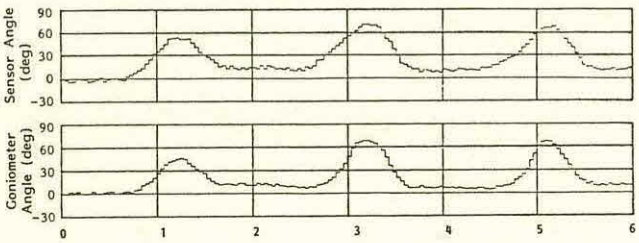

Figure 2.a Comparision of the Output of the Sensor to the Output of the Goniometer. (Sampling Frequency = 25 Hz)
* The steps in these curves are caused by the finite sampling rate and resolution of the measurement system.

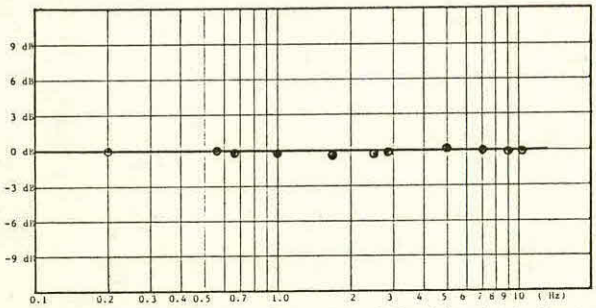

Figure 2.b Frequency Response of the Joint Angle Sensor.

ACKNOWLEDGMENTS

This research has been supported by the Rehabiliation Research and Development Service of the Veterans Administration.

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TOWARDS IMPROVED ACCESSIBILITY OF THE ICON EDUCATIONAL MICROCOMPUTER

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ABSTRACT

This paper and its accompanying poster describe the Icon educational microcomputer and recommendations for making this computer accessible to students with a wide range of physical disabilities. An Interface Unit (IU) has been developed that will allow a large number of alternate input devices to be used with the Icon. Devices that will be evaluated with the IU in the next phase of this project are briefly discussed as is the on-line help facility that is under development.

INTRODUCTION

In 1983 the Ontario (Canada) Ministry of Education issued a set of detailed functional specifications for an Educational Microcomputer. The goals of these specifications were not only to stimulate the development of a family of compatible computers explicitly designed to suit the educational environment but also to stimulate Ontario and Canadian hardware and software industries.

The first microcomputer to meet the Ministry's specifications and thus acquire OAEM (Ontario approved Educational Microcmoputer) status was the Icon computer. The Icon is a network-based microcomputer system consisting of a fileserver (the Lexicon) and from one to 32 Icon workstations. Both the Lexicon and the Icon are based on 80186 microprocessors. The Lexicon has 384K or 512K of memory and the Icon 384K which can be upgraded to 512K.

In parallel with the hardware development efforts, the Ontario Ministry of Education has embarked on a program to encourage the development of high-quality educational software by awarding software development contracts to Canadian companies and institutions. Currently, approximately 39 contracts have been completed and another 100 are under development. One of these contracts was awarded to the Hugh MacMillan Medical Centre and the Hugh MacMillan Centre School Entitled: "Towards Universality of Access to Information", its purpose is to provide access to the Icon computers by a wide range of physically disabled students.

OBJECTIVES

General

To create and evaluate an Interface Unit (IU) that will enable physically disabled students who cannot use the Icon keyboard and/or trackball to access the Icon via alternative keyboards, switches and alternative pointing devices.

Specific

(1) To evaluate the use of the Icon by physically disabled pupils; (2) to design and build a prototype interface that will allow for transparent connection of alternative keyboards; MOD- and definable-keyboards; single and multiple switches and alternative pointing devices; and (3) to evaluate the interface and the operation of the adaptive devices.

The project is divided into four stages:

1a) Preliminary design

1b) Design

2) Implementation and development
During this stage the second prototype interface unit will be developed and the first prototype will be tested in the classrooms of the Hugh MacMillan Centre School (HMCS).

3) Formative field evaluation
In this stage the second prototype will be tested both at HMCS and at other schools in Ontario.

EVALUATION

Stages 1a (Preliminary design) and 1b (Design) have been successfully completed. A total of 60 pupils using the Icon without any adaptive devices were evaluated by therapists, teachers and by an Education Resource Technician. Questionnaires focussing on educational and therapeutical aspects were administered in order to assess the type of problems encountered by physically disabled pupils.

Problems identified in the evaluation range from the weight, bulk and nonmodularity of the Icon computer to the size and colour of the cursor. The main recommendations stemming from the stage 1a and 1b evaluations included the provision of remote, detachable and redefinable keyboards, and keyboard systems. These changes came from observations of children who could not appropriately and comfortably reach all of the Icon's keys. The redefinable keyboards (see below) and the NRC MOD system (Nelson, Park and Korba, 1984) would expand the accessibility of the Icon.

Two other recommendations were the provision of a detachable/external trackball to allow the trackball to be relocated in a position easier for the student to reach. A keyguard was also a strong recommendation for students with targetting problems.

INTERFACE UNIT (IU)

The first prototype Interface Unit (IU) has been constructed and multiple copies of the IU board were produced for classroom evaluation. Development of an infra-red remotely linked version of the IU have commenced (D'Alessandro et al (these proceedings)). The Interface Unit is a 30 x 20 x 5 cm (12 x 8 x 2") box which is connected to the Icon. Alternative input devices can be plugged into the IU which decodes the input signals into Icon-keyboard or Icon-trackball compatible inputs. Thus the IU provides transparent input to the Icon. Details of the IU and its remote-linked version can be found in D'Allesandro et al.

IU PERIPHERALS

In stages 2 and 3 the Interface Unit will be evaluated and field tested with a number of alternate input devices and Icon adaptations. These devices, most of which are commercially available, are listed below. Instances of target users are provided, however these should be regarded only as potential applications of the devices.

1. Extended Keyboards

The IU will allow the use of parallel or serial keyboards. Some examples are: RCA, Apple, and IBM keyboards. These keyboards could be used for a child who is very small and cannot reach the Icon keyboard and see the monitor simultaneously.

2. Extended Trackball

The extended trackball is attached to the trackball port by a 6 foot cord. This extended trackball can be used by children who cannot be positioned close to the Icon because of seating restrictions. Other potential users are children who cannot use the action (selection) key and the trackball simultaneously because they have use of only one hand.

3. Mini Keyboard

The mini keyboard is a small keyboard with closely spaced keys. The keys can be activated with a conductive-tip pen. The conductive tip can also be attached to a head pointer, mouthstick, or thimble. The mini keyboard can be used with a child who has a very limited range of finger movement (e.g. children with Muscular Dystrophy, Juvenile Arthritis).

4. Unicorn Expanded Keyboard

The Unicorn Keyboard has 128 pressure-sensitive key areas. Individual keys can be grouped together to function as a single larger key. Custom layouts can be programmed and changed with the use of overlays. This can be used with children who have limited control of arm movement because of athetosis or spasticity (e.g. Cerebral Palsy).

5. King Keyboard

The King Keyboard has coloured enlarged (1-1/4 inch diameter) slightly recessed keys. This keyboard can be used with children who cannot isolate their index finger and control arm movement sufficiently for regular keyboard targetting.

6. Keyguard

The custom-made Icon keyguard allows hand stabilizing while targetting on keys. The keyguard also prevents activation of more than one key at a time. This device helps the child who has limited control of movement or tremors that cause more than one key to be hit accidentally.

7. NRC MOD Keyboard

The NRC MOD keyboard system (Nelson, Park, and Korba, 1984) allows severely disabled children to choose and enter commands or messages using single or multiple switches (see below). In the MOD system a second computer provides a scanning and selection display from which the child can make command selections.

8. Single and Multiple Switches

A wide variety of pressure, airflow, and touch activated switches is available. These switches can be used to control cursor movement or to make selections of commands or messages in the software. Switches can be used in conjunction with the NRC MOD keyboard or for Icon software that requires switch input.

9. Alternate Pointing Devices

A number of alternatives to the Icon's standard pointing device, the trackball, are being considered. These include a gated joystick, PC mouse and Mac-mouse. Target users of these devices would be students who cannot use the trackball or extended trackball.

TEACHER UTILITY

A software utility program is being prepared which will assist teachers and therapists in the use of the Interface Unit. In this utility program the teacher is led through a structured network of menu options which assists in the determination of the functional abilities and problems of prospective users. Suggestions for access adaptations and input devices are made. It is anticipated that this Teacher Utility will become a fully on-line help facility for both the technical aspects of the IU and to a limited extent for the more clinically oriented set-up aspects.

A large portion of the upcoming evaluation will be directed at establishing the boundaries and safe limits of clinically oriented help directed at teachers and therapists in school environments. These limits will depend on the complexity of the accessing problems presented by the student and on the expertise available on-site.

ACKNOWLEDGEMENTS

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P5.1

MUSCULOTENDON ACTUATOR MODELS FOR USE IN COMPUTER STUDIES AND DESIGN OF NEUROMUSCULAR STIMULATION SYSTEMS

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ABSTRACT

A computer model is needed to design functional neuromuscular stimulation (FNS) systems that will have the ability to activate many muscles to effect functional control of multi-articulated paralyzed limbs. Critical to this task is a model of each muscle and tendon (the musculotendon actuator) that will be stimulated. We have developed a dimensionless, first-order model of musculotendon contraction dynamics that is based on fundamental architecture and properties of muscle and tendon. Only three actuator-specific parameters (muscle strength, optimal muscle fiber length and tendon slack length) are needed to scale the generic representation to a specific actuator. We show that tendon can be a dominant factor in musculotendon mechanics. This model for contraction dynamics, when coupled to our first-order model for electrically-stimulated activation dynamics, will be used to study the trade-offs among alternative FNS systems.

INTRODUCTION

One of the obstacles preventing clinical use of NMS systems for restoration of upper or lower limb function is the lack of knowledge of how muscles in the able-body are normally coordinated and how a smaller subset of these muscles in the disabled ought to be controlled to restore function to paralyzed or paretic limbs. As multichannel implantable stimulators become available, the rehabilitation team must not only decide on the subset of muscles to be stimulated but, together with the potential user, the team must also be cognizant of the performance to be expected from the (presumably) best choice. It is necessary, therefore, to develop a computer model of the neuromusculoskeletal system to assist in the development of multi-muscle stimulation systems for functional control of multi-articulated paralyzed limbs.

Important to the development of a computer model of the neuromusculoskeletal system is a model of each muscle that is, or might be, a candidate for stimulation. Though it is true that muscle has been well-studied, it is also true that there exists a large repertoire of muscle models (i.e., many mathematical formalisms to represent muscle's properties). These models can be classified, however, into two basic groups: those used to formalize the mechanical properties of muscle at the holistic level, and those used to formalize muscle's sub-cellular properties. Regardless of the approach, though, it might seem that one still needs many parameters to model a given muscle. Tendon is also an element of the "musculoskeletal system," and perhaps an important one, albeit its effect on musculotendon properties is not well documented.

Our approach to modeling muscle and tendon, i.e., the musculotendon actuator, is:

(1) to base the model on musculotendon architecture and fundamental mechanical properties of muscle fibers and tendon;

(2) to structure the model so that its dynamics are low-order;

(3) to find a dimensionless representation of musculotendon dynamics.

This approach is particularly suited to the understanding of the interrelationship of fundamental musculotendon features on musculotendon mechanics and to the modeling of many musculotendon actuators. Our work to date suggests that only a few parameters, used to scale a nominal representation, are needed to model a specific human musculotendon actuator and that tendon compliance of some actuators can dramatically affect musculotendon mechanics.

MUSCULOTENDON MODEL

For the lumped-parameter mechanical model for tendon and muscle (Fig. 1), we assume that tendon is a conservative spring with variable stiffness k^{T}, that passive muscle PE develops force F^{PE} as would a spring with variable stiffness k^{PE}, and that muscle, once activated, develops active force F^{CE} from its contractility properties. The relationship between electrical pulse stimulation and activation $a(t)$ is discussed elsewhere in this volume. The purpose here is to formally present our model for musculotendon contraction dynamics.

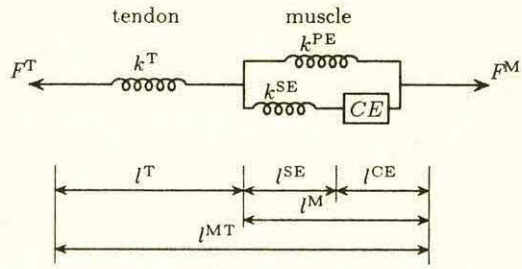

Figure 1.

We further assume that the properties of active muscle are given by 2 elements: a contractile element CE described by a force-velocity (f-v) relationship dependent on activation $a(t)$ and muscle length l^{M} (note that muscle fiber lengths are assumed equal), and a series elastic element SE to account for muscle's short-range stiffness k^{SE} (e.g., stiffness of cross-bridges among myofilaments). The contractile mechanism of muscle is assumed, in steady-state, to develop isometric force $\tilde{F}^{\mathrm{CE}}_{iso}$ that depends multiplicatively on its length l^{M} and its activation $a(t)$. Actually, it is through isometric force $\tilde{F}^{\mathrm{CE}}_{iso}$ that the f-v curve is assumed to depend on $a(t)$ and l^{M}.

Dimensional Analysis and Definitions

The dimensional quantities involved are force, length and time. We choose for the normalization parameters

F^{CE}_o = maximum isometric active muscle force

l^{M}_o = length of muscle at which F^{CE}_o is developed

$\tilde{v}^{\mathrm{CE}}_m$ = normalized maximum shortening velocity of the contractile element $(10 \cdot \mathrm{s}^{-1})$

Some definitions needed are noted below; others are introduced later.

F^i = force in ith element

$\tilde{F}^i = F^i/F^{\mathrm{CE}}_o$ = normalized force in ith element

l^i = length of ith element

$\tilde{l}^i = l^i/l^{\mathrm{M}}_o$ = normalized ith element length

$v^i = \dfrac{dl^i}{dt}$ = velocity of ith element

$\tilde{v}^i = \dfrac{d\tilde{l}^i}{dt}$ = normalized ith element velocity (1)

$\tau = \tilde{v}_m^{CE} \cdot t$ = dimensionless time

$\tilde{v}_\tau^i = \dfrac{d\tilde{l}^i}{dt}$ = dimensionless ith element velocity

$k^i = \dfrac{dF^i}{dl^i}$ = stiffness of ith element (i = T, PE, SE)

$\tilde{k}^i = \dfrac{d\tilde{F}^i}{d\tilde{l}^i}$ = normalized ith element stiffness (i = T, PE, SE)

where i = T, M, PE, SE, or CE, except as noted.

Tendon

The model's tendon element represents the physiological tendon both external and internal to the muscle. We assume tendon strain ϵ^T is the same everywhere and that the normalized tendon force-strain curve is given in Fig. 2 [1,2]:

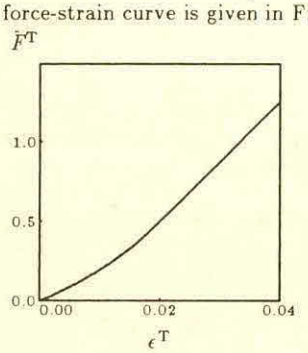

Figure 2.

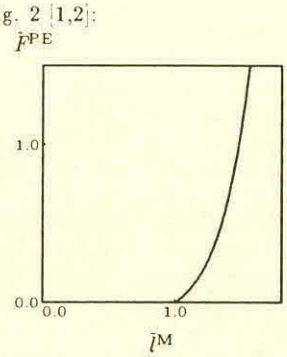

Figure 3.

Some definitions related to tendon are:

$$\begin{aligned} &l_r^T = \text{tendon slack length;} \\ &\tilde{l}_r^T = l_r^T / l_o^M = \text{normalized tendon slack length;} \\ &\epsilon^T = (l^T - l_r^T)/l_r^T = (\tilde{l}^T - \tilde{l}_r^T)/\tilde{l}_r^T = \text{tendon strain.} \end{aligned} \tag{2}$$

We assume that the $\tilde{F}^T$-ϵ^T curve (Fig. 2) is the same among musculotendon actuators but that l_r^T and $\tilde{l}_r^T$ are not. It is also assumed that tendon stress when $\tilde{F}^T = 1$ is 32 MPa, or

$$\sigma_o^T \equiv \sigma^T_{\tilde{F}^T=1} = 32 \text{ MPa}. \tag{3}$$

Passive Muscle

Passive muscle held at lengths $l^M > l_o^M$ develops force. We assume that this (normalized) force arising from either interfiber connections or elastic structures internal to the muscle fiber depends on (normalized) muscle length and that this $\tilde{F}^{PE}(\tilde{l}^M)$ relationship (Fig. 3) is the same among actuators [3].

Active Muscle

The isometric *force-length curve* of fully-activated muscle ($\tilde{F}_o^{CE}(\tilde{l}^M)$; Fig. 4; $a(t) = 1$) used in our human muscle model is based on the amount of overlap of myofilaments expected from human myofilament lengths [4,5] under the assumption of sarcomere and muscle fiber homogeneity.

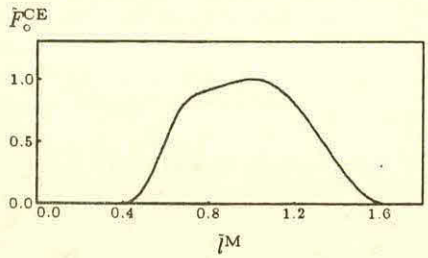

Figure 4.

The $\tilde{F}_o^{CE}(\tilde{l}^M)$ curve is assumed to be the same among actuators. Less than fully-activated muscle (i.e., $a(t) < 1$) is assumed to have an isometric force-length relationship that is identical in shape but reduced in amplitude; i.e.,

$$\tilde{F}_{iso}^{CE}(\tilde{l}^M, a(t)) = F_o^{CE}(\tilde{l}^M) \cdot a(t). \tag{4}$$

where $\tilde{F}_{iso}^{CE}$ = normalized isometric force of active-muscle.

The *f-v characteristic* of active-muscle is accounted for by the CE. Actually, what is needed in our formulation is a v-f relationship and again this relationship (Fig. 5), when made dimensionless, is assumed to be the same among specific actuators.

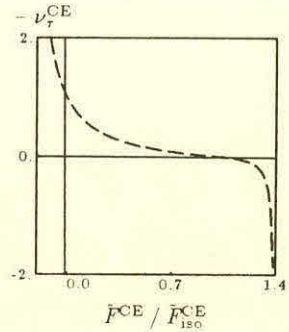

Figure 5.

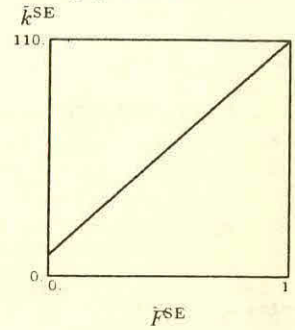

Figure 6.

Note that $\tilde{v}_\tau^{CE}$ = dimensionless isotonic velocity of active-muscle.

We use a modified version of Hill's equation [6] for this relationship to account for either the shortening or lengthening of activated muscle. Notice that the isotonic velocity of active-muscle $\tilde{v}_\tau^{CE}$ depends on length $\tilde{l}^M$ and activation $a(t)$ through $\tilde{F}_{iso}^{CE}$. To scale the dimensionless v-f relationship to obtain the absolute v-f relationship, one needs to multiply $\tilde{v}_\tau^{CE}$ by $[\tilde{v}_o^{CE} \cdot l_o^M]$ and $\tilde{F}^{CE}/\tilde{F}_{iso}^{CE}$ by $[F_o^{CE} \cdot \tilde{F}_{iso}^{CE}(\tilde{l}^M, a(t))]$. Notice also that with this formulation maximum shortening velocity of isotonic muscle is independent of $\tilde{l}^M$ and $a(t)$, and equal to $\tilde{v}_m^{CE} \cdot l_o^M = 10 \cdot l_o^M \text{m} \cdot \text{s}^{-1}$.

Stiffness of active-muscle [7] is accounted for by the SE and, when put in dimensionless form, assumed to be the same among actuators (Fig. 6).

MUSCULOTENDON ACTUATOR DYNAMICS

To find the dynamics of the musculotendon actuator in dimensionless form we find first the dimensionless dynamics of muscle alone and then the dimensionless dynamics of the musculotendon system.

Muscle Dynamics

From Fig. 1, it is seen that

$$\begin{aligned} &\tilde{F}^M = \tilde{F}^{PE} + \tilde{F}^{SE}; \qquad && \tilde{l}^{PE} = \tilde{l}^M; \\ &\tilde{F}^{SE} = \tilde{F}^{CE}; && \tilde{v}_\tau^M = \tilde{v}_\tau^{SE} + \tilde{v}_\tau^{CE}. \end{aligned} \tag{5}$$

Since

$$\frac{d\tilde{F}^M}{d\tau} = \frac{d\tilde{F}^M}{d\tilde{l}^M} \cdot \frac{d\tilde{l}^M}{d\tau}. \tag{6}$$

one can derive from eqs. 1, 5 and 6 the dimensionless muscle dynamics

$$\frac{d\tilde{F}^M}{d\tau} = \left(\tilde{k}^{SE} + \tilde{k}^{PE}\right) \cdot \left[\tilde{v}_\tau^M - \left(\frac{\tilde{k}^{SE}}{\tilde{k}^{PE} + \tilde{k}^{SE}}\right)\tilde{v}_\tau^{CE}\right]. \tag{7}$$

where $\tilde{k}^{M} \equiv \tilde{k}^{PE} - \tilde{k}^{SE}$ = muscle stiffness and $\tilde{k}^{PE}$ is given, in effect, by Fig. 3 and depends on $\tilde{l}^{M}$; $\tilde{k}^{SE}$ is given by Fig. 6 and depends on $\tilde{F}^{SE}$ or, equivalently, on $\tilde{F}^{M}$ and $\tilde{l}^{M}$; and $\tilde{v}_{\tau}^{CE}$ is given by Fig. 5 and depends on $\tilde{l}^{M}$, $a(t)$ and $\tilde{F}^{CE}$ (or, equivalently, on l^{M},$a(t)$ and $\tilde{F}^{M}$ since $\tilde{F}^{CE}$ is a function of $\tilde{F}^{M}$ and $\tilde{l}^{M}$). Thus, eq. 7 can be written

$$\frac{d\tilde{F}^{M}}{d\tau} = f(\tilde{F}^{M}, \tilde{l}^{M}, \tilde{v}_{\tau}^{M}, a(t)). \tag{8}$$

to show clearly that the rate of change of force developed by muscle depends on its force, length, velocity and activation.

Musculotendon Dynamics

In a similar way and recognizing that

$$\begin{aligned} \frac{d\tilde{F}^{T}}{d\tau} &= \frac{d\tilde{F}^{T}}{d\tilde{l}^{T}} \cdot \frac{d\tilde{l}^{T}}{d\tau} = \tilde{k}^{T} \cdot \tilde{v}_{\tau}^{T}; \\ \tilde{v}_{\tau}^{T} &= \tilde{v}_{\tau}^{MT} - \tilde{v}_{\tau}^{M}; \\ \tilde{F}^{M} &= \tilde{F}^{T} \end{aligned} \tag{9}$$

one has with substitution of eq. 7 into eq. 9 the dimensionless musculotendon dynamics

$$\begin{aligned} \frac{d\tilde{F}^{T}}{d\tau} &= \left(\frac{\tilde{k}^{T} \cdot \tilde{k}^{M}}{\tilde{k}^{T} + \tilde{k}^{M}}\right) \cdot \left[\tilde{v}_{\tau}^{MT} - \left(\frac{\tilde{k}^{SE}}{\tilde{k}^{M}}\right) \cdot \tilde{v}_{\tau}^{CE}\right]; \\ &= \tilde{k}^{MT} \cdot \left[\tilde{v}_{\tau}^{MT} - \left(\frac{\tilde{k}^{SE}}{\tilde{k}^{M}}\right) \cdot \tilde{v}_{\tau}^{CE}\right]. \end{aligned} \tag{10}$$

where $\tilde{k}^{MT} \equiv (\tilde{k}^{T} \cdot \tilde{k}^{M})/(\tilde{k}^{T} + \tilde{k}^{M})$ = musculotendon stiffness. One can also show that

$$\begin{aligned} \tilde{k}^{T} &= f_1(\tilde{F}^{T}); & \tilde{k}^{SE} &= f_3(\tilde{F}^{T}, \tilde{l}^{MT}); \\ \tilde{k}^{PE} &= f_2(\tilde{F}^{T}, \tilde{l}^{MT}); & \tilde{v}_{\tau}^{CE} &= f_4(\tilde{F}^{T}, \tilde{l}^{MT}, a(t)). \end{aligned} \tag{11}$$

and thus eq. 10 becomes

$$\frac{d\tilde{F}^{T}}{d\tau} = f(\tilde{F}^{T}, \tilde{l}^{MT}, \tilde{v}_{\tau}^{MT}, a(\tau)). \tag{12}$$

DISCUSSION

The dimensionless dynamics given by eq. 10 (or eq. 12) depends on only one parameter specific to each musculotendon actuator being modeled; i.e., $\tilde{l}_{\tau}^{T}$, which is the ratio of tendon slack length l_{τ}^{T} to optimal muscle fiber length l_{o}^{M}. The two other specific parameters are F_{o}^{CE}, muscle strength, and l_{o}^{M}, optimal muscle fiber length, but these parameters do not appear in the dimensionless dynamics because they are two of the three normalization parameters (the other is $\hat{v}_{m}^{CE}$). Of course, the force trajectory $\tilde{F}^{T}$ also depends on how it is activated, $a(\tau)$, on its kinematic state $\tilde{l}^{MT}$ and $\tilde{v}_{\tau}^{MT}$, and on the initial force. The actual dynamics for a specific actuator is found by converting the dimensionless variables into actual physical ones by using the three normalization parameters.

To illustrate the effect of $\tilde{l}_{\tau}^{T}$ on musculotendon mechanics we show in Figs. 7-8 the isometric musculotendon force-length curve ($\tilde{F}^{T}$ vs. $\tilde{l}^{MT} - \tilde{l}_{\tau}^{T}$) and the build-up in force during an isometric contraction ($\tilde{F}^{T}$ vs. τ) for two realistic (say, lower extremity) actuators; one with a stiff ($\tilde{l}_{\tau}^{T} = 1$) and the other with a compliant tendon ($\tilde{l}_{\tau}^{T} = 5$). The large differences substantiate our assertion that tendon is at times a dominant factor in musculotendon mechanics.

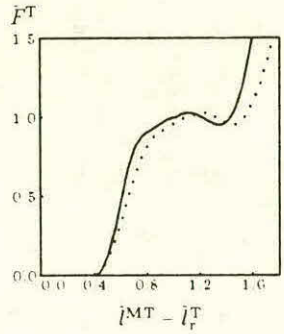

Figure 7.

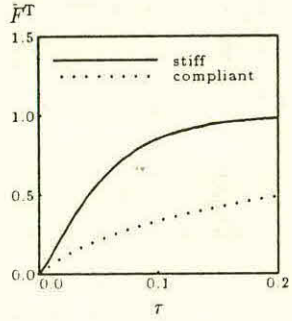

Figure 8.

This model for musculotendon contraction dynamics is, therefore, low-order (first-order), dimensionless, and applicable to any actuator on specification of three specific parameters (F_{o}^{CE}, l_{o}^{M}, l_{τ}^{T}). The model does not change, except in algebraic complexity, if pinnation of muscle fibers is included, though another specific parameter would then appear [8]. Complementary work by us on musculoskeletal geometry is ongoing to find nominal values for l_{τ}^{T} and l_{o}^{M} of human lower extremity actuators (this volume). Since we model activation dynamics as a first-order process as well (this volume), we believe that estimation of the parameters associated with this process (e.g., "recruitment curve" and F_{o}^{CE}) for each stimulating electrode is critical to the estimation of the overall musculotendon dynamics.

ACKNOWLEDGEMENT

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A PLANAR MUSCULOSKELETAL MODEL FOR STUDYING POSTURE INDUCED BY FUNCTIONAL NEUROMUSCULAR STIMULATION

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ABSTRACT

An inverted double-pendulum model of standing has been used to date to study some basic control issues. Using a constant feedback gain as a controller to stabilize a linear representation of this inherently unstable system, we found that the simulated nonlinear system's response was virtually the same as the linear system's one when initial joint offsets from the vertical were imposed. It was also found that this feedback control law functioned well when external disturbances simulating hand-held objects were applied. System response was very sensitive, however, to underestimation of body segment parameters. Finally, we have developed for future use an analytic expression suited to modelling muscle dynamics associated with neural pulse train activation.

INTRODUCTION

Although functional neuromuscular stimulation (FNS) has potential in restoring mobility to paraplegics, problems exist that must be solved before FNS becomes clinically practical. One of the problems is the need to develop a highly reliable control system and to study the trade-offs among alternative control system designs. We have chosen to develop a computer model for use in such studies and design. The composite model to date consists of a musculotendon model, a sagittal plane skeletal model and a musculoskeletal geometrical model.

Here we report our initial attempt to combine these three main elements of the model into the construction of a control problem and our results on observability, controllability and linearization of the control problem. In addition, we have developed a model for musculotendon activation dynamics effected by electrical stimulation which, when combined with our contraction dynamics model (this volume), provides a complete model for electrically stimulated musculotendon dynamics.

SKELETAL MODEL

The skeletal model is planar, accounting for motion in the sagittal plane (Fig. 1). It is composed of five segments, the two feet, the two shanks, the two thighs, trunk and head, and the two upper extremities. Each of these segments is assumed to be uniform and rigid so that the mass center does not change relative to each body-fixed reference frame. For postural control, it is assumed that the feet are kept flat on the ground.

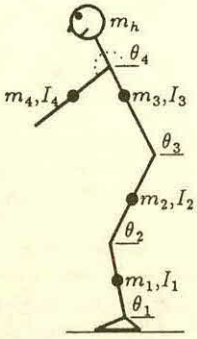

Fig.1 Skeletal Model

MUSCULOTENDON MODEL - ACTIVATION DYNAMICS

Musculotendon dynamics is decomposed into two parts in series, activation dynamics and contraction dynamics (Fig. 2). The input to the activation dynamics, $u(t)$, corresponds to the external stimuli in the FNS train. A rectangular pulse train is used to represent the stimuli. Therefore, $u(t)$ can be characterized by three parameters, pulse amplitude (u_m), pulse width (d) and pulse frequency (f). It is assumed, without loss of generality, that pulse width and pulse frequency modulation are employed by the FNS system. Hence, pulse amplitude is assumed to be constant. The output of the activation dynamics is the activation, $a(t)$, which is defined as the normalized level of active state; *i.e.*, if $a(t)$ is 1, in the steady state the musculotendon actuator, when held at a constant length, will develop a constant force corresponding to its isometric force-length curve, and if $a(t)$ is zero, no muscle force is generated regardless of musculotendon length.

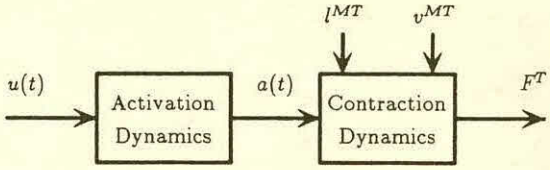

Fig.2 Musculotendon Model

Note that l^{MT} and v^{MT} represent the musculotendon length (origin-to-insertion length) and its shortening velocity, respectively.

We have studied two models for activation dynamics. The first is based somewhat on first-order enzymatic reaction kinetics, a process that presumably occurs in muscle fibers:

$$\frac{da(t)}{dt} = (k_2 + k_1 u(t)) \cdot (u(t) - a(t)). \tag{1}$$

This model has two desirable characteristics: (i) $u(t)$ is linear with $a(t)$ in the steady state and (ii) the developing rate and the relaxing rate of muscle force can be adjusted by the two constants, k_1 and k_2, respectively.

For studying FNS systems, one can also use a simple analytic expression to relate the stimulus parameters to the activation:

$$a(t) = a_m \exp\{-T_c(t - kT_s)\}$$
$$\text{for} \quad kT_s < t \leq (k+1)T_s, \quad k = 0, 1, 2, \cdots, \tag{2}$$

where $T_s = 1/f$ and T_c is the relaxation time constant. In order to account for the recruitment curve typifying electrically stimulated muscle (3), a_m can be expressed as a function of pulse width d:

$$a_m = \begin{cases} a_{max}, & \text{if } d \geq d_{sat}; \\ f(d), & \text{if } d_{thr} < d < d_{sat}; \\ 0, & \text{if } d \leq d_{thr}, \end{cases} \tag{3}$$

where d_{thr} and d_{sat} are constants.

Coupling either form of the above activation dynamics to musculotendon contraction dynamics, we can use the complete musculotendon model to find the actuator force output F^T developed by any input pulse train, given the musculoskeletal system in which the physical constraints are acting on the actuator.

MUSCULOSKELETAL GEOMETRICAL MODEL

For our initial studies of control issues, we have to date greatly simplified musculoskeletal geometry. Later, we will use the more accurate representation for musculoskeletal geometry under development in our laboratory (this volume). First, it is assumed that the knee is locked, say, by a brace; thus the number of degrees-of-freedom is reduced (Fig. 3). Second, the fact that range of motion is quite small in postural control around the upright position is used by us to assume that (i) musculotendon length for each actuator is constant and thus musculotendon force developed by each actuator is also constant and (ii) the moment arms of musculotendon actuators are constant regardless of joint angles.

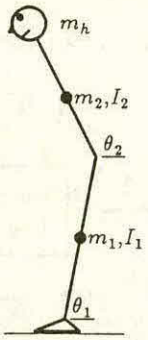

Fig.3 Simplified Skeletal Model

CONTROL PROBLEM FORMULATION

The control problem formulated and studied to date is a much simplified version of the one that we will soon study. Nevertheless, the study of a less complex version is useful for it provides insight into the directions that should be pursued in the future. First, notice that if arm movement is assumed to be quasi-static, one can replace arm movement by a torque and a force acting at the shoulder to see how system performance is affected by movement of the upper exremities. Second, we are assuming at the moment that musculotendon actuators have no dynamics. The tendon force F^T is hence assumed proportional to the level of activation $a(t)$. Third, although we used pulse trains to test the feasibility of the model for activation dynamics, we have not yet used pulses as inputs to musculotendon actuators in the control problem but rather have assumed that the controls are a set of activations, where each $a(t)$ is continuous over $[0,1]$. Fourth, we have assumed that two actuators (an agonist-antagonist pair) control each joint: that is, a triceps surae group and a tibialis anterior group are used to control the ankle, and a single-joint flexor and a single-joint extensor are used to control the hip. Finally, we have assumed that co-contraction cannot occur.

FNS CONTROL SCHEMES

The dynamics described above were linearized and combined to give the linear time-invariant state equations:

$$\frac{d\mathbf{x}(t)}{dt} = \underline{A}\mathbf{x}(t) + \underline{B}\mathbf{a}(t),$$
$$\mathbf{y}(t) = \underline{C}\mathbf{x}(t), \tag{4}$$

where $\mathbf{x}(t)$, $\mathbf{a}(t)$ and $\mathbf{y}(t)$ are the state, the input and the output, respectively. Neither process noise nor measurement noise was assumed to exist.

Constant output feedback was used as the basic control law:

$$\mathbf{a}(t) = -\underline{K}\mathbf{y}(t). \tag{5}$$

Two methods were used to compute the output feedback gain. One was the pole assignment technique (5) by which one can obtain desirable transient and steady state system responses. The other method used was linear optimal control theory for which the problem is to find the optimal output feedback gain that minimizes some performance criterion (4),

$$J = \frac{1}{2}\int_0^\infty (\mathbf{y}^T(t)\underline{R}_1\mathbf{y}(t) + \mathbf{a}^T(t)\underline{R}_2\mathbf{a}(t))dt, \tag{6}$$

where $\underline{R}_1$ is a constant positive semidefinite matrix and $\underline{R}_2$ is a constant positive definite matrix. In order to have the system output follow given command trajectories, we applied another control scheme, called output tracking. Details of how to use this scheme are well illustrated in (1). Briefly, the technique assumes that the command vector, which is the same dimension as the output vector, is a polynomial function of time:

$$\xi(t) = \mathbf{c}_{n_p}t^{n_p} + \cdots + \mathbf{c}_1 t + \mathbf{c}_0, \tag{7}$$

where n_p is the polynomial order and $\mathbf{c}_i$'s ($i = 0, \ldots, n_p$) are constant vectors. The control law then becomes

$$\mathbf{a}(t) = -\underline{K}\mathbf{y}(t) + \sum_{i=0}^{n_p} \underline{H}_i \xi^{(i)}(t), \tag{8}$$

where $\underline{K}$ is determined by either the optimal control law or the pole assignment technique, $\underline{H}_i$'s ($i = 0, \ldots, n_p$) are constant matrices, and $\xi^{(i)}(t)$ is the i-th time derivative of $\xi(t)$.

RESULTS AND DISCUSSION

We found that the analytic expression (2) does convert pulse trains to an analog activation signal $a(t)$, which, when applied to our model of contraction dynamics, produce isometric force trajectories typical of muscle excited by FNS. However, this model accounts for neither tension enhancement found among type F and type S muscle units (2) nor muscle fatigue. Since muscle fatigue is an important factor in designing FNS control strategies, our immediate future work is directed toward modification of our muscle model so as to account for muscle fatigue and energetics.

In order to check how valid linearization is about the upright position, we computed and applied a constant feedback gain (5) to both the linearized and the original nonlinear system. The responses of the two systems associated with the return of the body to the vertical from an initial vertical deviation were compared. The results showed that linearization is valid for a wide range of joint motion ($50° \leq \theta_i \leq 130°, i = 1, 2$). When passive joint torques were added, however, the responses of the two systems differed, since the passive joint torques were modelled as highly nonlinear functions of the joint angles. One might be able to reconcile this difference by decomposing the nonlinear

system into a set of piecewise linear systems with each linear system operational over a defined state space. The control laws described in the previous section worked well when external disturbances, to represent arm movements such as holding a cup, were applied.

We also found that desired transient and steady state responses can be obtained by adjusting the locations of the closeloop poles. It should be noted, however, that in some cases a variety of feedback gains can be used to position the poles at desired locations, though each output feedback gain produces a quite different response. Therefore, one needs to consider how to choose the one that generates the best performance.

Our simulation results also showed, as expected, that it is impossible to stabilize the system by using position feedback only. Hence, either velocity feedback is needed in addition to position feedback, or additional states will have to be estimated, or system stability will have to be improved by adding compensator dynamics.

Since it is desirable to have paraplegics don/doff as few gadgets as possible, it is necessary to minimize the number of sensors attached to their bodies. We have specifically studied the usefulness of measuring ground reaction forces (GRF) and center of foot pressure (such as with sensors in shoes) with particular emphasis on how feasible it is to neglect the horizontal component of the GRF. The results indicate that the horizontal component is much smaller than the vertical component, so the former can be neglected in state variable estimation.

We also found that control laws generated by using segment parameters 10% smaller than the nominal can induce disastrous performances when applied to the nominal system. Performance was especially sensitive to upper segment parameters.

ACKNOWLEDGEMENTS

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A MUSCULOSKELETAL MODEL OF THE HUMAN LOWER EXTREMITY

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ABSTRACT

A model of the human musculoskeletal system is needed to understand the function of different muscles during movement, especially with regards to functional neuromuscular stimulation. We have developed a musculoskeletal model of the human lower extremity to study the effects of musculoskeletal geometry and muscle parameters on muscle torque in the sagittal plane. Results for the static properties of hip muscles show that generally individual muscle torque is maximized since peak muscle force occurs at joint angles corresponding to maximum moment arms. Because peak isometric torque occurs at different joint angles for each muscle, net muscle torque is maintained over a broader range of hip joint angles than that for any single muscle. By describing the torque producing capability of each lower extremity muscle, the model provides a basis for understanding muscular coordination.

INTRODUCTION

Simulation of the human musculoskeletal system is needed to understand the function of different muscles during movement, especially in the context of functional neuromuscular stimulation (FNS). Besides our use of models to understand how muscles are coordinated during standing and walking in able-bodied persons, we can also use musculoskeletal models to determine the optimal muscle coordination to compensate for lost muscle function in disabled persons. Such understanding would greatly contribute to our efforts to design FNS systems for persons with paralyzed lower extremities.

To understand muscular coordination, both the force producing capability of individual muscles and the geometry of the musculoskeletal system must be considered. The influence of musculoskeletal geometry is two-fold: (*i*) musculoskeletal geometry determines moment arms, and therefore, the magnitude of muscle torque generated by a given muscle force, and (*ii*) the amount of force produced by a muscle varies with musculotendon length, which is a function of musculoskeletal geometry.

How muscle force varies with musculotendon length depends also on the slack length of the tendon in series with the muscle fiber; the effect of tendon slack length on muscle torque has not been previously reported.

Our primary interest is development of a musculoskeletal model for application to FNS systems for standing and walking. Presently, our model allows only joint motion in the sagittal plane, although the principles apply equally well to joint motions in other planes. Our model includes muscles of the entire lower extremity, and incorporates both the length and velocity dependent properties of muscle. For brevity, we present only results for the static properties of hip muscles.

MUSCULOSKELETAL GEOMETRY: METHODS OF COMPUTATION

General Approach

The lower extremity was modeled as four planar, rigid-body segments: pelvis, thigh, shank, and foot. Segments were joined by single degree-of-freedom (SDF), frictionless joints at the hip, knee and ankle, so that the instantaneous joint centers of rotation were uniquely determined from the joint angles.

The torque produced by each muscle was the product of muscle force, dependent on the musculotendon length, and the corresponding moment arm.

Calculation of musculotendon length and moment arm were based on straight-line models of each muscle. Muscles were assumed to pass from origin to insertion in a single straight line. For some muscles, a single straight-line model was not adequate, and additional straight line segments which passed through "effective" origin and/or insertion points were used.

Musculotendon length and moment arm calculations required coordinates for the muscle origins and insertions. Origin and insertion coordinate data were provided primarily by Brand *et al.* (1). Each muscle was given a set of three-dimensional coordinates expressed in one of three cartesian reference frames fixed to the pelvis, femur, or tibia (1). Brand *et al.* (1) provides either an actual or an effective attachment for each muscle, but not both. Since effective attachments underestimated musculotendon length, we estimated the missing coordinate data using measurements taken from a dried skeleton.

Vector Subtraction Method

Musculotendon length was calculated by two methods: vector subtraction and integration of moment arm. For muscles which act in one or more straight lines from origin to insertion, musculotendon length can be calculated using vector subtraction. We will consider two cases. First, for a uniarticular muscle modeled as a single straight line, musculotendon length is the magnitude of the vector difference between origin and insertion coordinates. The moment arm is calculated from a vector cross product; however, in our planar analysis the moment arm is used as a scalar.

Let l^{MT} be the musculotendon length, $\vec{OI}$ the vector from origin (O) to insertion (I), $\vec{HO}$ the vector from the joint center of rotation (H) to the origin, and m_h the moment arm associated with joint H (Fig. 1). Then:

$$l^{MT} = |\vec{OI}|, \tag{1}$$

$$m_h = \frac{|\vec{HO} \times \vec{OI}|}{|OI|} \tag{2}$$

Second, for a biarticular muscle acting at joints H and K, we model the musculotendon length as the sum of three straight lines: from actual origin (O) to effective origin (O_e), from effective origin to effective insertion (I_e), and from effective insertion to actual insertion (I) (Fig. 2). The moment arm is calculated using the effective attachments:

$$l^{MT} = |\vec{OO_e}| + |\vec{O_eI_e}| + |\vec{I_eI}|, \tag{3}$$

$$m_h = \frac{|\vec{HO_e} \times \vec{O_eI_e}|}{|O_eI_e|}, \tag{4}$$

$$m_k = \frac{|\vec{KI_e} \times \vec{I_eO_e}|}{|I_eO_e|}. \tag{5}$$

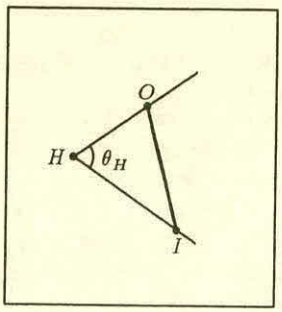

Figure 1.

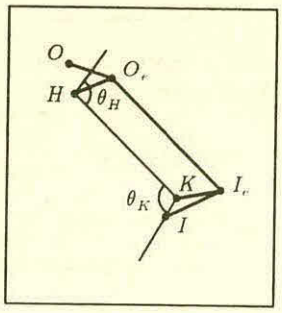

Figure 2.

Integration of Moment Arm Method

For the some muscles, (*e.g.*, rectus femoris and hamstrings) another approach must be taken for calculating musculotendon length. Since the knee's center of rotation varies with joint angle (4, 6) and the patella enhances the moment arm of the knee extensors (4), experimental data were substituted for calculated moment arms for rectus femoris (4) and the hamstrings (6). Musculotendon lengths were calculated using the mathematical relationship between joint angle, moment arm, and musculotendon length.

For a uniarticular muscle crossing a single degree of freedom joint H, the moment arm (m_h) is related to changes in musculotendon length (dl^{MT}) and joint angle ($d\theta_h$) (2):

$$m_h = \frac{dl^{MT}(\theta_h)}{d\theta_h}. \tag{6}$$

For a biarticular muscle the musculotendon length depends on both proximal and distal joint angles, θ_h and θ_k. Again, assuming SDF joints, then

$$dl^{MT}(\theta_h, \theta_k) = \frac{dl^{MT}(\theta_h, \theta_k)}{d\theta_h} d\theta_h + \frac{dl^{MT}(\theta_h, \theta_k)}{d\theta_k} d\theta_k. \tag{7}$$

Because the distance between joint centers is large relative to the moment arms at the joints, we assume that the configuration of one joint does not affect the moment arm at the other joint. Mathematically,

$$dl^{MT}(\theta_h, \theta_k) = \frac{dl^{MT}(\theta_h)}{d\theta_h} d\theta_h + \frac{dl^{MT}(\theta_k)}{d\theta_k} d\theta_k, \tag{8}$$

or, making use of (6),

$$dl^{MT}(\theta_h, \theta_k) = m_h d\theta_h + m_k d\theta_k. \tag{9}$$

Equation (9) shows that we can calculate changes in the musculotendon length of a biarticular muscle as if it were two uniarticular muscles. Integrating (9) we get:

$$l^{MT}(\theta_h, \theta_k) = \int_{\theta_{hr}}^{\theta_h} m_h d\theta_h + \int_{\theta kr}^{\theta_k} m_k d\theta_k + l^{MT}(\theta_{hr}, \theta_{kr}), \tag{10}$$

where $l^{MT}(\theta_{hr}, \theta_{kr})$ is the musculotendon length when θ_h and θ_k are at their reference values, θ_{hr} and θ_{kr}, respectively. These reference values are selected so that $l^{MT}(\theta_{hr}, \theta_{kr})$ can be computed using the vector subtraction method.

Furthermore, since $l^{MT}(\theta_{hr}, \theta_{kr})$ can be computed using vector subtraction for different values of θ_{hr}, then m_h can be calculated using (4), and integration at joint H is not necessary. We choose θ_{hr} equal to θ_h, and (10) becomes:

$$l^{MT}(\theta_h, \theta_k) = 0 + \int_{\theta kr}^{\theta_k} m_k d\theta_k + l^{MT}(\theta_h, \theta_{kr}). \tag{11}$$

MUSCLE MODEL AND SPECIFIC PARAMETERS

Identification of Muscles to be Modeled

To simplify the musculoskeletal model, we reduced the number of muscles to be modeled by neglecting some muscles and lumping others. Muscles were neglected if they functioned primarily in the transverse or frontal planes (*i.e.*, obturator internus), or had a small physiological cross-sectional area and negligible moment arm in the sagittal plane (*i.e.*, pectineus). Muscles were lumped if they had similar moment arms in the sagittal plane, and similar physiological parameters (*i.e.*, semitendinosus, semimembranosus, biceps femoris (long head)). Lumping was accomplished by taking an average of coordinates and muscle parameters, weighted by the physiological cross-sectional area (A^M) of each muscle.

Muscle Parameters

We assert that all muscles and their tendons are described by the same, dimensionless musculotendon model (this volume). The dimensionless model is scaled to individual muscles by three parameters: the maximum isometric strength (F_o^{CE}), the optimal muscle fiber length (l_o^M), and the tendon slack length (l_r^T). Additionally, the pinnation angle at optimal muscle fiber length (α_o) is needed to calculate muscle force. We used data for l_o^M, A^M, and α_o reported by others (3, 7). However, to account for the difference between measured sarcomere length and predicted optimal sarcomere length, we multiplied optimal muscle fiber lengths by 2.8/2.225 (this volume). Since muscle, strength (F_o^{CE}) is proportional to A^M (7), we multiplied the reported A^M by a factor so that the peak isometric hip extensor torque matched experimental data.

No experimental data exist for tendon slack lengths in human muscles. However, indirect experimental data were used to select tendon slack length according to the following criteria:

i. since passive force is developed in muscle fibers only at lengths greater than optimal (this volume), tendon slack length was adjusted so that the joint angles at which optimal muscle fiber lengths occurred coincided with the onset of passive joint torque measured at the hip (8), and

ii. since tendon slack length determines the joint angle at which maximum muscle force occurs, tendon slack lengths were further adjusted so that the sum of the maximum isometric torque curves for either hip flexor or extensor muscles coincided with the *in vivo* maximum isometric hip flexor and extensor torque data, respectively (5).

RESULTS AND DISCUSSION

Eleven hip muscles were modeled to study the effects of musculoskeletal geometry and tendon slack length on maximum isometric hip torque in the sagittal plane. The maximum isometric torque generated by three hip extensor muscles is shown in Fig. 3. Differences in moment arms and muscle-specific parameters among muscles produced isometric torque peaks at different joint angles for each muscle. The operating range of each muscle was unique, and net muscle torque was maintained over a broader range of hip joint angles than that for any single muscle.

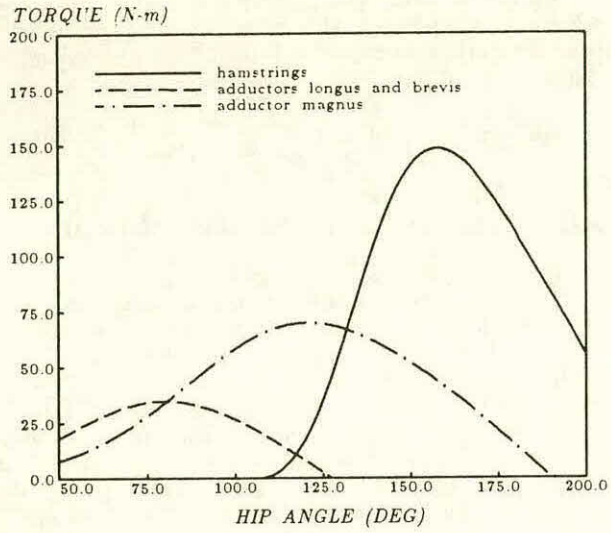

Figure 3.

For each muscle, tendon slack length determined the joint angle at which maximum force was developed. For all hip muscles except tensor fascia latae, tendon slack lengths were such that muscles produced peak muscle force at joint angles near maximum moment arm, thus maximizing muscle torque.

In addition to tendon slack length, muscle torque is a function of optimal muscle fiber length and moment arm. Moment arm determines how musculotendon length varies with joint angle (6), and optimal muscle fiber length determines how muscle force varies with musculotendon length (this volume). Thus, the influence of tendon slack length on muscle torque was much greater in muscles with large moment arms (*i.e.*, hamstrings) or relatively short optimal muscle fiber lengths (*i.e.*, tensor fascia latae) than in muscles with small moment arms (*i.e.*, gluteus minimus) or relatively long optimal muscle fiber lengths (*i.e.*, sartorius).

With our model of the human lower extremity, we have completely specified the muscle parameters and musculoskeletal geometry of each muscle. Since these parameters determine the torque producing capacity of each muscle, the function of different muscles during movement can be studied. Thus, our model provides a basis for understanding muscular coordination in both able-bodied persons and in disabled persons using functional neuromuscular stimulation.

ACKNOWLEDGEMENTS

This work was supported by NIH grant NS17662 and by the Veterans Administration. The authors thank C. Cady and G. Yamaguchi for their assistance in figure preparation.

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CHARACTERIZATION OF A LATERAL PREHENSION-RELEASE GRASP USING FUNCTIONAL NEUROMUSCULAR STIMULATION

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ABSTRACT

The use of force recruitment curves is demonstrated to facilitate the setup of hand grasp elicited by functional neuromuscular stimulation (FNS). A lateral prehension-release grasp with a linear thumb motion and desired contact force vector is determined by cocontracting muscles in patterns established by the recruitment curves of the thenar muscles.

INTRODUCTION

Functional neuromuscular stimulation of paralyzed muscle has been used to provide hand grasps in quadriplegic subjects [1]. A single input command controls the stimulation parameters to several forearm and hand muscles through a coordination algorithm to produce the desired grasp. Previously, coordination algorithms have been developed emperically by general observation of the output characteristics of the electrodes [3]. It is possible, however, to refine a grasp in a more qualitative manner using the measured force and position output characteristics for each electrode. This paper reports on a complete lateral prehension-release grasp that was characterized and subsequently optimized using data obtained from force and position recruitment curves.

METHODS

This study has been performed on one subject (JHJ), a C5/C6 level quadriplegic who has been actively involved in the FNS program since 1978. The subject has 14 percutaneous intramuscular electrodes chronically implanted in his left forearm and hand. The electrodes were implanted in the following muscles: Flexor digitorum superficialis (FDS), Adductor pollicis/first dorsal interosseous (AdP), Flexor pollicis longus (FPL), Flexor digitorum profundus (FDP), Extensor pollicis longus (EPL), Extensor digitorum communis(EDC), and Abductor pollicis brevis(AbPB). JHJ also has a fused interphalangeal joint of the thumb, tenodesed tendons of the FDS, the FDP, and the EDC, and a Zancolli Lasso procedure ot the FDS.

Force and position measurements of the thumb and finger were made to evaluate the lateral grasp pattern.The subject's arm was held in a cast which covered the distal upper arm down to the wrist. The wrist was in a neutral pronation-supination position and slightly extended (10-20 degrees).

Experiments consisted of three to four second periods of stimulation followed by at least 60 seconds between trials involving the same muscle. The applied stimulation was a constant current biphasic pulse with a pulse width between 0 and 200 uS. The interpulse interval and stimulus amplitude were fixed throughout the experiment at values of 80 msec and 20 ma respectively.

Force measurements were made using a two-directional force transducer. The thumb is placed into a machined thumb cast that fits into a cylindrical hole in the end of a metal beam. Thumb force is then measured by strain gages placed on all four sides of the metal beam. These gages are placed in such a way that they measure shear forces and are thus independent of the placement of the thumb on the transducer. The output is in the form of electrical signals that are proportional to the two orthogonal components of the thumb force vector. The force transducer was attached to a stereotaxic frame which allowed the transducer to be fixed in the desired orientation.

During each force trial the stimulation pulse width was stepped from zero to the pulse width being tested and the force output of the thumb was recorded. The static force output for each value of input pulse width was determined by averaging the force output over the last half second of the 3 second trial. The force output generally did not increase after the first second of stimulation. The force output was sampled at 100Hz.

Thumb position transduction was performed using a magnetically coupled ball and socket joystick which was connected to the tip of the thumb by a lightweight metal rod. As the thumb moved through its range of motion it moved the rod connected to the joystick. The output of the ball and socket transducer gave proportional position signals in two orthogonal directions: abduction-adduction and flexion-extension. The position coordinates for the end of the thumb could thus be obtained for each stimulation level. Position trials lasted 4 seconds and consisted of two two-second halves. In the first half of each trial the muscle was stimulated at maximum pulse width so that the thumb was in full extension. Stimulation was then decreased to the value being tested and thumb position was determined to be the average value over the last half second of stimulation. This protocol always kept the thumb movement in the same direction, preventing any inconsistency in the position measurement due to hysteresis.

Finger position was measured using a similar ball and socket device which was positioned on the back of the hand over the second metacarpal. The rod was attached through a slot that protruded above the proximal phalange of the index finger. This device only measures the angle of the metocarpal-phalangeal joint.

RESULTS

The characterization of the entire prehension-release grasp is shown in Figure 1 and consists of three sections: (1) finger closing from 0 to 20% of the command range, (2) thumb movement from

full extension to contact with the fingers from 20 to 50% of the command range, and (3) thumb to finger contact force regulation from 50 to 100% of the command range.

Finger closing

The external goniometer used to measure finger position measured only the angle of the MP joint of the index finger. Because the finger tendons were tenodesed, measurement of the index digit movement reasonably represents the motion of all digits. Figure 2 shows that the position output varies almost linearly with the pulse width input. Because of this, no attempts were made to further linearize the finger motion.

Thumb Motion

Figure 3 shows that the thumb motion with the initial stimulus parameters was highly nonlinear. The thumb moved from full extension to the lateral aspect of the index finger within a few percent of the command range. This was due to a very high gain region in the EPL force output. The FPL and EPL cocontraction parameters were modified [4], with the result that the thumb position moved smoothly and nearly linearly over the 20-50% command range as shown. The entire range of thumb motion requires a variation of only a few microseconds in pulse width of both the EPL and the FPL electrodes.

Thumb Force

Figure 4 shows the force vectors from three thumb electrodes available for use in force regulation. The vectors are measured with the thumb positioned

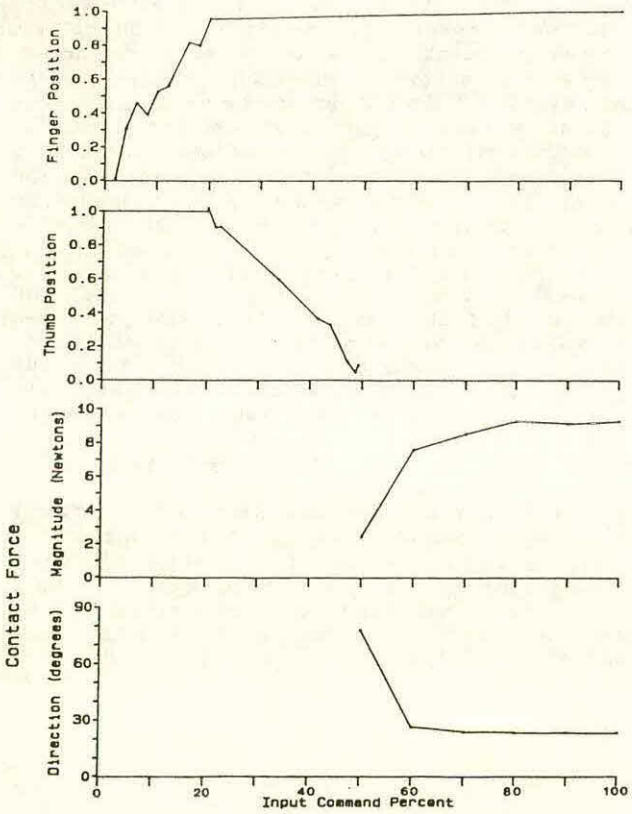

Figure 1. Complete Lateral Prehension-release Grasp.

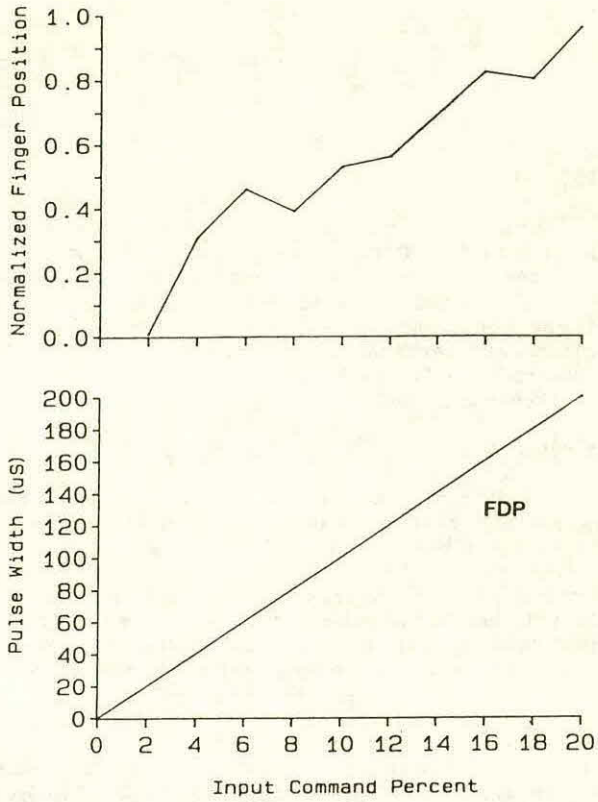

Figure 2. Finger Position Change Induced with FDP Stimulation.

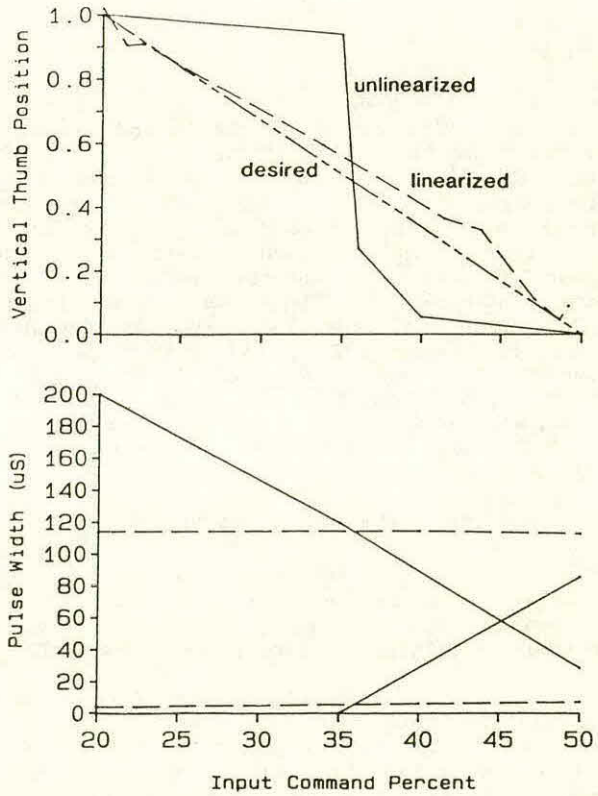

Figure 3. Thenar Position with Change in Input Command.

slightly above the lateral aspect of the index finger. The force vector information allows the selection of both force magnitude and direction by choosing the proper individual vector combinations so that they sum to the desired force vector output. The figure shows that the AbPB and FPL muscles act in similar directions and have comparable magnitude so that it was only necessary to use one of the two in a functional grasp. If a higher pinch force was desired then both muscles could be contracted simultaneously.

Figure 5 shows the thumb force magnitude and direction for the initial and modified hand grasps. It was noted by visual observation that the thumb tended to abduct as it contacted the index finger to such an extent that it impaired the functional use of his lateral grasp. This is evident from the initial force direction curve which shows a direction change of 30 degrees at the 60% command level. In order to eliminate this undesired abduction, a force vector was added in the adduction direction by stimulation of the AdP. As shown in the force direction plot from the modified grasp, this smoothed out the force direction and effectively removed the abduction component at the 60% command level. The initial 80 degree force direction in this plot is at a low force level and does not impair the function of the hand grasp. For this example we did not attempt to linearize the force magnitude. This can be done by varying the coordination algorithm in a manner similar to the position linearization [2]. Since the force direction is constant, it will not be altered by the change.

CONCLUSIONS

The use of force and position recruitment curves has been demonstrated to be useful in the optimization of FNS hand grasp systems. Results have been presented showing the use of position and force vector information to augment one subject's grasp. Force vector information has been shown to allow the choice of both force magnitude and direction by the proper selection of individual electrodes. Position recruitment data has been shown to be useful in the production of a linearized command vs. position output.

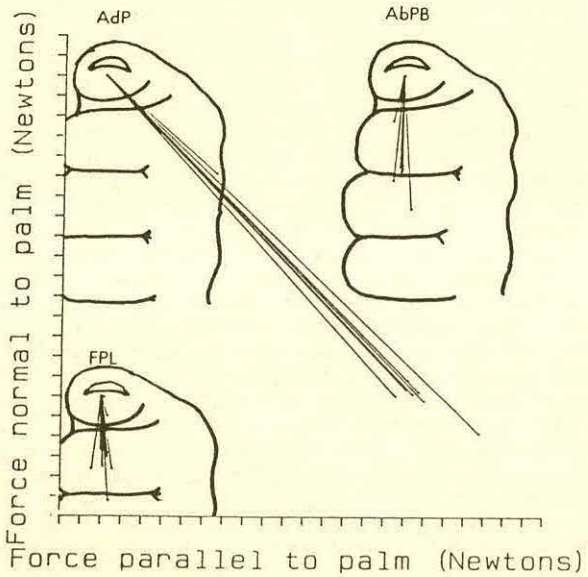

Figure 4. Force Vectors for the Indicated Muscles. Each vector represents a different stimulation pulse width to the respective electrode.

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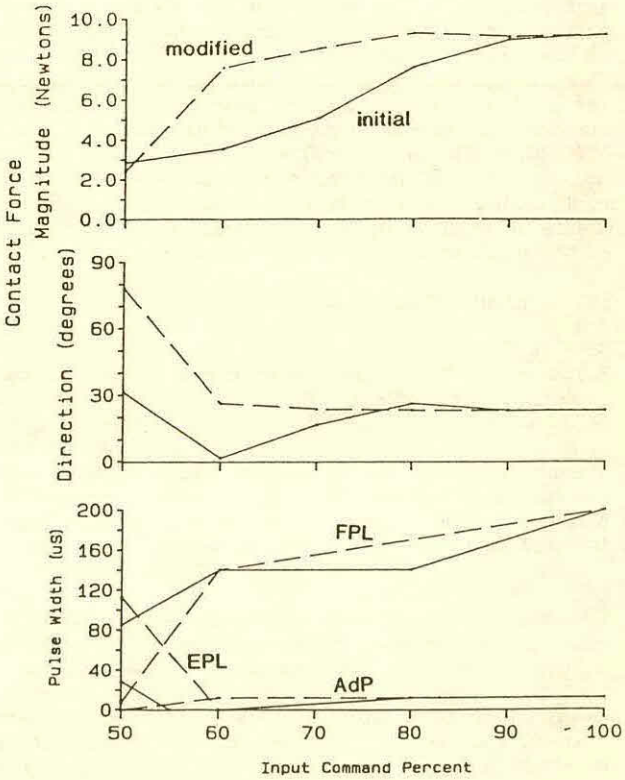

Figure 5. Force Vectors of Thenar Muscles.

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P6.2

EVALUATION OF SHOULDER POSITION AS A COMMAND CONTROL SOURCE FOR USE WITH NEURAL PROSTHETIC DEVICES BY QUADRIPLEGIC INDIVIDUALS

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ABSTRACT

Evaluation of shoulder position as a command control source for orthotic / prosthetic systems. The experimental procedure includes the following areas of study: range of shoulder motion; changes in the range of shoulder motion during postural movement and task performance; use of proprioceptive feedback for maintainance of shoulder position; independence of shoulder movement in horizontal and vertical positioning; variations in the dynamics of fast and slow shoulder movements over the range of shoulder motion; and the ability to make incremental movements of the shoulder and the corresponding number of discrete levels that can be achieved. Experiments have been carried out with two normal subjects and two C-5 level quadriplegics.

INTRODUCTION

The purpose of this study is to evaluate the use of shoulder position as a source for command control signals. Shoulder position has been used for several years as a command control source in upper extremity neural prosthetic devices for quadriplegic individuals(1). The success of the shoulder control in these devices has been varied, depending on the individual. The goal of this study is to characterize the essential elements of shoulder motion for use as a command control source. These studies are valuable for three purposes; for design of more effective command control algorithms, for a better understanding of the limitations that are specific to each individual user with respect to the different levels of injury, and for comparing the effectiveness of the shoulder with other command control sources.

MATERIALS AND METHODS

Subjects.

Subjects for this procedure were two normal and two quadriplegic volunteers. The subjects sat in either a chair or a wheelchair, in front of a video display monitor. Position transducers were attached to the sternum and shoulders with double sided adhesive tape (as described below). The subjects were requested to perform a series of tests described below which measure the essential elements of shoulder movement.

Equipment.

Two-axis position transducers have been developed to measure protraction-retraction and elevation-depression of the shoulder(2). The transducer is arranged in a ball and socket configuration. The ball is moved by means of an attached rod and sensors in the socket transduce movement. The body of the transducer is mounted on the sternum of the subject, and a telescoping rod runs from the ball towards the shoulder and is attached to the skin about half way along the length of the clavicle. The rod is mounted approximately parallel to the clavicle. Two shoulder position transducers were used so that the movement of both shoulders is measured simultaneously.

The experiments are run on a PDP-11/73 computer. The computer reads the transducer outputs, stores the data on disk, and generates a visual display for the subject. The visual display is generated on a 13" color display monitor.

Experimental Procedure.

One shoulder is chosen as the primary command control site. Normally, the command shoulder is contralateral to the extremity used for the neural prosthetic device. Certain portions of the experimental procedure are carried out using both shoulders, whereas most of procedure examines only the command control shoulder. The experimental procedure is outlined as follows:

Range of shoulder motion. The range of motion is defined by the extremes of the volitional shoulder range of the subject. This is the region in which command control tasks can be performed. The range of motion is measured with the subject sitting in a normal upright posture and with both arms down at the sides. The trials are repeated for both shoulders, each while the opposite shoulder is at a neutral (relaxed) position, and then with both shoulders simultaneously. No visual feedback is given to the subject during the range of motion trials. The subject is asked to move to each of the following positions during each trial: neutral (no movement), full elevation, full depression, full protraction, full retraction, or a combination of these (full elevation and protraction for example). The subject is also asked to slowly trace out a full range circle with their shoulder, which encompasses their range of motion.

The range of motion defines the principle horizontal and vertical axes that the subject uses when asked to move along these directions, in addition to the magnitude of the motions. These axes are used to define an experimental reference coordinate system and set of reference points that are utilized in some of the subsequent sections of the experiment. The principle axes of the coordinate system are the two straight lines which best approximate the horizontal (x) and vertical (y) shoulder movements of the subject. This produces a coordinate system that is not nesessarily orthogonal with respect to the transducer, but that is orthogonal with respect to the subject's internal proprioceptive and perceptive reference frame. This coordinate system is used to define five reference points, one where the two axes cross, and one at 75% of the extreme along each of the four coordinate directions (i.e. up, down, forward, and back). In the case of some quadriplegic subjects who have poor protraction / retraction and poor active shoulder depression, this coordinate frame is simplified to include only the neutral and 75% of elevation points, plus a point midway between these two points. The reference points are used so that shoulder movements can be analyzed systematically over the whole range of motion.

In addition to examining the range of motion, we also examine how the range of motion changes with respect to posture and movements of both the ipsilateral forearm

and the contralateral shoulder and forearm. Here the range of motion is measured as above, except that the ipsilateral or contralateral forearm is maintained at certain reaching positions during each of the trials. These movements simulate the conditions that occur during normal tasks performance.

The ability to maintain a shoulder position. This part of the experiment is designed to test the ability of an individual to maintain a desired shoulder position using only proprioceptive feedback. The position of the shoulder is displayed to the subject as horizontal versus vertical shoulder motion. The motion is displayed in the subject's coordinate reference frame. During each of the trials, one of the five reference positions is displayed as a target for the subject. With the aid of visual feedback, the subject is asked to move his shoulder to the position corresponding to the displayed target position. When the subject has reached the target position, the visual shoulder position feedback is blanked out and the subject is asked to try to maintain that position for a specified period of time (5-20 seconds). After this time has elapsed, the feedback is turned back on and the subject is told to relax. The trajectory of the shoulder movement is redisplayed for the subject immediately following each trial so that he can determine how well he is performing.

This section of the experiment examines how well a subject can maintain a constant shoulder position, using only the intact proprioceptive senses. This may be an important factor in determining whether the subject needs to be supplied with additional feedback information. These tests will also establish how long an individual can tolerate holding a constant shoulder position, which is a factor involved in designing a command control technique to lock a prosthetic device at a specific position or force.

Independent control of two axes. This section of the experiment is designed to test the the ability of the subject to control horizontal shoulder movements independently from the vertical shoulder movements. Immediately before each trial begins, the subject is informed exactly what that trial will involve. Like the previous set of trials, one of the five reference points is displayed, and the subject is asked to move to that position. When the display blanks out, the subject makes the predetermined movement. For example, the subject might be asked to move upward and then forward. The visual display then returns and the subject is told to relax. As before, the trajectory of the shoulder movement is redisplayed for the subject immediately following each trial.

The goal of these trials is to determine the feasibility of obtaining two independent command signals from the shoulder. These trials measure the ability of the subjects to control horizontal and vertical shoulder movements independently. If subjects can adequately control two degrees of freedom without having to compensate using visual feedback, then implemenation of a two degree of freedom prosthetic system may be feasible.

Dynamic movements over the range of motion. The next section of the experimental procedure studies the dynamics of fast and slow shoulder movements over the range of shoulder motion. These experiments are identical to the previous set, except that the subject is asked to jerk his shoulder in along the appropriate direction, instead of making normal slower movements. For example, the subject might be asked to move his shoulder to the upward reference point, and then jerk his shoulder forward, or jerk his shoulder down. Again, the subject gets to the target reference position using visual feedback, but does not get any visual feedback during the jerking motion.

The primary goal of these trials is to determine how quick movements vary over the range of motion, and how they compare with the slower proportional command movements. This group of trials will also help us determine when we can discriminate between different types of movements. This is important for establishing what types of logical signals we can derive from the shoulder, in addition to proportional position commands.

Incremental movements of the shoulder. Thefinal group of experiments involves the ability of subjects to make incremental movements of the shoulder. In each of these trials, the subject again has to move to one of the reference points, with the visual feedback display on. From this point, the subject is asked to make a series of the smallest possible incremental changes in shoulder position along a specified direction. The subject is cued every five seconds to make another step in the same direction. In addition to the audio cue, the display blanks out when the subject is supposed to make the movement, and reappears after the movement has been made. The subject steps through part of or the entire range depending on conditions such as the size of the steps or the fatigue of the subject.

The incremental movements of the shoulder give us information about the maximum number of discrete levels that can be achieved using the shoulder as a proportional command source. This is related to how well the subjects can make fine command adjustments. We can also determine if the ability varies over the range of motion.

RESULTS

Figure 1(a) shows the range of motion of a normal subject and figure 1(b) shows the range of motion of a C-5 level quadriplegic subject. These figures show horizontal versus vertical movement of the shoulder. Four separate trials are shown in each diagram, one for a full range movement in each of the four directions: up, down, forward, and back. Each trial starts from the neutral shoulder position. Both of these particular subjects have a larger vertical range than horizontal range. The normal subject has some range in all four directions, but most of the range is up. The quadriplegic subject, on the other hand has very

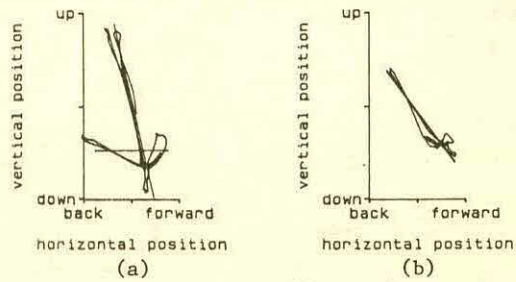


Figure 1. Range of shoulder motion.
(a) Normal subject. (b)Quadriplegic subject.

little horizontal range or shoulder depression, because of weakened musculature, but has fairly strong shoulder elevation. The crossed lines through the trials in figure 1(a) represent the axes that define the subjects horizontal and vertical movements. Figure 1(b) shows only a vertical line, because the horizontal movement is too weak to be consistent. These axes are used to define the reference points that are used through out the experiment.

Figure 2 shows the ability of a normal subject to maintain the shoulder at each of the five reference positions. This figure shows five seconds of five different trials, during the time when the visual feedback is turned off. The circles show the position of the reference points, and the dots within the circles show the position of the shoulder while the visual display is turned off. Most of the trials show very little change over this period, but occasionally the shoulder drifts as shown in the upper trial.

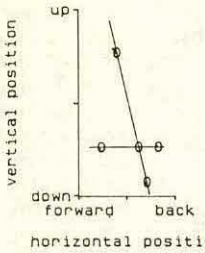


Figure 2. Maintainance of shoulder position.

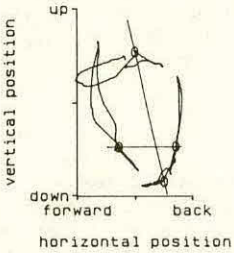
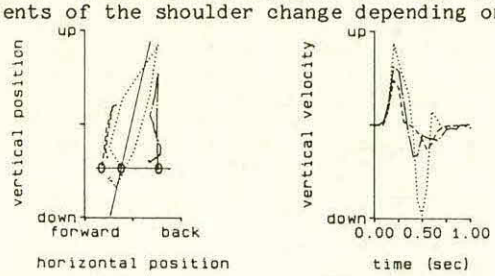


Figure 3. Two axis shoulder movement.

Figure 3 shows the ability of a normal subject to control horizontal and vertical movements independently. In each of the eight trials shown here, the subject was asked to move to the displayed reference point, and then to make a certain movement when the visual feedback was blanked out. This figure demonstrates the subject's ability to move forward and back from the up and down positions, and up and down from the forward and back positions, even when operating near the extremes of the range. This subject has very little range in the forward and down direction, so it is not possible to separate the axes in this region.

Figure 4(a) shows three different trials where the subject was asked to jerk the shoulder quickly upwards, from three different starting positions. Figure 4(b) shows the same movements plotted as the vertical velocity versus time. There is a significant difference between the velocity achieved in the center of the range and the the velocity achieved near the extremes. These results demonstrate that dynamic movements of the shoulder change depending on the

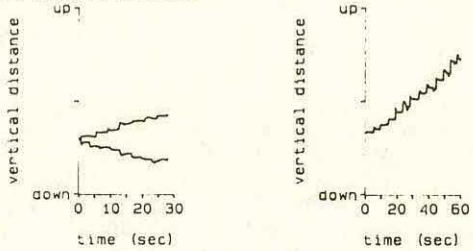


(a) (b)

Figure 4. Quick upward movements of the shoulder.

initial state, and that we need to account for this variability in designing command control techniques.

Figure 5 (a) and (b) show vertical position versus time for incremental shoulder movements. Figure 5(a) shows the upwards and downwards steps of a normal subject. Based on the number of steps in these trials, this subject should be able to achieve approximately 60 discrete shoulder positions along the vertical axis. Figure 5(b) shows a C-5 level quadriplegic subject. This trial shows upwards steps over the whole range of motion. This subject was able to achieve approximately 16 steps over the full range. On a relative scale, the average step size of the quadriplegic are about 1.5 times the average step size of the normal subject.



(a) (b)

Figure 5. Incremental shoulder movements. (a) Normal subject. (b) Quadriplegic subject.

CONCLUSIONS

There are two groups of command signals that can be obtained from a command control source, namely proportional (continuous) and logical signals. To process either type of signal optimally we need to know the characteristics of the signal generator. In this case, a number of different aspects of shoulder motion need to be examined to fully realize its potential as a command source. These experiments are providing a more complete knowledge of how the shoulder performs as a command control site. This data is also giving us a better understanding of how the impairment of shoulder motion affects the performance of quadriplegics in using neural prosthetic hand systems. These experiments will help us to establish the best route to take in the development of better command control techniques.

ACKNOWLEDGMENTS

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IMPLANTABLE GAIT STIMULATION: A SYSTEM OVERVIEW

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ABSTRACT

A sixteen channel implantable stimulation system is being developed to supplement intact neural control of gait in spinal cord and head trauma patients at Rancho Rehabilitation Engineering Center (RREC). The implanted hardware consists of one to two eight channel radio frequency powered and controlled stimulators with platinum disk nerve cuff electrodes. The external patient hardware consists of a small, battery powered, microprocessor-based controller to direct the operation of the implanted stimulators. The controller receives its operating instructions and data from a laboratory computer system over a removable communications link. Stimulation patterns may be quickly determined or modified using the graphics capabilities of a laboratory computer system and transferred to the external controller for use in subsequent runs. Patient performance is measured by the computer system during runs and compared to target performance at the conclusion of each run.

INTRODUCTION

As the efficacy of utilizing multi-channel Functional Electrical Stimulation (FES) continues to increase, so does the need to improve the devices with which it may be applied to spinal cord and head trauma patient population. Clinically, multi-channel stimulation has been realized using surface electrodes in two through six channel systems to help patients in gait training programs (1,2) and through the use of trans-cutaneous coiled wire electrodes, as many as thirty-two channels have been utilized experimentally to help complete paraplegics to take steps, climb and descend stairs (3).

In its development, the use of FES systems for gait assist have encountered a number of problems. One of the problems associated with the use of electrical stimulation in a chronic application is the problem associated with the maintenance of the stimulation system. Electrode lead breakage and other forms of physical damage can lead to a continual and costly operating life for the stimulation system. One of the methods for avoiding this and other problems is to use implantable electronics and electrodes. Although the body can present a rather harsh internal environment, it is nonetheless very stable and free from external abuses. It also provides an excellent opportunity for direct interface between the electronics of the stimulus system and the neuromuscular system of the patient.

An early attempt to utilize an implantable stimulation system is described by a group at Rancho Los Amigos Hospital in 1968 (4). A single channel RF powered device was implanted in the medial thigh of stroke patients and a single bipolar electrode was attached to the post-tibial branch of the peroneal nerve to elicit dorsiflexion of the involved foot during swing phase of gait. This device was successful in that it achieved a great level of function for those patients implanted with the device and provided long term evidence for the safety of the package and electrode configuration.

Recently, multi-channel implantable systems have started to appear. At Case Western Reserve University, Peckham, et al, have developed an implantable eight channel stimulation system to be used in their upper extremity research (5). Their device can supply up to 20 mA and pulse widths from 0 to 256 uSec, the pulse width being controllable via the RF link. A newer version will have five bits of current amplitude control over the RF link as well. Thoma, et al, have implanted a pair of eight channel devices in a paraplegic patient with an electrode placement scheme resulting in four active channels (6). The devices of both groups greatly reduce the amount of external devices required to achieve multi-channel stimulation.

In light of the usefulness of the 1968 RREC application, the safety and effectiveness of the nerve cuff electrodes, and the consistency of the resulting stimulation, and the need for ever increasing number of channels of stimulation required for effective gait applications, and a tremendous need to reduce the external equipment to an absolute minimum, a series of multi-channel implantable stimulation systems are currently under development at RREC. The system described here is the first multi-channel device in our implantable stimulation program.

LABORATORY DESCRIPTION

The gait laboratory is a new facility being developed especially to support the research with the implantable gait stimulation system. The laboratory will house all of the equipment and personnel required to train implant candidates prior to and after surgical implantation as well as to determine gait stimulation patterns used by the external controller to control the implant devices in all modes of operation. The laboratory equipment directly involved with the implant operation can be divided into three categories: the laboratory computer system, the gait walkway, and the implant/external controller/physical therapist controller system (Figure 1).

Laboratory Computer

Prior to an experimental run, a Digital Equipment Corporation MicroVax II computer will be used to formulate control sequences for each of 16 possible channels and to transfer stimulation data table information to/from the external controller via a removable hardware link. During the run, the computer will collect goniometric, electromyographic (EMG), and force data from the

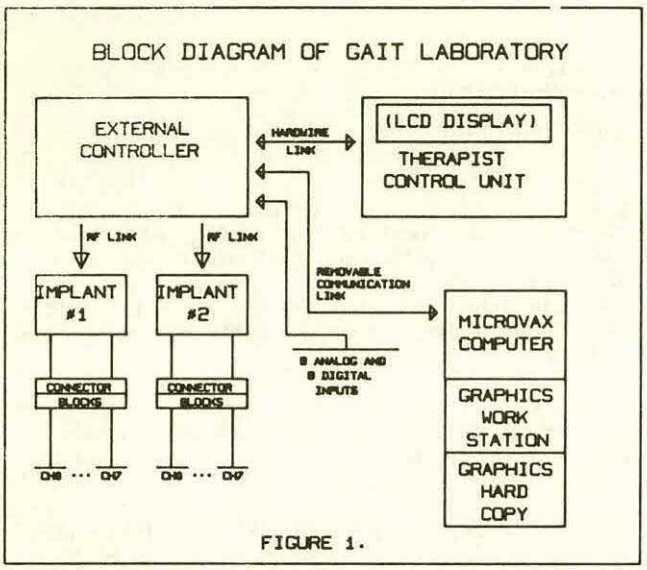


FIGURE 1.

patient as he/she proceeds along the gait walkway during experimental runs. This data will then be analyzed and results presented to the clinicians using the laboratory. In this way, parameters for gait stimulation may be quickly determined and analyzed for modification by the clinicians.

Graphics Workstation

To present the results of data analysis in an easily interpreted manner, and to facilitate the modification of data tables corresponding to gait stimulation parameters, a high speed, high resolution, color graphics workstation will be used. Data table information which corresponds to pulse widths for individual pulses within a channel sequence must be presented in an understandable manner to the clinician making decisions about the modification of stimulation parameters for subsequent runs. By using the power of graphics, the channel information may be quickly reviewed and modified using a "mouse" and sent back to the laboratory computer for processing and transmission to the external controller. Additionally, muscle activity and limb movement simulations will be possible to perform in the future to test stimulation patterns before they are empirically tested with the patient. Also available to the clinician will be a hard copy of the graphics screen information to keep records of patient progress.

Gait Walkway

The gait walkway will be approximately eight meters long and will have color video recording equipment and a Gait Stride Analyzer (7) available for patient performance measurement. Force measurements will be available from instrumented shoes and canes, not from forceplates. Detailed gait analysis complete with EMG measurement is available at the Pathokinesiology Laboratory on the grounds of RREC.

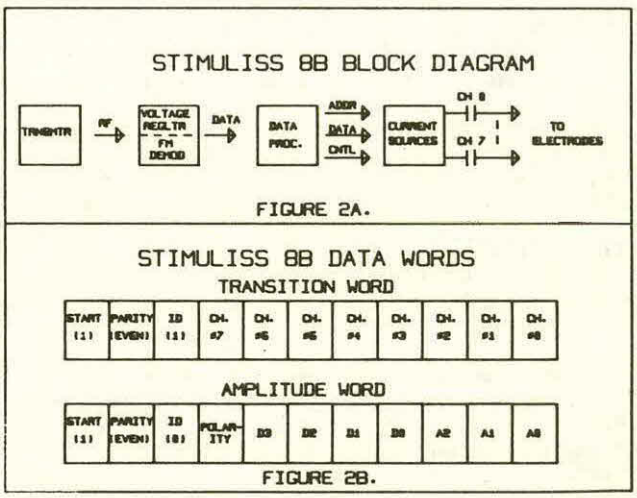
IMPLANT SYSTEM

Implantable Stimulator

The implantable stimulator utilizes a three chip set of integrated circuits developed at Stanford University Integrated Circuits Laboratory, the "Stimuliss 8B" (8). These chips were developed to be utilized in Stanford's auditory prosthesis program funded by the National Institutes of Health. Although the chips were incorporated into a working prototype of the cochlear implant, a final application for the chips has yet to be produced. We were very fortunate to receive several sets of these chips for use in our gait application. The three chips comprise the three system block diagram functions shown in Figure 2. In our application, the chips and their associated electronics are enclosed inside a titanium, hermetically sealed can, with tantalum feedthroughs for electrode and three antenna connections. A shielding technique allows the placement of the receiving antenna directly on top of the can, reducing the package size substantially while sacrificing insignificant signal loss due to the proximity of receiving coil and metal can.

The implantable stimulator receives its information and power from a 20 MHz transmitter modulated with an External Timing Control Code, composed of two words, a "Transition" word, and an "Amplitude" word. Each word uses 11 bits in a bit stream with floating word frames, so the start bit indicates the beginning of a word. A parity bit checks for an odd number of bit errors in a word and thus protects from erratic information being sent to the implant. An ID bit distinguishes between an Amplitude word and a Transition word. The remaining eight bits contain state commands for the Transition word and current level commands for the Amplitude word.

The Transition word is used to change the state of any of the eight channels. If a one occurs in channel bit positions, then that output cycles to its next state of three states, both directions of current and charge balancing (charge balancing means shorting the electrode to body ground, thereby recovering the residual charge left on the electrode). The Amplitude word selects the current amplitude of the state sequence of a given channel. It has three bits to individually address the eight channels, four bits to adjust the current magnitude, plus one more bit to set the current polarity. The magnitude bits set the peak current of the pulsatile waveforms, while the polarity bit selects one of the two possible



pulsatile sequences available for each channel: "sourcing current", "sinking current", then "charge balancing" or "sinking current", "sourcing current", and then "charge balancing".

A special case of the Transition word, the "Master Reset" word, is used to initialize the implant from time to time and is required by the implant to guarantee the integrity of the incoming information.

The implant is approximately 1.75 inches in diameter and 0.375 inches thick. Protruding from the edge of the package are two sets of four leadwires terminating one inch away in two connector blocks. Attached to these connectors are up to eight electrode leads, terminating in monopolar platinum disk nerve cuff electrodes. The platinum disk has a surface area of 1.5 mm^2 and an impedance of approximately 1 KOhm. The implant will be placed surgically in the medial thigh of each leg and transmitting coils will be placed over each device.

In practice the implant will be delivering variable ulse width biphasic pulses, with amplitudes from 200 uA to 2.25 mA. Pulse width is dependent upon the bit transfer rate of the external controller and the controller's discretion. Using a 4 MHz clock frequency, we are able to attain a 500 KHz bit transfer rate from the external controller to the implant devices. This gives us pulse widths ranging from 22 uSec to 300 uSec.

External Controller

Supplying information and power to the implant units will be a microprocessor based external controller which is worn on a belt by the patient. (Please see companion paper describing this device.) This controller utilizes a Complementary Metal Oxide Semiconductor (CMOS) microprocessor, the Motorola MC68HC11, for low power portable operation, and circuitry needed to convert stimulation parameter information into the control words required by the implant and the modulation circuitry to implement the transmission process.

The controller will have sufficient memory to contain pulse information for 16 channels in several modes of operation: transition from sitting to standing, transition from standing to sitting, transition from standing to walking, continuous walking, etc., in the form of data tables. The elements of these tables will correspond directly to the pulse width required for a particular muscle group to produce a desired amount of force. In this manner, a modulated sequence of muscle forces may be defined to realize the desired trajectory of the patient's lower extremities.

Therapist Control Unit

Communicating to the external controller worn by the patient will be a Therapist Control Unit. This device is carried by the physical therapist or other clinician walking with the patient during gait runs. With this device, which is also microprocessor based, the therapist will be able to locally adjust scale factors for any of the channels of the implanted stimulators, to interrogate the status of the implant system, to trigger or modify the operation of the external controller, and to generally assist the external controller in implant system operation. While the external controller is rather devoid of displays and user modifiable controls, the Therapist Control unit will have a two line, 16 character alphanumeric liquid crystal display (LCD) and a number of controls which allow extensive interaction between the therapist and the implant/controller system.

DISCUSSION

The implantable stimulation system and laboratory described above comprise the latest efforts of RREC in the area of gait assist. The flexibility and power of this system will allow research in this vital area to continue at our facility and allow us to make invaluable contributions to this field.

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IMPLANTABLE GAIT STIMULATION: CONTROLLER DESIGN & OPERATION

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ABSTRACT

A small, portable, battery powered, microprocessor based controller has been developed at the Rancho Rehabilitation Engineering Center (RREC) for the purpose of sending coordinated nerve stimulation patterns to a pair of implanted 8-channel nerve stimulators in order to control the different muscle groups used for gait via a trancutaneous RF link. Due to the uniqueness of each gait impaired individual's disabilities, a practical gait controller must simplify the complex interaction required when iterating new gait stimulation patterns. The gait controller accepts and remembers test stimulation patterns for the various muscle groups from a laboratory computer, accepts physical therapist control commands, transmits a bilaterally coordinated nerve stimulation pattern for gait to the individual, collects response data, and allows a physical therapist to interactively modify transmitted patterns. When used in conjunction with a laboratory computer and a clinical team, successful stimulation patterns can be quickly determined and applied.

INTRODUCTION

A key project at the Rancho Rehabilitation Engineering Center has been the development of an external controller for transmitting the critical nerve stimulation patterns required for ambulation in gait impaired individuals. The purpose of this paper is to present the external controller operational requirements and to describe fully the characteristics of the subsystems of the implemented hardware. A companion paper gives an overview of the clinical environment (1) in which the implant system and external controller will be used.

BACKGROUND

In the Rancho gait system, a good part of the procedure to determine the stimulation patterns for gait is a trial and error process, and a practical gait controller must make the process simple. Each muscle group used in gait has a specific set of pulse amplitude, width, and repetition rate values developed for it on a graphics workstation and stored in a DEC MicroVax II computer. The sets are time related (coordinated) to each other in order to produce smooth, dynamically balanced motion in both limbs. Due to the uniqueness of each individual's disabilities, the sets of stimulation parameters must be tailored to the individual. The final nerve stimulation patterns are determined iteratively by evaluating the results of applying the pattern to the impaired individual and modifying the pattern if the desired results are not obtained.

The stimulation patterns above are sent via a removable communications link from the laboratory computer to a portable, battery powered, microprocessor based external controller. A physical therapist (P.T.) has a device (called the P.T. controller) with a communication link with the external controller, above, with which modes may be selected and channel data may be scaled locally. Prior to a trial run, the stimulation patterns are down-loaded into the external controller from the laboratory computer. The down-load communication link is then removed and the clinical team begins evaluating the response of the individual to the stimulation. The patterns may then be modified on the graphics workstation, down-loaded, and then evaluated again. The procedure is continued until desired response has been elicited.

SYSTEM DESCRIPTION

The block diagram of the clinical system is seen in Figure 1. The external controller, the major topic for this dicussion, has interfaces to all system components. There are four input/output interfaces in the external controller system.

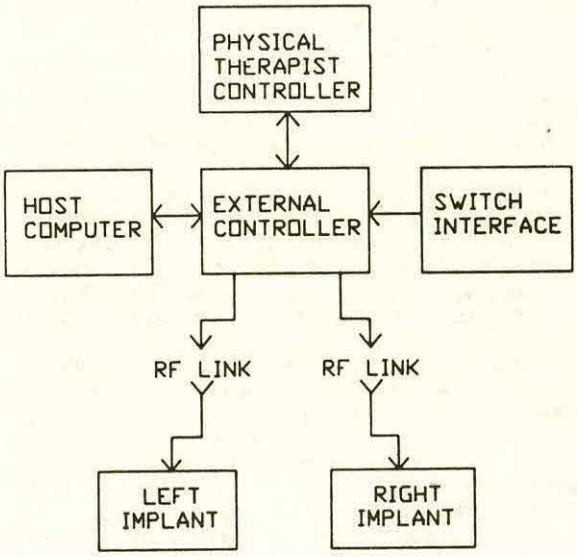

Figure 1. External Controller relative to Implanted Stimulation System

First is a dual RF transmitter to drive the two implanted stimulators. The P.T. Controller interface allows control of the gait sequences, modification of stimulation values, and external controller to host communication. The laboratory computer interface is for the bidirectional upload and down-load of data from external controller RAM memory. The switch interface is for the

collection and conditioning of external switch and analog signals from such sources as goniometers, foot and crutch switches, EMG amplifiers, etc.

The P.T. Controller communicates with the external controller through a three wire serial interface. The P.T. Controller allows the operator to select a number of gait modes, such as, exercise modes, standing, sitting, quiet standing, level walking, walking up and down inclines, etc. The operator is allowed to modify the amplitude of the stimulation for each channel by manipulating a scale factor which multiplies the output amplitude of any specific set of stimulation values by a constant. The system status of both the P.T. Controller and the external controller can be displayed on a Liquid Crystal Display (LCD) on the P.T. Controller.

The labortory computer interface is made from a standard two wire RS-232, full duplex, 9600 baud communication link. The cable is easily removable and is removed after the stimulation patterns have been down-loaded.

The switch interface is designed to take a mixture of digital data and analog data. The digital data is from foot and crutch switches, but would be amenable to any TTL level type switches. The analog data input is principally goniometer data, but many analog signals could be conditioned and accepted as input.

The hardware subsystem classifications are as follows: 1) CPU Board, 2) Switch Interface Board, 3) Formatter Board, 4) Bit Pumper Board, 5) Dual Transmitter Board, 6) Packaging. Figure 2 is a functional block diagram of the subsystems.

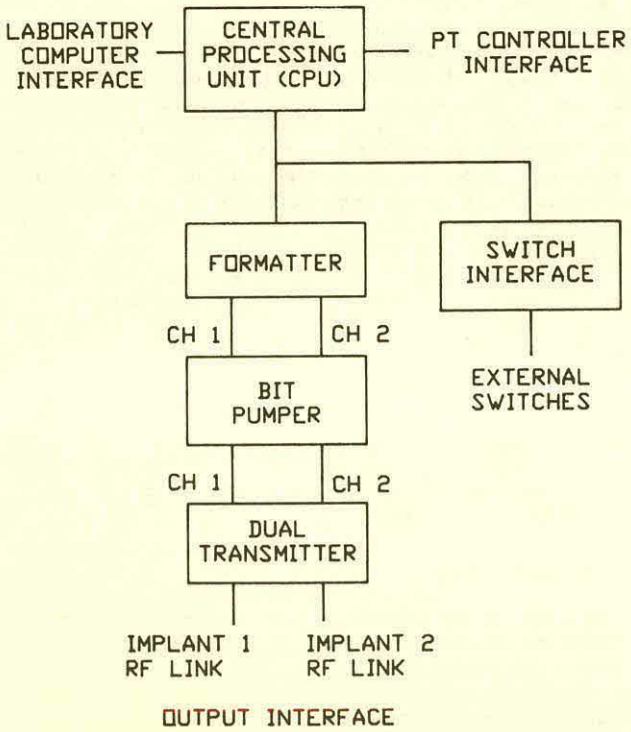

Figure 2. Subsystem Interconnect

1. CPU Board. The Central Processing Unit (CPU) board is responsible for data manipulation, subsystem timing, and communication with the laboratory computer and with the P.T. Controller. The microprocessor selected for the CPU is the Motorola MC68HC11. A major factor in the selection of the MC68HC11 for this application was its low power consumption due to its fabrication in CMOS. On-board the chip is Read Only Memory (ROM) used for program storage, a small amount of Random Access Memory (RAM) and electrically alterable ROM, two serial Input/Output (I/O) ports, an analog to digital converter (A/D), an interrupt processing structure, a clock generator, and a timer.

Commands received from the laboratory computer and from the P.T. Controller are processed by the microprocessor. In general, the program instructs the CPU to look for a command from the P.T. Controller. Then, upon recognition of a valid operation command, the CPU takes its instruction from another portion of the program dedicated to performing the command operation. Upon completion of the command operation, the CPU is instructed to return back to the portion of the program which has the CPU looking for a command from the P.T. Controller.

The chip has an on-board asynchronous Serial Communication Interface (SCI) port through which the laboratory computer is interfaced whereas a synchronous Serial Peripheral Interface (SPI) port is used for the P.T. Controller interface. The baud rate of the SCI is 9600 baud so data may be transferred to/from the laboratory computer at approximately 960 characters/second. The SPI clock rate is 1.0 Mhz, and so during gait operation commands and responses between the external controller and P.T. Controller may be processed very quickly with little effect on pulse timing.

The RAM requirements for storing the data tables are quite large. Initial estimates of memory requirements to support the stimulation patterns for several modes of operation, 16 channels deep, is greater than 90,000 bytes. To accommodate this, there are four pages of 32K (1K=1024) bytes for a total memory capacity of 128K bytes. The RAMs are, as is the CPU, constructed with CMOS technology for low power consumption. In order to achieve the packing density needed because of restricted board size, the chips are piggy-backed with the chip select lines being connected separately.

The microprocessor on-board timer is used to provide a time base interrupt that maintains the orderly deliverly of sequential stimulation data (pulse amplitude, pulse on/off, channel, right or left transmitter) to the formatter circuitry. The timer has about ten related 16 bit registers for a number of timing functions of critical importance to the external controller system.

The Bus interface is used to communicate with the other boards in the system. Only two other boards are controlled by the CPU: the Switch interface board and the Formatter board. The Formatter board drives the Bit Pumper board and the Bit Pumper board drives the Dual Transmitter board. the bus interface is a bidirectional buffer and is

capable of both sending and receiving data.

2. Switch Interface Board. The Switch interface board accepts eight external analog inputs as well as eight external digital inputs. Trigger sites used to initiate stimulation sequences will vary from individual to individual depending on the disabilities, so to maintain maximum flexibility, the inputs are configured as generalized inputs which can be defined and configured under program control. The analog inputs can also double as quasi digital inputs since the allowable input voltage range is limited from zero to plus five volts and a window could be defined easily to correspond to states of interest. The analog signals are converted to eight bit words continuously by a CMOS analog to digital converter.

3. Formatter Board. The responsibility of the formatter board is to sequentially send eleven bit control words defining the generation of a single stimulus pulse on a single channel to the bit pumper board. The eleven bit control words used by the implants are described in detail in the companion paper (1).

The implant accepts basically two types of information: an 11 bit transition word and an 11 bit amplitude word. Transition words contain information about the channel selected, the type of word (a transition word in this case), and a word transmission error check. It can also contain a master reset code. Amplitude words are used to set the amplitude and polarity of the pulses, but they also include a transmission error check and a bit that identifies the word as an amplitude word. By cleverly combining the codes that turn on (or off) the various output driver transistors (polarity control), and by setting the amplitudes on consecutive transmissions, a current waveshape of almost any form can be generated.

A typical sequence of control words might be as follows: 1) Send out a master reset transition word, 2) Send out a transition word selecting channel 5, 3) Send out an amplitude word that specifies a positive polarity and an amplitude of 2 ma, 4) Send out an amplitude word 300 microseconds later that specifies a positive amplitude of 0 ma (effectively turning the pulse off). The resulting waveform from the preceeding sequence would be a 2 ma positive pulse that is 300 microseconds wide occurring in channel 5.

The CPU sends the formatter board two bytes (which makes a 16 bit word of which only 11 bits are used by the formatter) for each word to be sent to the implant. Timing information corresponding to word delivery by the bit pumper board circuitry and pulse width is stored in 16 bit registers for formatter on-board regulation of data transfer timing.

4. Bit Pumper Board. The bit pumper board takes the 11 bit words from the formatter board and converts it into a pulse width modulation signal which will gate the 20 Mhz carrier from the dual transmitter board giving bursts of 20 Mhz carrier of two precise widths. The bit pumper is a dual channel device capable of simultaneously providing modulation for implant 1 and implant 2.

5. Dual Transmitter Board. The transmitter board simultaneously sends out radio frequency (RF) signals to both implants. The bursts of 20 Mhz RF provide both power and stimulation information for the implants. Information to the implant takes the form of narrow pulses of RF representing zeros and wide pulses of RF representing ones. The time to transfer one bit is 2 uSec.

The RF energy is transferred to the implants via flat disks of coiled wire 2.5 inches in diameter. The antennas are supported directly over the implant and they are large enough so that some misalignment is allowed. Each transmitter uses, on the average, 700 milliwatts of power and represents the greatest use of power in the external controller.

6. Packaging. The external controller package size is 3.25" thick x 6" wide x 6.25" deep. It holds five printed boards and a mother board. The battery pack for the controller is a separate removeable module measuring 1.5" x 3.25" x 6" and mounts on the top edge of the package. A strap goes around the back of the individual's neck and the package rests in front at the waist. Cabling to trigger switches, antennas, and miscellaneous analog and digital signals comes from the bottom of the package. There are two telephone style headset jacks which allow the laboratory computer and the P.T. Controller to be plugged in.

DISCUSSION

The external controller described in this paper plays an integral part in the gait implant program at RREC. Due to its powerful microprocessor and flexible design, it can easily be configured and programmed to control the complex operation of two independent implantable stimulation devices. The numerous interfaces and open design of this device provides flexibility, and the generous memory space allows great amounts of stimulation data to be manipulated. Even the operating nature of the device may be changed by utilizing RAM (in addition to EPROM) for operating instructions as well as data, this being transferred via the laboratory computer communication link. Thus custom fitting the operation of the implants to the needs of the patients and research studies will be a valuable asset to this new and important development in functional electrical stimulation gait assist research.

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CONSTANT POSTURE RECLINING SEATING SYSTEM

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ABSTRACT

A wheelchair seating system is being developed to allow the user to alter seating position without altering the seat base/back angle, thereby redistributing pressure on the skin without disrupting posture. The device may be attached to currently available reclining wheelchairs and the motorized linear displacement mechanisms used on power recline wheelchairs.

INTRODUCTION

Clinical observations indicate the repeated use of reclining seat back wheelchairs reduces the occurence of decubitus ulcers (pressure sores) while presenting new problems to the user population. Repetition of the reclining motion significantly alters posture of the seated individual so that the user slides forward on the seat base and assumes a slouched posture. This requires the user to seek assistance in reassuming proper posture. The motion of the reclining seat back creates skin shear, a transverse force on the skin due to friction between the cushion and the skin. Skin shear may further contribute to pressure sores, and requires the addition of complex mechanisms to displace the seat back of the reclining wheelchair.

The purpose of this project is to design a seating system that provides a redistribution of pressure on the user/wheelchair interface without creating skin shear, or interfering with the posture of the seated individual. A review of current wheelchairs reveals no designs meeting this requirement suitable for adult populations. However, the S.T.S. Travel Chair provides this function for pediatric applications. The Travel Chair consists of a rigid seat base/back assembly which pivots with respect to the wheelchair frame. Constant posture is maintained by the rigid seat, and skin shear is avoided because the seat does not translate with respect to the user. It would be desirable to configure a similar seating system (Figure 1a and 1b) suitable for adult as well as pediatric populations. With the aid of rehabilitation engineers, a physical therapist, a local wheelchair dealer, and users of currently available reclining wheelchairs, we determined functional objectives for a seating system of this type. The objectives are to provide a seating system that:

- adaquately redistributes pressure without interfering with posture or creating skin shear.
- can be easily adapted to a variety of users of diverse dimensions.
- is operable without assistance by users who have upper limb paralysis.
- can be easily attached to available reclining wheelchairs without reconfiguring the chair.
- does not interfere with normal wheelchair operation.
- allows attachment of common accesories for the user population such as cushions and control systems.
- does not provide the potential for injury to the user or individuals in contact with the user.
- is structurally sound, and stable in all possible configurations.

MATERIALS AND METHODS

Wood and foam core prototypes were constructed and evaluated in meeting the functional objectives. These models were refined with input from rehabilitation engineers, a physical therapist and users of reclining wheelchairs. Analysis of reclining wheelchairs provided information concerning wheelchair center of gravity and stability. Anthropometric data was used to determine the location of the user center of gravity and body weight load on individual components of the design (1,2). Dimensions were provided from wheelchair catalogues and anthropometric publications.

Final frame design was achieved with the aid of a CAD system, and finite element analysis was used to provide stress/strain information for material selection. The finite element analysis included the ninety-fifth percentile male body weight of 220 pounds (3). Two models were analyzed, one with full body weight on the seat base and the other with full body weight on the seat back. Both models accounted for a one thousand pound force due to the motorized recline mechanism. Stresses calculated were less than three times the yield strength of the material.

A final design prototype of the rigid frame and pivoting mechanisms was constructed from T-6062 aluminum tubing of one inch and seven-eighth inch outer diameter (0.083 inch wall thickness), one-eighth inch aluminum plate, commonly available aluminum clevises and plastic saddles. Oil impinged bushings that allow rotational motion about two axes were used to provide a pivot point and attachment of the rigid seat to the wheelchair frame. This prototype was attached to a reclining wheelchair (figures 2 and 3). Mathematical analysis of the completed prototype indicates stability of a Rolls Invacare wheelchair in a static condition if the system is reclined to 90 degrees (seat base becomes vertical). The wheelchair may become unstable if operated in a reclined state, requiring the use of wheelie bars or a kill switch to prevent a potential for injury.

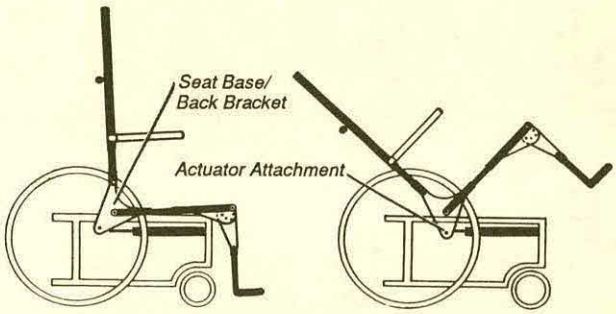

Figure 1a: Resting Position *Figure 1b: Reclined Position*

FINAL DESIGN DESCRIPTION

Assembly

The Constant Posture Reclining Seating System consists of a rigid tubular frame independent of the wheelchair frame, attachments for the linear displacement drive or other power mechanisms, and an axle and bearing assembly. The rigid frame is attached to the wheelchair frame at the seat base/back

intersection by the axle and bearing assembly. The assembly of the seating system on a reclining wheelchair requires the removal of the wheelchair seat back, arm rests and sling seat or pad. Two bearings in cushion blocks are bolted to the horizontal members of the wheelchair frame where the standard reclining seat back was bolted. The axle spans the bearings and is continuous with the rigid frame. Standard wheelchair seating pans and cushions are attached to the frame to provide a seat back and base. The tubular frame is attached to the motorized linear displacement device by two aluminum plates, one on each side of the seat back/base intersection. These plates allow the linear displacement actuator to rotate the frame about the axle. A drawing of the complete configuration appears in figure 4.

Tubular Frame

The tubular frame imitates the functions provided by the wheelchair frame and provides a rigid structure to be rotated about the pivot. The frame is constructed of aluminum tubing of one inch and seven-eighths inch outer diameter, allowing the attachment of accessory devices and control systems by conventional means. Armrests are attached to the frame and maintain the arms in a constant position with respect to the body as the system reclines. The armrests are easily removed or rotated with conventional pin systems. Leg rests also maintain the legs in a constant position with respect to the body as the seat is reclined. The foot height can easily be changed, and the leg rests can be removed or rotated out of the way. The seat back and base are bolted to the aluminum tubing, and a head rest can be attached above the seat back. The frame is assembled by telescoping seven-eighth inch diameter tubing inside one inch tubing and bolting or spot-welding the assemblies together. The one inch diameter tubing connects by the use of clevises at the seat back/base junction. Eight pieces of one inch diameter tubing are necessary for the assembly; these can be stocked in standard sizes. Fine dimension adjustments can be made by varying the overlap of the tubing. The seat back/base angle is maintained (ninety degrees in the prototype) by two aluminum plates. These plates also allow the attachment of the power recline drive.

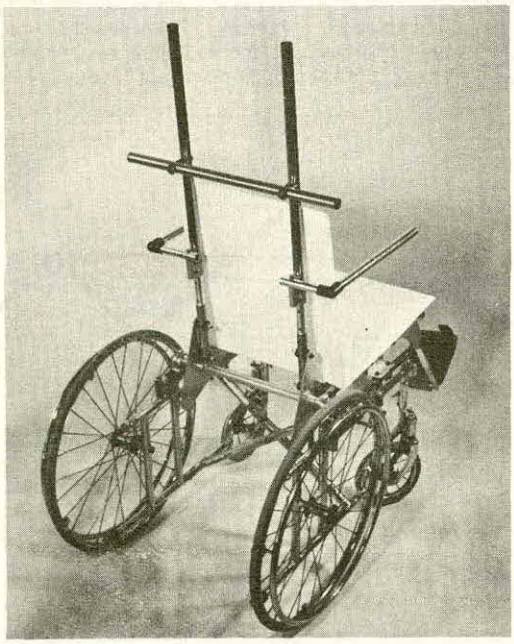

Figure 2: Prototype Mounted to Reclining Wheelchair

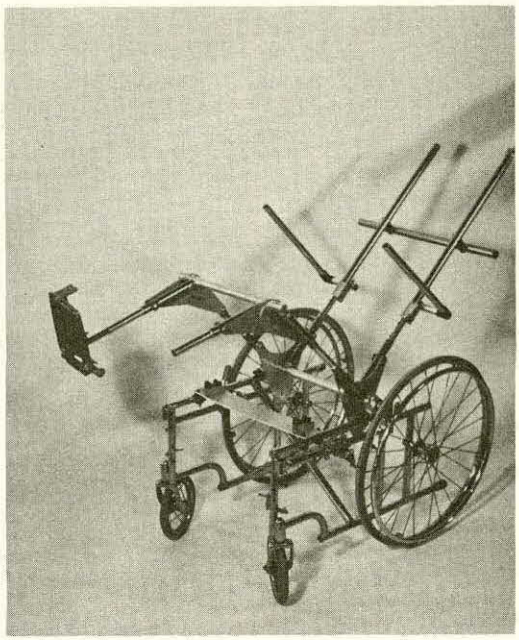

Figure 3: Reclined Prototype

An axle spans the frame at the seat back/base intersection, providing an attachment to the frame as well as the pivot point for the recline motion. The axle rests on two bushings which allow rotation about two axes. These bushings allow the frame to recline if the axle is not perpendicular to the bearing plane. This safety feature allows function of the unit if the seating system or wheelchair tubing bends.

Power System

A motorized lead screw actuator used in power recline wheelchairs attaches to the front of the existing wheelchair frame at a pivoting pin joint. The lead screw extends beneath the seat, and displaces a stainless steel rod which is fixed to the aluminum plates. This configuration resembles that of the current reclining wheelchair, requiring little modification of the existing frame. The actuator requires a twenty-four volt power supply, and switching systems from power recline wheelchairs can be used to operate the device. The actuator rotates the rigid frame about the pivot point by applying a force to the brackets that maintain seat back/base angle. The actuator must be able to generate a 1000 pound force to recline a 220 pound adult. To eliminate the possibility of damage to the lead screw assembly and plates, the system is to be operated in the resting position only (figure 1a) . A platform is provided so that the seat system is supported by the wheelchair frame, minimizing load due to operation on rough ground.

DISCUSSION

Initial review of the completed prototype with rehabilitation staff reveals that the completed prototype achieved the design criteria set at the onset of the project. The actual ability of the system in achieving pressure relief and maintaining constant posture will require evaluation. Analysis of the forces at the user/seat interface show significant pressure relief will be achieved in the reclined position. However, we have yet to measure the actual pressures associated with various seating positions. Performance-testing on an actual power wheelchair is necessary to prove the reliability and performance of the system. After the completion of reliability and performance evaluation, a refined and proven prototype will be evaluated by members of the user population.

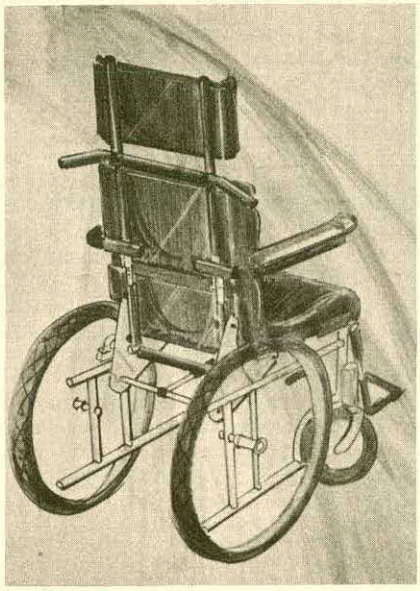

Figure 4: Drawing of Complete Configuration

ACKNOWLEDGEMENTS

This project was funded by The University of Michigan College of Engineering as part of an interdisciplinary engineering and industrial design course. The authors sincerely thank everyone involved in this project, especially: Don Guk Shin, Ph.D. for creative assistance, Mark Allen for industry input, the Design Laboratory Staff for expert consultation, the Rehabilitation Engineering staff at the University of Michigan Medical Center, Lexan Kreyt, Physical Therapist and Brian West for the user's point of view.

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3. Abraham, S., Johnson, C.L. and Najjal, F. Weight and height of adults 18-74 years of age. Vital and Health Statistics, Series 11, Number 211 (1979a)

ADDRESS

Mark W. Capper
Pulmonary Division
3916 Taubman Health Care Center
University of Michigan Medical Center
Ann Arbor, Michigan 48109-0360

HOUSING

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

Demonstrates adaptions required in average dwellings, both apartments and houses, to meet needs of wheelchair bound occupant, and shows how domestic independence can be achieved with a minimum of expense. At the same time it shows many of these modifications would be welcome in a home for non-handicapped. Film then addresses the fact that the home cannot be treated in isolation, but as part of the community picture.
Review: Film received extensive praise for its superior approach to problems of housing and its addressing the economic aspects of special design.

The Netherlands, 16 mm., color, 30 mins., 1974. Producer: Burgwal Films. Sales distributor: Netherlands Society for Rehabilitation. Rental from Rehabfilm: Members: $45.00; Non-members: $60.00.

NO STEPS TO CONQUER

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

A tour of a 13-flat apartment complex built for the disabled by the John Grooms Association in London. Complex developed for residents confined to wheelchairs. Demonstrates many features of a well-planned housing unit for the disabled, including Fokus-designed kitchen, Wessex hoist, Clos-o-mat toilets, and so on.
Review: Good overview of barrier free designing, with relatively little footage spent on self-promotion.

England, 16 mm, color, 20 mins., 1974. Producer: William Hamer Production for John Grooms Association. Sales distributor: William Hamer Productions. Rental from Rehabfilm: Members: $45.00; Non-members: $60.00.

DISABLED ACCESS TO TECHNOLOGICAL ADVANCES

Kay Houston
Access to Independence, Inc.
Madison, WI, USA 53704

This video-tape shows how computer technology can be used to enhance employment options in a variety of fields for persons with severe physical disabilities. Featured are three people using computer technology to help achieve their employment goals: a counselor using computer retraining drills and training in word processing; a planner using a lap computer with a speech synthesizer for communication as well as a keyboard emulator to access the office computer; a law student describing how word processing and access to legal databases will make him more productive. Other employment related issues are also discussed.

There are two versions of the videotape--one directed toward social service providers and one directed toward employers. Copies may be purchased from:

Access to Independence, Inc.
1954 East Washington Avenue
Madison, WI 53704
608-251-7575 voice and TDD

DISCUSSION WITH DR. FRANK BOWE

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

Discussion edited from a presentation made by Dr. Frank Bowe at AT&T. Focus of the discussion is the attitude towards disabled people of private sector employers. Discussion covers what needs to be done to increase employment figures, job evaluation suggestions, reasonable accommodation and possible areas of support by state rehabilitation agencies.

USA, U-matic, color, 30 mins.,1981. Producer: American Telephone and Telegraph Corporate Television. Rental from Rehabfilm:Members: $35.00; Non-members: $50.00.

VOCATIONAL REHABILITATION ENGINEERING

Leah M. Ross
Rehabilitation Engineering Center
2021 North Old Manor
Wichita, Kansas 67208

Several successful vocational rehabilitation engineering modifications are described, the majority of which are for industrial machine tools in a factory setting. Modifications in clerical settings are described as well as one designed for an agricultural worker.

The appropriate audience for this presentation is the rehabilitation professional as well as industrial and technical personnel who are unfamiliar with rehabilitation engineering applications.

This production is available in either 1/2 inch VHS or 3/4 inch U-matic format for rental or purchase. The Rehabilitation Engineering Center is the distributor.

THE BOTTOM LINE

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

Dramatization of hard-pressed head of small company who is trying to replace a plant supervisor. Personnel manager suggests a man with right qualifications who is a wheelchair user. Through persuasion and a meeting with a wheelchair user who is successful in business, the prospective employer changes his attitude.
Review: Convincing hire-the-handicapped film. Touches on all the usual employer concerns, is well-acted. Highly recommended to personnel managers, EEO officers and employment advocates.

USA, 16 mm.,color, 15 mins. Producer and sales distributor: Learning Systems, a Division of Mississippi Methodist Rehabilitation Center. Rental from Rehabfilm: Members: $45.00; Non-members: $60.00.

TECHNICAL AIDS IN SWEDEN

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

General overview of current technical aids devised to assist in functional indpendence of the elderly, stroke patients, and individuals with multiple sclerosis. The film emphasizes electrical eqiupment such as devices for pageturning, wheelchair propulsion, remote control units of lights and appliances, and alarm systems of those who live alone. Also demonstrates the use of phone-dialing equipment, a foot-operated typewriter, and a chin control wheelchair. Narrator describes diagnosis of individual and then demonstrates appropriate device.
Review: Good straightforward overview of technical aids now employed for a number of chronic somatic disabilities. Appropriate for clinical use, as well as patient education.

Sweden, 16 mm, color, 20 mins., 1976. Producer: and sales distributor: Hildegard Halliing, M.D. Rental from Rehabfilm: Members: $45.00. Non-members: $60.00.

SHELTERED WORKSHOPS

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

Survey of a number of competitive sheltered workshops in Australia, investigating problems of management and employees. Many interviews with workers, managers, intercut with work scenes and teaching scenes. Includes some information on barrier-free design and aids to daily living as they apply to work setting. Disorders depicted include paraplegia, cerebral palsy, various forms of mental or developmental retardation.
Review: Excellent introduction to the subject to which the film is confined. Useful in any setting where employment of the handicapped is an issue.

Australia, 16 mm., color, 26 minutes, 1971. Producer and sales distributor: Australian Commonwealth Film Unit. Rental from Rehabfilm: Members:$45.00; Non-members: $60.00.

REHABILITATION ENGINEERING ACCESS TO EMPLOYABILITY

Leah M. Ross
Rehabilitation Engineering Center
2021 North Old Manor
Wichita, Kansas 67208

The history of a disabled individual for whom modified assistive devices were designed and fabricated is described. The individual was enabled to perform light bench work on small electronic components with the assistance of rehabilitation engineering. There is a human interest emphasis.

The appropriate audience for this presentation is rehabilitation professionals, vocational counselors, and/or disabled individuals and their advocates.

This production is available in either 1/2 inch VHS or 3/4 inch U-matic format for rental or purchase. Rehabilitation Engineering Center is the distributor.

"A DIFFERENT APPROACH"

SOUTH BAY MAYORS' COMMITTEE
FOR EMPLOYMENT OF THE HANDICAPPED
Manhattan Beach, CA 90266

The film is a scripted comedy. It shows a fictional government committee which has hired a young, avant garde filmmaker to produce their new film on employment of the handicapped.

On viewing the first answer print of what they expect will be a typically maudlin, emotional appeal to hire the handicapped, they are shocked to see the subject treated as a musical comedy, with irreverent humor, a singing and dancing handicapped chorus, today's top television celebrities and the story of a film within a film.

"ADA" has garnered twenty awards for excellence, including an OSCAR nomination. It has excited the handicapped as well as the professional rehabilitation community more than any other motion picture of its kind.

THE ROAD BACK TO FARMING

Medical Center Rehabilitaion Hospital
Box 8202 University Station
Grand Forks, North Dakota 58202

Media: 3/4" videotape, color Length: 30 min.
Audience: General public, disabled farmers, bankers, rehab professionals, DVR & service clubs.
Description: This videotape presents 8 farmers who suffered disabilities & successfully returned back to farming. The objective of the videotape is to increase awareness that farmers, through adaptation & other adjustments, can successfully carry out farm operations. Examples of equipment adaptations & other adjustments individuals have made in order to return to their farming operation are shown. The case histories include 3 females & five males. Diagnoses include BK amputation, paraplegia, fore-quarter amputation, stroke, arthritis, fibrositis & multiple trauma.
Cost: Rental 1-Week (may be applied to purchase price)$40/Purchase:$350.
Produced, edited & photographed by Karen Bommersbach, Multimedia Coordinator & developed by the Office of Education & Professional Standards, Sharon Rask, Director, Medical Center Rehabilitation Hospital.

WHEELCHAIR FACTORY

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

Tour of a factory at Japan Sun Industries, Beppu, where disabled workers, most of them wheelchair-bound, using modified equipment, manufacture and distribute wheelchairs. Shows design, testing and manufacture of wheelchairs. Also describes the original condition of the factory and the adaptations made to make production possible. Particular attention is paid to a ball-link machine operated by a cerebrally-palsied man with poor hand coordination. A number of wheelchairs are shown, including one which can be used for boarding planes, another with an attachment for bracing a wheelchair on an escalator. There is also a tour of the Tetra-Ace house completely adapted for unassisted living for below C-6 quadriplegics. Recommended to vocational rehabilitation workers.

Japan, 16 mm., color, 20 mins., 1977. Producer: Y. Nakamura, M.D. Sales distributor: Japan Sun Ind. Rental Rehabfilm: Members: $45.00. Non-members: $60.00.

JAPAN SUN INDUSTRIES

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

Documentary on the programs at Japan Sun Industries in Beppu, Japan touching on residential, employment, social, recreational and technical assistance in rehabilitating their clients for a return to society. Assessment, wheelchair selection, competitive industry in which the handicapped work as well as own stock, tool adaptation, the Tetra-Ace model house for quadriplegics with a wide variety of technical aids, are all demonstrated. Employment includes radio production for Sony and precision manual furniture building.
Review: Good general film on rehabilitation in Japan, though marred by simplistic narrative and awful music. Worthwhile for its selections on technical aids and employment.

Japan, 16 mm, color, 30 mins.,1975. Producer and sales distributor: Japan Sun Industries. Rental from Rehabfilm: Members: $45.00. Non-members: $60.00.

BLIND IN INDUSTRY

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

Presents the cases of three totally blind workers in an automotive, cosmetic, and electronic factory setting, and the managers who hired them. Bruce works four to five semi-skilled jobs in Ford Motor Company. He was hired under pressure by a reluctant manager, who is now satisfied and expresses his desire to hire more people of same caliber. Co-workers concur. Mrs. Skyhill works for Elizabeth Arden Company in production and has raised the productivity of the entire unit through her enthusiasm. Employers are consistently satisfied.
Review: Excellent film. Interviewees are credible, the film is well-edited and professional. Recommended to anyone demonstrating employability of the handicapped.

New Zealand, 16 mm., color, 12 mins., 1974. Producer: Reynold's Television Ltd. Sales distributor: Royal New Zealand Foundation for the Blind. Rental from Rehabfilm: Members: $45.00; Non-members: $60.00.

INTERRUPTIONS

The Stanfield House Films/Media
P.O. Box 3208
Santa Monica, CA 90403

Seven individuals whose lives were interrupted are featured along with the employers who gave them the opportunity to prove that people with disabilities also have capabilities.

THE DISABILITY MYTH II

Stanfield House Films/Media
P.O. Box 3208
Santa Monica, CA 90403

Humor is enlightening, and flashbacks to a performance by Ed Rice, a disabled comedian whose wry comments about his disability produce great response from his audience and insight, make this film particularly engaging. The drama, however, is in the high unemployment rate facing disabled persons. With the majority of the disabled unemployed, this film explodes the popular myths which are, in part, responsible for the appalling statistics. Narrated by John Hurt, of variety of encounters are shown where disabled individuals are facing the myth dilemma.

POWERED MOBILITY FOR THE VERY YOUNG CHILD

Mette Norgaard
Everest and Jennings
3233 East Mission Oaks Blvd.
Camarillo, CA 93010

Motorized wheelchairs offer immobile children a safe and efficient method of independent movement. They can learn to drive skillfully, with minimal professional invovlement, within a short period of time and at a much younger age than one would have been thought possible. Powered mobility motivates further movement behavior by the children, and like ambulation, promotes their intellectual, social and emotional development.

In this program, Charlene Butler, Ed.D., and Susan Harris, Ph.D./RPT, express their opinions and philosophies on when, how and why to provide children with powered mobility, and its benefits. Complete transcripts of the program and copies of this video are available through your local Everest and Jennings representative. It is an enlightening video program with little time spent on E and J promotion.

THE EFFECTS OF MOTOR DEPRIVATION ON LEARNING

Danny McCulloch, P.&O.T. and Margaret Martin, O.T.
Hugh MacMillan Medical Centre, 350 Rumsey Road;
Toronto, Ontario, Canada M4G 1R8

The fact that the cerebral palsied child is deprived of many experiences is universally known. We should also consider how motor deprivation affects the acquisition of conceptual, perceptual and behaviour skills. A brief overview of normal motor development leads to specific discussion of sucking/rooting, vision and grasping. The development and reflex learning and accommodation is compared with the pathological reflex and movement patterns of the child with cerebral palsy. Bower states that "every intellectual capacity, even these characteristics of the most cognitive development appears in a simplified form in the sensormotor period". The sensori-motor components of egocentrism, object permanence, cause and effect and social learning and behaviours are developed with specific references to work done by Monnier, Rosser and Shantz.

The role of occupational/physical therapy in remediation is discussed including facilitating the child to replicate the normal patterns of development, providing sensory programs to diminish hypersensitivity and to increase body awareness, positioning to encourage eye-hand coordination, interfacing to educational toys and activities and providing independent mobility at an early age. Based on a blending of neuro and behavioural psychology and therapy expertise and experience, the authors feel that the above observations and treatment management concepts are of major clinical importance.

RACHAEL'S TEAM

William Whiteford & Susan Hadary Cohen
University of Maryland at Baltimore
Department of Physical Therapy
Video Services

RACHAEL'S TEAM is a 22 minute program which includes interviews of Rachael's parents,educators, and health professionals. Working together as a team, these adults in her life describe their in-put to supporting Rachael in the mainstream. Video footage includes special transportation, specialized seating, use of the computer, and use of the prone board -- all in the educational environment. Demonstrated by the program is potential for independent living and special education.

NO MORE SPECIAL ME

The Kids on the Block, Inc.
822 N. Fairfax St.
Alexandria, VA 22314

The Kids on the Block, Inc. is an internationally known company that teaches children about disabilities and differences through the use of puppetry and educational curricula. There are presently over 700 organizations implementing Kids on the Block programs in 49 states and 12 countries.

"No More Special Me" was filmed as a T.V. special in Australia and won First Prize in Special Education at the 1982 Rehabilitation International Film Festival. The Kids on the Block puppets encourage non disabled children to learn about their disabled peers. "No More Special Me" shows this live interaction as well as discusses the philosophy and intent of The Kids on the Block.

"No More Special Me" is available for rental and purchase and is accompanied by a Leader's Guide for Workshop and Classroom Activities.

FOR THE HAND OF A CHILD

H. Clifford Chadderton
The War Amputations of Canada
2827 Riverside Drive
Ottawa, Ontario, K1V OC4

The bio-engineering research project was conducted by the University of New Brunswick and funded by government, private corporations and charitable institutions.

The video details the evolution of the myoelectric arm, starting with the large and awkward electronic devices of the 1960's through to the sophisticated systems available today. The challenge for researchers was to design an arm small and durable enough for infants and small children.

Interviews with the parents of Champs Jonathan Rector and Sara Kerr add insight into the complexities of artificial limb design. Both youngsters received myo-electric arm fittings at the University clinic.

This documentary follows the search for the ultimate in an artificial replacement for the human hand - one that can be fitted for a very young child.

CURRENT MYOPROSTHETIC DEVELOPMENTS AT THE HUGH MACMILLAN MEDICAL CENTRE

M. Mifsud, H.R. Galway, M. Milner
Hugh MacMillan Medical Centre, Rehabilitation Engineering Department; 350 Rumsey Road; Toronto, Ontario, Canada M4G 1R8

Four new devices developed to augment the fitting of myoelectrically operated upper-extremity prostheses are presented.

1) A miniature data acquisition device (DAD) to record both the number of open and close events and the cumulative event duration has been developed. DAD can be placed within the prosthesis to objectively collect information regarding prosthesis use during the amputee's normal daily activities.

2) A self-adaptive myoelectric processor (SAMP) is a 3-state system developed to: facilitate the fitting and training process; optimize the calibration settings as required; motivate prosthesis use through immediate operation; and minimize health care cost and disrupted schedules for recalibration purposes.

3) A microcomputer-aided myoelectric control assessment program (MCAP) developed for routine clinical use to objectively measure, in the clinic, amputee control of a visual hand image with their prescribed myoelectric prosthesis.

4) A microcomputer-based data acquisition system to augment the process of identifying suitable muscle control sites. Graphical and numeric display of the myoelectric voltage signal levels from up to 2 separate muscle groups is complemented with the display of previous results obtained allowing mapping of the muscle bulk area.

PROSTHETIC TERMINAL DEVICE FOR PLAYING THE PIANO

D.J. KOESTER, S.P. LEVINE, W.S. JOCZ, K.D. BUI

UNIVERSITY OF MICHIGAN MEDICAL CENTER
Ann Arbor, MI 48109-0032

This project involves the development of a device to enable a person fitted with a below-elbow prosthesis to play the piano. The client, R.M. is a ten year-old girl who has both natural ability and a strong desire to play music. She has a congenital deficiency of the left arm, a disability equivalent to a short below-elbow amputation.

We continue to work closely with R.M., her piano instructor, family, and Rehabilitation Engineering staff. Early in the project a prototype was constructed to allow R.M. to practice with a "two fingered" left hand. Hard work and effort on the part of R.M. and her instructor are evident in the video presentation and have continued. This prototype, which has fixed finger positions, is to be replaced by a more sophisticated device with variable finger spread and wrist rotation. Technical difficulties have delayed delivery of this device; however, the delay has given R.M. time to develop a large repertoire from which she has performed in recitals with her peers.

Recently an interim device that enabled finger spread via cable to a foot-control has been delivered to R.M. Due to the success of the interim prototype and client acceptance, a design similar to this device will be finalized.

PROSTHESIS PLUS TECHNIQUE EQUALS INDEPENDENCE PARTS I AND II

MEDICAL CENTER REHABILITATION HOSPITAL
Box 8202 University Station
Grand Forks, North Dakota 58202

Media: 3/4" videotape, color Length: Part I 30 min./Part II-37 min. Audience: Upper extremity amputees & their families, physicians, P.T.'s, O.T.'s, students, prosthetists, DVR counselors, nurses, health care educators, home health care providers & health care professionals.

Description: Blaine, a bilateral upper extremity amputee (left-fore quarter amputee, right short above elbow) explains & demonstrates techniques for ADL's & various other activities. Demonstrations are given in bathing, shampooing, dressing, personal hygiene, toilet activities, cooking, homemaking, driving hints, leisure time, & shop activities.

Cost: Rental-1 Week (may be applied to purchase price)$60.00/Purchase:$375.

Produced, edited, & photographed by Karen Bommersbach, Multimedia Coordinator & developed by the Office of Education & Professional Standards, Sharon Rask, Director, Medical Center Rehabilitation Hospital.

BEAD SEATING SYSTEM (BSS)

Douglas A. Hobson
University of Tennessee, Memphis
Rehabilitation Engineering

The video presentation shows the step-by-step process involved in the client assessment, system fabrication, and final fitting of the Bead Seating System (BSS). The BSS emphasizes therapy evaluation and the use of simulation in the decision making process.

The Bead Seating System utilizes the vacuum consolidation process in the direct fitting of custom contoured seating components. Both seat and back modules are fabricated independently and finished using a simplified vacuum forming process. Total time for fabrication is less than one working day per system.

This video presentation is basically for seating clinicians. Copies of the video can be obtained from the University of Tennessee, Memphis - Rehabilitation Engineering Program or RESNA on a rental basis.

EASY ROLLING

MEDICAL CENTER REHABILITATION HOSPITAL
Box 8202 University Station
Grand Forks, North Dakota 58202

Media: 3/4" videotape, color Length: 18 min.
Audience: Wheelchair users, family members, & health care professionals.
Description: Program shows wheelchair maintenance and repair: tightening of spokes, truing of wheels; removal, packing & installation of bearings; proper tightness of rear and front wheels; & replacement of handrim clips. Safety tips for falling forward and backward are also included. Program has descriptive visuals showing step-by-step procedures for easy understanding.
Cost: Rental-1 week (may be applied to purchase price) Purchase-$125

Produced, edited, and photographed by Karen Bommersbach, Multimedia Coordinator & developed by the Office of Education & Professional Standards, Sharon Rask, Director, Medical Center Rehabilitation Hospital.

ONE GIANT STEP

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

The film is a basic teaching seminar on the use of technical aids for those concerned with rehabilitating physically disabled persons. A range of aids available, from environmental control systems to communication aids are shown in the home, the workplace, chronic care homes and recreation facilities. With such aids the disabled person can become more independent --- turning on and off lights, radio and television sets, making phone calls and unlocking doors.

In one scene, a farmer who is quadriplegic, is shown operating a tractor with the use of technical aids. In another scene a young girl, born with only one arm, is fitted with a prosthetic limb and taught to activate the fingers of her new hand.

Canada, 1982, 16 mm/videocassette, 40 mins. Produced by Canadian Rehabilittion Council for the Disabled. Rental in the United States by Rehabfilm: Members: $45.00; Non-members, $60.00.

MINSPEAK

Bruce Baker and Jan Eisenbarth
Council for Exceptional Children
1920 Association Drive
Reston, VA 22091

Bruce Baker and Jan Eisenbarth discuss the fundamentals of Minspeak, and iconic computer interface technique for speeding up man-to-machine communications. Semantic compaction, whose commercial name is Minspeak, is a widely used system for communication enhancement. It employs icons along with numbers and letters to enable its physically and/or cognitively impaired users to communicate more rapidly. This videotape explores the fundamentals of the system and shows a brief communication interaction between Ms. Eisenbarth and a novice user, Mr. Anthony Brown, of Corapolis, Pennsylvania. Some of the standard Prentke Romich control interfaces, are demonstrated as well.

All Rehabilitation professionals interested in communication disabilities may wish to watch this clarifying explanation and demonstration. Copies of this program can be obtained from the Prentke Romich Company, 1022 Heyl Road, Wooster, Ohio 44691, (216)262-1984.

MULTICHANNEL SPINAL CORD STIMULATION FOR MOTOR DISORDERS

Joseph M. Waltz, M.D.
Department of Neurosurgery
St. Barnabas Hospital
New York, NY 10457

During the past 10 years we have used spinal cord stimulation in the treatment of over 850 cases with motor disorders. This therapeutic application has been expanded by us to include cerebral palsy and other related neuromuscular disorders, such as dystonia, torticollis, and degenerative diseases. Examples of the therapeutic effectiveness of spinal cord stimulation in these various disorders will be shown. IMPROVEMENTS NOTED:
MOTOR FUNCTION-SPASTICITY, ATHETOSIS, DYSTONIC POSTURING, ADDUCTOR & PAINFUL SPASMS DECREASED; RANGE OF MOTION, HAND FUNCTION & DEXTERITY IMPROVED.
POSTURE-SPINE STRAIGHTER, SITTING BALANCE, HEAD & NECK STABILITY & TORTICOLLIS IMPROVED.
WALKING-ATAXIA DECREASED; STABILITY, BALANCE & GAIT PATTERN IMPROVED.
PSEUDOBULBAR-SPEECH, SWALLOWING & DROOLING IMPROVED.
BLADDER-CONTROL IMPROVED.
Results indicate that the spinal cord level stimulated, the electrical field polarity and the stimulation frequency are critical and must be individualized regardless of etiology. These analyses can be done by the patients primary physician after the implant by the neurosurgeon.

CHANGES

CRYSTAL PRODUCTIONS
P.O. Box 12317
Aspen, CO 81612

A sensitive definition of the unique problems and lifestyle changes which follow spinal cord injury. Changes is a intensely subjective and evocative study. Photographed entirely from wheelchair point of view, by a paraplegic, the film's story is told by the patients, graduates, and medical staff of the Craig Rehabilitation Hospital. An outstanding educational and awareness tool for the general public, students, hospital staff, relatives of the injured, and patients. AWARDS: Winner of SILVER HUGO, Chicago International Film Festival; BRONZE AWARD, International Film and TV Festival; CERTIFICATE OF PARTICIPATION, National Educational Film Service.

VHS & BETA II	#840	$89.50
3/4" U-MATIC	#940	$134.50
16MM	#140	$300.00

16MM Rental $35.00 (3 Days)

OUTSIDE

CRYSTAL PRODUCTIONS
P.O. Box 12317
Aspen, CO 81612

Everyone wants a life that brings satisfaction. As all able-bodied persons face decisions about their futures, the handicapped person must also make choices following rehabilitation as to goals and life's work, even though those choices may be physically limiting. In this film, people with spinal cord injuries, who feel they were not deprived of choice because of a physically disabling accident, talk about the diverse and challenging opportunities that were open to them in their new life outside the rehabilitation center.

This upbeat film about altered lives by Barry Corbet, cord-injured filmmaker and author, has allowed the people who are meeting their goals to talk about themselves, their working, playing, loving, laughing, and living. It is not only an inspiration to the handicapped but to the able-bodied.

VHS & BETA II	#846	$89.50
3/4" U-MATIC	#946	$134.50
16MM	#146	$300.00

16MM Rental $35.00 (3 Days)

"IT'S A NEW DAY..."

SOUTH BAY MAYORS' COMMITTEE
FOR EMPLOYMENT OF THE HANDICAPPED
Manhattan Beach, CA 90266

A celebration of the new attitudes and technologies available to disabled people to increase their integration into the mainstream of life.

Using an original song performed by singer/composer Danny Deardorff, the film highlights disabled people using new devices such as a vertical-lift wheelchair, talking calculator, Opticon-scan and Porta-printer. They are seen in diverse and challenging life styles, including playing tennis and racquetball, dirt-biking, and performing such tasks as forest ranger, engineer, airline reservations agent, school teacher and psychologist.

A CRY FOR HELP

Gregory Johnston, AVS, Carrie Tingley Hospital,
PO Box 25447, Albuquerque, NM 87125-0447

This documentary was developed for a parent action group, Parents Reaching Out, in an attempt to bring the long-term needs of medically fragile children to the attention of the public and New Mexico law makers. It was influential toward the passing of a bill to establish a state funded program for the care of severely handicapped children in New Mexico.

This video tape is available in 1/2" VHS format for purchase.

THE GOOD NEIGHBORS OF ST. CLAIR COUNTY

Joe Dzenowagis; Macomb-Oakland Regional Center,
49599 N. Gratiot, Mt. Clemens, MI 48045.

Produced for the State of Michigan Department of Mental Health, this documentary examines the controversial and emotionally-charged issue of community placement for mentally ill people.

Mentally ill people, parents, community leaders, human service professionals, and neighbors of group homes are profiled in this 20 min. tape, available to community organizations for education and information purposes.

Recipient of six awards, including the President's Council on Employment for the Handicapped for Best Coverage of Disabilities, and First Place, National Media Awards.

LAINEY

Elaine Trefler, OTR
University of Tennessee Rehabilitation
Engineering Program; 682 Court Avenue;
Memphis, Tennessee 38163

There are thousands of school age children in the United States who are being excluded as active participants in their educational process. These are individuals whose intellectual abilities are within or above normal levels, but who are so severely handicapped that they cannot voluntarily use their body parts to speak, write, manipulate objects, sit or move independently.

This videotape presents the results of a cooperative effort between the Prentke Romich Company with a subcontract to the University of Tennessee Rehabilitation Engineering Program. The system developed is a modular component system which provides appropriate powered mobility, seating, control and interface with a communication aid and computer. It depicts the impact of technology on a young lady and her family and how she has been able to obtain a degree of independence.

EMERGING

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

Emerging is a celebration--without narration--of the emergence of society from the disabled dark ages into a present era of rehabilitative enlightenment. Theme number one is architectural barriers. We move on to communication, featuring Blissymbolics, a symbol system used by cerebrally palsied children. Next, Ability. We watch a recreation of the famous movie scene in "Best Year of Our Lives" in which a handless pianist plays. Also blind dancers and scenes from an Olympics. Open your ears during the Olympics footage-the sound editing is fabulous. Finally, a wedding and family life, implying successful integration.

Canada, 1980, 16 mm/videocassette,color, 16 mins. Produced by the Canadian Rehabilitation Council of the Disabled. Distributed for rental in the United States by Rehabfilm: Members: $45; Nonmembers: $60.00.

AN ORDINARY LIFE

The Stanfield House
P.O. Box 3208
Santa Monica, CA 90403

A look at some of the advances which are helping the disabled to find a voice and a place in our society. In an integrated school, devices are shown which assis t both teacher and student to achieve a normal educational environment. In a computer consultant firm, the nine shareholders are disabled. The film also visits a totally accessible shoppin g center and a community living center in which residents have 24 hour assistants to help them maintain an ordinary life. The computer has revolutionzed access to information for the disabled who could not have this access with books and papers. In the community, a center which develops devices for making homes more usable and comfortable is demonstrated.

AGAINST ALL ODDS

Stanfield House Films/Media
P.O. Box 3208
Santa Monica, CA 90403

This film features three creative achievers, accomplished professionally and inspiring to others. Roseanne La Flamme is a winning Olympic Athlete and ski instructor. Melody Elliot Clarke i s a professional artist/choreographer. Wayne Pronger is a somgwriter/composer. Roseanne is also a triple amputee, Melody is a paraplegic and Wayne has severe cerebral palsy. This film confronts such issues as acceptance and understanding , marriage relationships, psychological growth, vocational decisions.

YOU HAVE GOT THE POWER

Stanfield House Films/Media
P.O. Box 3208
Santa Monica, CA 90403

An affirmative statement about the limitless boundaries of human resiliency concerning five individuals who have something in common. Each of them has an arm or a leg amputated because of bone cancer. Each has had to overcome the rage and self-pity resulting from their profound sense of loss. Each has confronted the questions any of us would ask if we were placed in their situation - Will I still have my friends? Will I ever lead a normal life or will I just be a freak? Their singular drive to survive is candidly revealed.

PLAYSAFE II: Don't Let It Happen To You

H. Clifford Chadderton
THE WAR AMPUTATIONS OF CANADA

The personal experiences of ten amputee children are revealed in documentary style in this sequel to PLAYSAFE. Filmed in a variety of rural and city locations across Canada, the Champs discuss the price they paid for a few seconds of fun playing in unsafe areas.

Fashioned after the show-and-tell approach popular in school classrooms, PLAYSAFE II probes even further into the events leading up to the children's accidents in the hopes that children who see the film will indeed recognize such circumstances in time to avoid a similar mishap. Each Champ takes viewers along to the actual locations where their accidents occured and demonstrate how their artificial limbs work.

DRESSING TECHNIQUES FOR THE SPINAL CORD INJURY PATIENT FROM THE WHEELCHAIR AND BED

MEDICAL CENTER REHABILITATION HOSPITAL

Media: 3/4" videotape, color Length: 19 min.
Audience: Spinal cord injured clients & their families, O.T.'s, nurses, health care educators, & students, home health care providers & other health care professionals.
Description: A spinal cord injured individual (paraplegic) demonstrates 2 methods of lower extremity dressing. The first method demonstrated is dressing from the bed. The second method is dressing while in an upright position while sitting in his wheelchair. The individual who demonstrates his dressing skills is a registered O.T., & accompanies his demonstration with a narrative.
Cost: Rental 1-Week (may be applied to purchase price)$40/Purchase:$250.
Produced, edited & photographed by Karen Bommersbach, Multimedia Coordinator & developed by the Office of Education & Professional Standards, Sharon Rask, Director, Medical Center Rehabilitation Hospital.

WHERE THERE'S A WHEEL THERE'S A WAY

MEDICAL CENTER REHABILITATION HOSPITAL
Box 8208 University Station
Grand Forks, North Kakota 58202

Media: 3/4" videotape, color Length: 18 min., 30 sec. Audience: Physically challenged individuals, their families, health care professionals, general public, athletic coaches, recreation departments & park boards.
Description: The program stresses the similarities between playing basketball from a wheelchair & stand-up basketball. It de-emphasized disabilities & addresses the misconception surrounding the physically challenged. The tape addresses the physical & psychological benefits obtained through participation in team sports. The emphasis is on what the physically challenged can do & encourages participation in wheelchair athletics. Sports wheelchair equipment adaptations are demonstrated.
Cost: Rental-1 Week (may be applied to purchase price)$45/Purchase:$250

Produced, edited, & photographed by Karen Bommersbach, Multimedia Coordinator & developed by the Office of Education & Professional Standards, Sharon Rask, Director, Medical Center Rehabilitation Hospital.

DOWNHILL; Any Way You Can

H. Clifford Chadderton
The War Amputations of Canada
2827 Riverside Drive
Ottawa, Ontario, K1V OC4

A group of child amputees conquer the mile-high Mont Tremblant in Quebec during a week-long ski clinic sponsored by The War Amps. Coaching the youngsters is double-leg amputee Karl Hilzinger, Sports Consultant for the Association and former professional ski instructor. Mr. Hilzinger is the only one in the world to ski on "stubbies", or short artificial legs.

From the first run down the beginners' slope for five Champs who had never skied before to the final run down the toughest slope of the mountain by five who have experience, the film is a documentary of accomplishments of each Champ, not measured against each other, but against each Champ's individual progress, determination and desire to succeed. DOWNHILL is an incentive film to encourage amputees to take up this challenging sport.

IT'S ABILITY THAT COUNTS

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

Sports and leisure activities for the handicapped. Introduced by Sir Ludwig Guttman, originator of sports for the handicapped and Stoke-Mandeville games. High-speed competitive basketball playing by paraplegics. Aquatics for paraplegics and amputees, obstacle courses and rope-climbing for cerebral-palsied children, volleyball game of the blind, wheelchair polo, blind equestrians, and numerous other activities are intercut with scenes from the 1974 Stoke-Mandeville Games. Film stresses the importance of athletics for the disabled and the case of adaptions that can open a variety of sports to them.
Review: Recommended for teacher/training ideas it offers professionals, encouragement it offers to the handicapped and those concerned with them. Film technique is generally good.

England, 16 mm., and U-matic, color, 32 mins., 1975. Producer and sales distributor: Cinexsa Film Productions. Rental from Rehabfilm: Members: $45.00; Non-members: $60.00.

RESNA EDUCATION COMMITTEE

Elaine Trefler, OTR and James Connelly, Ph.D
University of Tennessee, Memphis - REP
682 Court Avenue, Memphis, TN U.S.A. 38163

The field of Rehabilitation Engineering brings together a diverse group of individuals committed to designing, developing, evaluating and providing external and internal devices that will put the benefits of technology to work for disabled people. This videotape production produced by RESNA, an organization concerned with the exploitation of science and technology in the rehabilitative process, provides the viewer with an overview of Rehabilitation Engineering activities nationwide. Specific research projects and service related issues are presented with the visuals being provided by Rehabilitation Engineering professionals.

The target audience consists of consumers, rehabilitation professionals and concerned citizens.

THE TERRY FOX MOBILE CLINIC

H. Clifford Chadderton
THE WAR AMPUTATIONS OF CANADA
2827 Riverside Drive
Ottawa, Ontario, K1V OCr

Named after Terry Fox, the one-legged runner whose Marathon of Hope inspired amputees and disabled people, the Mobile Clinic is an innovative concept in health care funded by The War Amps and staffed by the Royal Ottawa Regional Rehabilitation Centre.

The clinic is a self-contained trailer and van that brings treatment therapy and rehabilitative education to disabled patients, their families and medical professionals in outlying communities in Eastern Ontario and Western Quebec.

The video documents the presentation ceremony of the clinic in January 1983 and describes its many functions. A prosthetist shows four-year-old Angela Craig the mini-workshop where artificial limbs are fashioned and fitted for individual patients.

REHABILITATION ENGINEERING "A CHALLENGE FOR ALL"

Terry Willkomm
Management Consulting & Rehabilitation Services
Ankeny, Iowa 50021

This presentation is an overview of what rehabilitation engineering is and how everyone can become involved.

Primary objective of the presentation is to show professionals and consumers that they do not need to be an engineer to solve unique problems.

Specific areas addressed include utilizing: community resource people, places and unique materials; approaches to solving unique problems; low cost, low technology devices (over 46 slides of specific examples); accessing commercially available devices and fabricating materials.

S.T.A.M.P. : THE TEAM APPROACH USING BIOENGINEERING TECHNOLOGY

M. J. Coen, M.S.; Dr. R. A. Baird; Dr. H. H. Kent
Special Team for Amputations Mobility Prosthetics
117-S Long Beach V.A. Med. Center 5901 E. 7th St.
Long Beach, CA U.S.A. 90822

Special teams have been established at eight Veterans Administration Medical Centers nationwide to provide a state-of-the-art, multidisciplinary team approach to the care and rehabilitation of problem cases involving amputation, impaired mobility, and/or other self-care functional limitations. The latest medical technology is incorporated into the clinical, educational, and research programs at each of these eight V.A. medical centers. A bi-directional interaction between these regional centers and outlying V.A. medical facilities has been established to optimize patient care and information/ education exchange.

This presentation will give viewers a look at these teams nationwide, the team as established at the Long Beach VAMC, and the role of biomedical technology in the pre-op evaluation through post-op rehabilitation in current veteran medical treatment.

Target Audience: People involved with amputation/ mobility problems especially in V.A. patient care. Availability: Eight V.A. medical centers nationwide with regional coverage and general interest.

Video Proceedings
ICRRT II

UND Engineering Experiment Station
Medical Center Rehabilitation Hospital
1300 S. Columbia Road
Grand Forks, North Dakota 58202

The video proceedings of the Second International Conference on Rural Rehabilitation Technologies (ICRRT II) focuses on the special technological needs of disabled people in rural areas. The video proceedings highlight both the unique challenges confronting those involved in rural rehab technology, as well as highlighting some of the resources currently available to the needs of the nearly 9 million Americans and the untold citizens of the world with disabilities living in rural settings. The four major themes of ICRRT II were: 1) rural hospital rehabilitation roles; 2) international rehabilitation technologies; 3) ag-rehab info networking; and 4) rural rehabilitation technology resources and directions. ICRRT II hosted authorities from around the world with 210 participants from 30 states with the foreign countries of Canada and Mexico both represented within the program.

VOICE INPUT AND VOICE OUTPUT

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

Documents the development of computers that talk and computers that listen. Two researchers at the Div. of Computer Research and Tech. at NIH developed voice output technology,using a VORTRAX speech synthesizing system, for use with computer CRT's. A blind programmer demonstrates his scanning system. Similarly, a voice input system is used by a high-level quadriplegic employee. One interesting segment involves the necessity of getting the computer used to an individual's voice - a problem which has plagued other voice input systems in the past.
Review: Clear and interesting review of this important research, for other researchers in this area, computer programmers, prospective employers, and users.

USA, U-matic, color, 22 mins.,1980. Producer and sales distributor: Division of Computer Research and Technology, National Institutes of Health. Rental from Rehabfilm: Members: $30.00; Non-members: $45.00.

PREVENTING PRESSURE SORES

Gillette Children's Hospital
200 E. University Avenue
St. Paul, MN 55101

Pressure sores are a health risk for children, adolescents, and adults with spina bifida or spinal cord injury. A special clinic is offered at Gillette Children's Hospital to prevent pressure sores for young people with spina bifida. The goal of the clinic is to reduce the incidence of pressure sores through education and proper seating. A team of physicians, therapists, nurse and rehab. engineer works with the client and family to provide proper seating and to motivate the young person to take responsibilty for his or her health.

Education in the causes of pressure sores and preventative self-care habits is a critical component of the Pressure Ulcer Prevention clinic. Excellent seating cannot prevent pressure sores without active client participation. This film, edited with permission from the original 3M Company producer, is one of several educational tools used to teach clients about the basic causes and prevention of pressure ulcers.

NOT JUST A SPECTATOR

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

Depicts extremely wide array of sports and leisure activity for the disabled. Includes swimming, sailing, kayaking, rock-climbing, spelunking, birdwatching, wheelchair dancing, skeetshooting, cross-country hiking, water skiing, bowling on the green, and horse-back riding. Training in technique and safety precautions emphasized. Many examples of the efforts of traditional recreation organizations extending their services to the handicapped.
Review: Vigorous, upbeat narrative, excellent photography. Has consistently received excellent reviews from those concerned with recreation activites for the handicapped. Has not been surpassed for variety.

England, 16 mm., color, 35 mins., 1974. Producer and sales distributor: Town & Country Production, Ltd. Rental from Rehabfilm: Members: $45.00; Non-members: $60.00.

BOLD

CRYSTAL PRODUCTIONS
P.O. Box 12317
Aspen, CO 81612

Active participation in all kinds of outdoor sports has brought enjoyment to blind people and a healthy way to spend their leisure time. This film portrays how BOLD, the Blind Outdoor Leisure Development Program, involves blind people in downhill skiing, cross country skiing, ice skating, horseback riding, jogging, golfing, river rafting, and mountain climbing. John Denver narrates the film and participates in the downhill skiing sequences. A film which would be of interest to the general public of all ages as well as school service clubs, church groups, health care workers and rehabilitation counselors.

VHS & BETA II #842 $89.50
3/4" U-MATIC #942 $134.50
16MM #142 $225.00
16MM Rental $25.00 (3 Days)

NOT QUITE THE CANADIAN OPEN

H. Clifford Chadderton
THE WAR AMPUTATIONS OF CANADA
2827 Riverside Drive
Ottawa, Ontario, K1V OC4

The War Amputations of Canada produced a videotaped golf tournament featuring members of the CHAMP Program. Young amputees are encouraged to join the WINNER'S CIRCLE by developing their skills in recreational activities.

Sports Consultant Karl Hilzinger and Chief Executive Officer H. Clifford Chadderton provide play-by-play colour commentary. The performance by the young Champs is an inspiration not only to amputees, but to anyone interested in children, the disabled and the game of golf.

NO SIMPLE ROAD

CRYSTAL PRODUCTIONS
P.O. Box 12317
Aspen, CO 81612

The excitement and joy of handicapped children participating in outdoor sports and activities is reflected sensitively in this award-winning film produced by Children's Hospital. Children with muscular dystrophy, polio, cerebral palsy, spinal bifida, and other physical handicaps learn to play tennis, golf, ski, and experience the pleasures of rafting, fishing, hayrides, and mountain excursions. These experiences allow youngsters to better accept their disabilities, gain self-confidence, and react positively to their able-bodied peers. AWARDS: IFPA GOLD CINDY.

VHS & BETA II #848 $89.50
3/4" U-MATIC #948 $134.50
16MM #148 $235.00
16MM Rental $35.00 (3 Days)

OPEN BOATING

REHABFILM
Rehabilitation International USA
1123 Broadway Suite 704
New York, New York 10010

The film presents a recreational boating program designed for physically handicapped persons. Beginning with water familiarity activities, individuals with a wide range of physical disabilities are shown progressing through various stages in their acquisition of boating skills. Boating activities include sailing, rowing and canoeing. The film shows the teaching model used, equipment, adaptations and facility modifications. Review: Interestingly shot with good narration this is an excellent film for anyone interested in boating for the physically disabled, but especially useful for parks and recreation administrators.

USA, 16 mm, color, 30 mins., 1982. Producer: Richard Olcese and Karen Southerland. Sales distributor: City of Oakland Office of Parks and Recreation. Rental from Rehabfilm: Members: $45.00; Non--members: $60.00.

WILDERNESS INQUIRY

Wilderness Inquiry II/Ocatvio Molina
1313 Fifth Street S.E., Suite 327A
Minneapolis, MN 55414

An award winning 28-minute documentary about a 10 day wilderness canoe trip involving physically disabled people. Filmed on location in northwestern Ontario, this film highlights the methods in which persons with cerebral palsy, paraplegia, and blindness overcome the challenges of a true wilderness adventure. These adventures are organized by Wilderness Inquiry II,a non-profit Minneapolis based organization. This film has been shown on dozens of PBS stations across the United States and Canada.

For purchase information, contact the office of Wilderness Inquiry II at 612-379-3858, Voice or TTY.

TWO, THREE, FASTEN YOUR SKI

CRYSTAL PRODUCTIONS
P.O. Box 12317
Aspen, CO 81612

This award-winning film about a hospital rehabilitation program for amputee skiers portrays the fun and excitement from the skier's point of view. As one of the skiers says, "They shouldn't admire us because we ski with one leg. We like to think of three-track skiing as a separate and unique sport." Another says, "I always used to say that I can't, even before I tried things: but now I'm finding out that I can do things like that."

VHS & BETA II	#844	$89.50
3/4" U-MATIC	#944	$134.50
16MM	#144	$250.00

16MM Rental $30.00 (3 Days)

CONSEQUENCES

CRYSTAL PRODUCTIONS
P.O. Box 12317
Aspen, CO 81612

The exhilaration, excitement, and beauty of adventure sports are contrasted with the potential consequences - accidents, spinal cord injury and permanent paralysis. Breathtaking scenes of hang-gliding, surfing, diving, auto racing, climbing, skiing, and other sports are intercut with comments from spinal cord victims. They have experienced the thrill of these sports and discuss why they were injured and the changes in their lives since they have been in a wheelchair. Ideal for health and safety programs and for general awareness about spinal cord injury and the resulting paralysis. AWARDS: GOLD AWARD, Miami Film Festival; THE RED RIBBON AWARD, American Film Festival; CINE, Golden Eagle. Captioned film for the deaf also available.

VHS OR BETA II	#841	$69.50
3/4" U-MATIC	#941	$99.50
16MM	#141	$150.00

16MM Rental $20.00 (3 Days)

CHALLENGED EQUESTRIANS
The Use of the Horse for Special Populations

Barbara L. Glasow, PT
Winslow Therapeutic Riding Unlimited, Inc.
340A South Rt. 94
Warwick, N.Y. 10990

CHALLENGED EQUESTRIANS is a videotape that presents divisions of riding for the disabled from a particular perspective. Medical, educational and sport riding programs for the disabled can all be carried out in a therapeutically sound way with the main goal to provide the challenged individual the opportunity to interact with horses in ways that will facilitate his personal growth. The approach used by Winslow Therapeutic Riding Unlimited, Inc. of Warwick, NY, in its riding program is one which follows the natural learning process of a child. The progression may take a rider through hippotherapy, developmental vaulting, remedial vaulting, remedial riding or sport riding depending on the level of maturation and needs of the rider. The rationale of each stage of the progression is presented with a look at the professionals involved, the benefits of each stage and the different therapeutic techniques used.

The videotape is of interest to: riding instructors, therapists, educators and rehabilitation specialists. It is available for purchase from Winslow Therapeutic Riding Unlimited, Inc. for $69.95.

THE MOUNTAIN DOES IT FOR ME

CRYSTAL PRODUCTIONS
P.O. Box 12317
Aspen, Co 81612

Telling the story of teaching youngsters with cerebral palsy to ski, the film looks at both the special training and educational approach in the hospital. One of the skiers in the movie describes how he feels about skiing. "It's kind of like a dream because movement, like walking, doesn't come easily for me. But in skiing, I don't have to fight it. The mountain does it for me." AWARDS: American Film Festival BLUE RIBBON; International Film and TV Festival of New York BRONZE METAL; Samosonite International Ski Film Festival SPECIAL JURY MEDAL; Rehabilitation Film Festival FIRST PRIZE.

VHS & BETA II	#843	$89.50
3/4" U-MATIC	#943	$134.50
16MM	#143	$180.00

16MM Rental $25.00 (3 Days)

WHAT'S YOUR HANDICAP?

H. Clifford Chadderton
THE WAR AMPUTATIONS OF CANADA
2827 Riverside Drive
Ottawa, Ontario, K1V OC4

This is a story of a young amputee who overcomes his handicap and masters the sport of golf. Doug Paulsen of Manitoba, wears an artificial leg. At first, he fears it will limit his ability to learn the game. He soon proves himself wrong, with the help of some very special people.

Doug meets former CFL All-Star Karl Hilzinger, who lost both legs and partial use of his right arm, providing Doug with the incentive he needs to get over the psychological hurdle. Doug also receives some helpful tips from Canadian Golf Professional, Geroge Knudson.

"WHAT'S YOUR HANDICAP?" encourages amputees and disabled people to take up golf as a successful form of rehabilitation.

AMPUTEE GOLF: IT'S THE SHOT THAT COUNTS

H. Clifford Chadderton
THE WAR AMPUTATIONS OF CANADA
2827 Riverside Drive
Ottawa, Ontario, K1V OC4

Amputees who have mastered golf encourage all disabled people to play in this step-by-step approach to the sport. Endorsed by Canadian Tour Professional Jim Nelford, the show highlights the physical and psychological aspects of the game. Various golf shots are demonstrated,including the drive, chip shot, bunker shot and the put.

The host, War Amps' Chief Executive Officer Cliff Chadderton, explains how the disabled can learn the game and the special concepts amputees have developed to enjoy the sport.

The unique adaptive equipment The War Amps has developed to help amputees play golf is also demonstrated.

MOUNTAIN SILENCE

CRYSTAL PRODUCTIONS
P.O. Box 12317
Aspen, CO 81612

The film shows hearing-impaired young people participating in many outdoor summer and winter activities at Mountain Silence, a camp school for the deaf in the Rocky Mountains. Here, the children gain the confidence they need to become successful adults and find that the mountains themselves are great teachers for self-reliance. Skiing, rafting, horseback riding, dance, mime, crafts, backpacking, and mountain climbing are among the many activities available at the camp school where reinforcement of sign languages is also emphasized.

VHS & BETA II	#847	$89.50
3/4" U-MATIC	#947	$134.50
16MM	#147	$250.00

16MM Rental $35.00 (3 Days)

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