MEC '02

The Next Generation

University of New Brunswick's
MyoElectric Controls/Powered Prosthetics
Symposium

August 21-23, 2002

Fredericton NB CANADA

CONFERENCE PROCEEDINGS

Hosted by:
Institute of Biomedical Engineering

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Myoelectric Controls / Powered Prosthetics Symposium
Fredericton, NB Canada
August 21-23, 2002

Welcome from the MEC '02 Organizing Committee

Welcome to MEC '02. This three day international symposium and the related displays and courses promise to offer something for everyone in the powered upper limb prosthetics field.

The symposium features two keynote speakers and five invited speakers in addition to the scientific papers and commercial exhibits and presentations. The keynote speakers are Dr. **Robert N. Scott**, Professor Emeritus, Electrical and Computer Engineering at the University of New Brunswick, and Mr. **Steve Hughes**, Research Director at Queen Mary's Hospital in London, and Director, Biomedical Engineering at University of Surrey, in the UK.

Dr. Scott will talk about the development of the myoelectric controls field and Mr. Hughes will talk about his involvement in developing osseointegrated femoral implants to allow percutaneous, direct skeletal attachment of prosthetic limbs.

The program offers ample free time to meet other people who are attending and to visit the exhibits. We hope to see everyone at the Old Government House for the Wine and Cheese reception on Wednesday, August 21st, from 7 to 10 pm. The Old Government House is a Fredericton landmark and National Historic site located next to the Sheraton Fredericton Hotel.

We hope you enjoy MEC '02 and your visit to Fredericton. Please don't hesitate to ask if we help in any way during your stay.

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Wednesday, August 21, 2002

8:45 am	Welcome to MEC '02
9:00 am	**Keynote Address: Dr. Robert N. Scott** Professor Emeritus in Electrical Engineering, University of New Brunswick
10:00 am	Refreshment Break / Vendor Displays
10:30 am	**Theme: Prosthetics and Hardware** *Design of a clinically viable multifunctional prosthetic hand* *A myoelectrically controlled prosthetic hand for transmetacarpal amputations* **Invited Speaker / Session Chair:** Richard F. Weir
11:00 am	*Evaluation of a new flexion wrist integrated with electric hand* H. H. Sears *Optimal fixed wrist alignment for below-elbow, powered, prosthetic hands* J. S. Landry *Locking shoulder joint enhancements improve functionality* T. W. Williams III *Hybrid approach to bilateral UE prosthetic Rehabilitation* T. Farnsworth
12:00 pm	Lunch Break
1:30 pm	**Theme: Prosthetics** *Enhancing function: the glimcher test at its best (hybridization of servo and myoelectric control for the transhumeral level limb deficient individual)* *Experiences with the Otto Bock ergo elbow* **Invited Speaker / Session Chair:** John M. Miguelez
2:00 pm	*Solutions for ergonomics in high level amputation treatment* C. R. Hell *Advancement of upper extremity prosthetic interface and frame design* R. D. Alley *Results of a study with the goal to optimize harnesses by means of up-to-date biomechanical engineering* T. Bertels

	Progress towards a biomimetic prosthetic arm D. L. Russell
3:00 pm	Refreshment Break / Vendor Displays
3:30 pm	*EMG electrodes for myoprosthetics : design principals and application tips* G. Haslinger
	Electrodes installed in roll-on suspension sleeves W. K. Daly
	Custom silicone liners for upper extremity prosthetics I. Whatmough
	Experience with silicone suction sockets using myoelectric control J. E. Uellendahl
4:30 pm	Shuttle bus from UNB to the Sheraton Hotel
7:00 -10:00 pm	**Wine and Cheese reception at Old Government House**

Thursday, August 22, 2002

8:45 am	Daily Notices
9:00 am	**Keynote Address: Mr. Steve Hughes** Research Director at Queen Mary's Hospital in London, and Director, Biomedical Engineering at University of Surrey, in the UK
10:00 am	Refreshment Break / Vendor Displays
10:30 am	**Theme: Occupational Therapy and Case Studies** *The influence of concomitant diagnoses when treating the older UE amputee* **Invited Speaker / Session Chair:** Margaret F. Wise
11:00 am	*Prosthetic Management of an individual with "unique" multi-level limb deficiencies: a case study* J. W. Limehouse
	Elective amputation of Cerebral Palsy patient successfully wears electric prosthesis S. A. Mandacina

	Electronic prosthesis for the partial hand amputee J. M. Miguelez
	Clinical trials of the new Boston digital arm system T. Farnsworth
	Fitting of transcarpal myoelectric prosthesis with locking liner R. Dakpa
12:00 pm	Lunch Break
1:30 pm	**Theme: Hardware and Controllers** *Prosthetic actuation: a case for pneumatics* *Prosthetic control: a case for extended physiological proprioception* **Invited Speaker / Session Chair:** Dick H. Plettenburg
2:00 pm	*Real time personal computer (pc) modeling of a prosthesis controller based on the concept of extended physiological proprioception (epp)* T. R. Farrell
	Unique device-selection strategies for powered elbows C. Wallace
	Clinical experiences with animated prosthetics controller and LIMB LINK ™ J. W. Limehouse
	Mini joystick for upper limbs prostheses A. Davalli
3:00 pm	Refreshment Break / Vendor Displays
3:30 pm	*Microprocessor control features and control options* G. Sjonnesen
	New prosthetic controller expands capabilities W. J. Hanson
	Software/firmware tools to customize controller parameters in upper extremity, powered prosthetic systems D. Wells
	Comparative analysis of microprocessors in upper limb prosthetics B. Lake
4:30 pm	Shuttle bus from UNB to the Sheraton Hotel

7:00 pm	**Public Lecture – "The Branemark Approach to Osseointegration"** Mr. Steve Hughes *Wine and Cheese Reception and Tour of the recently expanded Institute of Biomedical Engineering facility on the University of New Brunswick Campus will follow the presentation.*

Friday, August 23, 2002

8:45 am	Daily Notices
9:00 am	**Theme: Research** *A study of EMG signals from limbs with congenital absence and acquired losses* **Invited Speaker / Session Chair:** P. J. Kyberd
9:30 am	*Experience with the intelligent hybrid arm systems* A. S. Poulton
	Distributed forelimb pressures for prosthetic control S. L. Phillips
	Control and signal processing concepts for a multifunctional hand prosthesis M. Reischl
	Myoelectric prosthesis with sensorial feedback A. Rios Poveda
	Neuro-fuzzy logic as a control algorithm for an externally powered multifunctional hand prosthesis A. B. Ajiboye
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12:00 pm	Closing Comments
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A myoelectrically controlled prosthetic hand for transmetacarpal amputations

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Harold Sears
Evaluation of a new flexion wrist integrated with electric hand

11:15 – 11:30
John Landry
Optimal fixed wrist alignment for below-elbow, powered, prosthetic hands

11:30 – 11:45
T. Walley Williams III
Locking shoulder joint enhancements improve functionality

11:45 – 12:00
Troy Farnsworth
Hybrid approach to bilateral UE prosthetic Rehabilitation

DESIGN OF A CLINICALLY VIABLE MULTIFUNCTIONAL PROSTHETIC HAND

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INTRODUCTION

We are in the process developing a new multifunctional hand mechanism in the hopes of providing a new mechanism that will have superior function over today's single degree-of-freedom (DOF) mechanisms and yet be clinically viable. However, this is no easy task. There have been a multitude of multifunctional hands built, all of which have failed to find clinical application as an artificial hand replacement. During the 60s and 70s much time, effort, and money was invested in the development of externally-powered multi-functional hand-arm systems. Prime among these being the Edinburgh Arm [1], the Boston Arm [2,3], the Philadelphia Arm [4,5], the Belgrade hand [6,7], the Sven Hand [8], the Utah Arm [9]. However while many of today's commercially available externally-powered *elbow* systems (Boston Elbow – Liberating Technology Inc [10], NYU Elbow – Hosmer-Dorrance, and Utah Elbow – Motion Control [11]) owe their origins to this era of upper-limb research no multifunctional hand mechanisms made the transition from the Laboratory into clinical practice.

The Sven hand never found widespread clinical use, even though it was extensively used in multi-function control research using pattern recognition of myoelectric signals [12]. Henry Lymark, the director of the Handikappinstitutet of Stockholm, Sweden, later created a simplified version of the Sven hand called the ES hand. This hand was an attempt to produce a more robust and hence more clinically viable version of the Sven hand. Unfortunately, Lymark died soon after the initial development of this hand and the project was never continued. The Philadelphia Arm [4,5] also never found clinical use, but like the Sven hand found use as a research tool for multifunction control using weighted filters for the pattern recognition problem. The Belgrade hand too was never used clinically but has ended up in the robotics field in the form of the Belgrade/USC robotic hand [13].

More recent multifunctional hand mechanisms are the Montreal hand [14], the Southampton Hand [15], The Rutgers Hand [16], and the Karlsruhe Research Center humanoid hand [17]. The Montreal/Lozac'h's hand is a single DOF hand, i.e. opening & closing, but because the hand has articulated fingers that move independently of each other it is capable of forming an adaptive grip with which to grasp objects. An adaptive grip requires much lower forces than conventional prosthetic hands to hold objects. Also, *operating under the assumption that a practical prosthetic hand can have only one axis of rotation for the thumb, Lozac'h [18] performed a series of experiments and determined that the preferred working plane of the thumb lay between 45 and 55 degrees*. The Montreal hand looked very impressive when demonstrated but the articulated fingers may be too fragile to withstand the rigors of daily life.

The Southampton Hand [15] also has articulated digits enabling an adaptive grasp. But this hand uses a system of "hierarchical artificial reflexes" to automate the control process. In this hand the operator is taken "out of the loop" and onboard processing is used with sensors in the hand to tell the hand what grasp pattern to adopt. The user only provides a conventional single DOF open/close EMG signal. The idea is that by allowing the processor to take control it reduces the mental loading on the user. *A major factor in the success or failure of this type of control is user confidence in the mechanism allowing the user to relinquish control to the artificial reflexes*. The Southampton hand is a work in progress that has yet to gain clinical acceptance, but has found some application as an end effector for the MANUS medical manipulator.

The Rutgers multifunctional hand [16] has gained considerable media attention. This hand uses tendon movement to actuate pneumatic sensors. These sensors are interposed between a prosthetic socket and superficial extrinsic tendons associated with individual finger flexors. The control scheme is interesting and demonstrates the feasibility of multifunctional control. However the idea of using pneumatic sensors is not in itself new, only the way the pneumatic sensors are implemented is new. In the past pneumatic or pressure sensors were tried and abandoned due to problems discriminating between user commands and external disturbances caused by interaction with the real world.

The humanoid hand developed at the Karlsruhe Research Center is a very recent multifunctional hand that is driven by what are claimed to be a new class of pneumatic actuator [17]. The actuators consist of cavities that change their size when inflated with a pressurized gas or liquid. The five-fingered hand resembles the functions and anatomy of the human hand. All fingers and the wrist can be positioned independently. The compact size of the actuators allows for their full integration into the finger. A safe grasp is accomplished through the flexible self-adaptability of the actuators. The maximized contact surface allows for the use of relatively low pressures. It remains to be seen whether this hand will be clinically viable. Robustness and how it is to be controlled remain issues for concern.

In the end, most multifunctional hand designs are doomed by practicality, even before the control interface becomes an issue. Prosthesis users are not gentle with their devices; they expect them to work in all sorts of situations never dreamed of by their designers. Most mechanisms fail because of poor durability, lack of performance and complicated control. *No device will be clinically successful if it breaks down frequently.* A multifunctional design is by its nature more complex than a single DOF counterpart. Articulated joints on fingers are more likely to fail than monocoque, or solid finger, designs. To be clinically viable, the fingers and thumb of most prosthetic hands are non-articulated and have a single axis of rotation. This minimizes the number of moving parts, reduces complexity and increases robustness. Palmar prehension is achieved by single joint fingers that are fixed in slight flexion at a position approximating the interphalangeal joint. The resulting finger shape also creates a concave inner prehension surface that can be used to provide cylindrical prehension [19]. Another practical consideration is performance. The hand must be able to generate enough torque and speed, and have a sufficient width-of-opening to be useful to the user. *The pinch force of a multifunctional hand does not have to be as high as that of current commercially available single DOF hands because of the adaptive nature of their grip.* But they should still be capable of high speeds-of-opening and have a pinch force of at least 68 N (15 lbs_f) in accordance with Peizer et al [20].

However, in spite of these issues we believe a compromise must be reached if increased function is to be achieved. Some of the robustness and simplicity of single DOF devices must be traded to achieve the increase in performance possible with a multi DOF hand. This leads us to the question:

Is There an Optimal Number of Degrees-of-Freedom for Multifunctional Hands?

A possible compromise to the dilemma of robustness versus increased function is to limit the device to those degrees-of-freedom necessary to replicate Keller et al.'s [21] grasp patterns. This idea of providing sufficient DOFs to recreate Keller et al.'s grasp patterns turns up in many unrelated fields. Professional SCUBA diver gloves trade function for warmth in order to extend dive times. A mitten is warmest while a glove with individual fingers is the most functional. Professional SCUBA diver gloves are a compromise having the thumb and index fingers free and the middle, ring, & little fingers together. This configuration affords the diver the basic prehension patterns of the hand while at the same time keeping the bulk of the hand warm.

In the area of remote manipulation, the SARCOS system [22] uses a three DOF hand for the slave manipulator terminal device and limits the hand of the operator to the same three DOF when in the master arm. In this mechanism the thumb has two DOFs while the three fingers (middle, ring & little) have the third DOF. The index finger is rigid. For space suit gloves the Direct-Link Prehensor [23,24] limits the motions of the operator's hand to three DOF. Space suit gloves are bulky and stiff due to the suit's pressurization. This stiffness results in limited external dexterity. Also, as in the case with diver's gloves, tactile sensation and manual dexterity are lost because the hand is gloved. The Direct-Link Prehensor is a two finger device with the thumb mounted at 45° to the fingers.

In the area of surgery, Beasley [25] described a surgical procedure to provide a functional four DOF hand for persons with C5-C6 quadriplegia. This procedure makes possible precision prehension with careful positioning of the stabilized thumb to oppose the actively flexed index and middle fingers. The result is a functional hand that retains some of its sense of touch. For functional electrical stimulation (FES), surgical techniques have been combined with implantable four channel FES electrodes to enable patients with flail arms to reproduce the palmar and lateral grasp patterns of Keller et al. [26].

In each of these situations sensation (feedback) was compromised [by gloves or remote nature of the terminal device] or muscular function (control) was impaired. In all cases enabling the hand to recreate Keller et al.'s prehension patterns optimized hand function vs. the number of available control sources. Therefore it would appear that an artificial hand capable of reproducing these grasp patterns would require three or four DOFs: one, more usually two, for the thumb; one for the index finger; and one for the middle, ring and little (MRL) fingers which are combined to move as a unit. Limiting the thumb to a single DOF and orienting it such that it operates along its preferred plane – 45° reduces the number of DOFs to be controlled to three.

Furthermore, if one considers the role a single DOF thumb operating along the 45° plane, it becomes apparent that the final prehension pattern adopted by a hand is a function thumb position control. i.e. if the thumb engages both the index finger and MRL finger unit, then tri-digital pinch (three jaw chuck) results. If the thumb only engages the index finger, tip prehension results. If the thumb closes on the side of the index finger, lateral prehension results. Power grasps result from digit shape and a wide width-of-opening. Thus it is possible through appropriate control of the thumb to determine the resulting prehension pattern of the hand.

A system of this kind (where the timing, or speed, of thumb is controlled) can be configured so a single "open" signal drives all digits (fingers & thumb) back to their start positions and two "close" signals, one for the index and MRL finger drives and a second for the thumb drive, control hand closure. This implies a one and half DOF control system. *It should be noted that prehension (grasping) should not be confused with manipulation. For dexterous manipulation many more degrees-of-freedom of control are required.*

IN CONCLUSION

Using these ideas we are building a multifunctional hand mechanism that we believe will be robust and simple enough to be clinically viable. This mechanism is based on a three motor hand with a two motor wrist. We are incorporating a wrist because the wrist is used so extensively in the positioning of the hand in space. One motor will drive a single DOF thumb that will operate along the preferred 45° plane; one motor will drive the index finger; one motor will drive the middle, ring and little fingers as a unit; one motor will provide wrist extension/flexion and one motor will provide wrist rotation. The drive trains for these motors will be based on the drive train developed for our partial hand mechanism [27]. The digits will be solid non-articulated designs to improve robustness, although some articulated fingers might be tested just for fun. This mechanism is to be controlled using a 4 DOF fuzzy logic based myoelectric controller we are also in the process of developing [28].

ACKNOWLEDGMENT

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REFERENCES

1. Simpson, D.C., "An Externally Powered Prosthesis for the Complete Arm." Biomed Eng, 4(3) 106-110,119, 1969.
2. Mann, R.W., "Efferent and Afferent Control of an Electromyographic Proportional Rate, Force Sensing Artificial Elbow with Cutaneous Display of Joint Angle." Proc Symp on the Basic Problems of Prehension, Movement, and Control of Artificial Limbs, Institution of Mechanical Engineers, London, pp. 86-92, 1968.
3. Mann, RW Reimers SD "Kinesthetic Sensing for the EMG Controlled Boston Arm." MMS-11(1) 110-115, 1970
4. Taylor, D.R., & Wirta, R.W., "Development of a Myoelectrically Controlled Prosthetic Arm." Proc 3rd Intn'l Symp on External Control of Human Extremities, Dubrovnik, Yugoslavia, Aug 25-30, 1969. In: Advances in External Control on Human Extremities, Yugoslav Committee for Electronics and Automation (ETAN), 1970.
5. Taylor, D.R., & Finley, F. R., "Multiple-Axis Prosthesis Control By Muscle Synergies." In: The Control Of Upper-Extremity Prostheses and Orthoses, Herberts, P., Kadefors, R.,

Magnusson, R.I., & Petersén, I., (Eds), Charles C. Thomas, Springfield, Illinois, pp. 181-189, 1974.
6. Razic, D., "Kinematics Design of a Multifunctional Hand Prosthesis." Proc 4th Intn'l Symp on External Control of Human Extremities, Dubrovnik, Yugoslavia, August 28-Sept 2, 1972. In: Advances in External Control of Human Extremities, M.M. Gavrilovic & A.B. Wilson, (Eds), Yugoslavia Belgrade ETAN, pp177–183, 1973.
7. Stojiljkovic, Z.V., & Saletic, D. Z., "Tactile Pattern Recognition by Belgrade Hand Prosthesis." Proc 5th Intn'l Symp on External Control of Human Extremities, Dubrovnik, Yugoslavia, Aug, 1974. In: Advances in External Control of Human Extremities, ETAN, 1975.
8. Herberts, P., Almström, C., & Caine, K., "Clinical Application study of multifunctional Prosthetic Hands." J Bone & Joint Surgery, No. 60, pp. 552, 1978.
9. Jacobsen SC Knutti, DF. Johnson RT & Sears HH "Development of the Utah Arm." BME-29(4) 249-269, 1982.
10. Williams, T.W. "Use of the Boston Elbow for High-Level Amputees," Chapter 19. Comprehensive Management of the Upper-Limb Amputee, Atkins DJ & Meier RH (Eds), Springer-Verlag, NY, pp211-220, 1989.
11. Sears, H.H., Andrew, J.T., & Jacobsen, S.C., Experience with the Utah Arm, Hand, and Terminal Device. Comprehensive Management of the Upper-Limb Amputee, Atkins, D.J., & Meier, R.H., (Eds), Springer-Verlag, New York, pp194-210, 1989.
12. Lawrence, P.D., & Kadefors, R., "Classification of Myoelectric Patterns for the Control of a Prosthesis." In: The Control Of Upper-Extremity Prostheses and Orthoses, Herberts, P., Kadefors, R., Magnusson, R.I., & Petersén, I., (Eds), Charles C. Thomas, Springfield, Illinois, 1974.
13. Beattie, D., Iberall, T., Sukhatme, G.S., & Bekey, G.A., "EMG Control for a Robot Hand Used as a Prosthesis." Proc. 4th International Conf Rehab. Robotics (ICORR), Wilmington, Delaware, Bacon, D.C., Rahmin, T., & Harwin, W.S., (Eds), Applied Science & Engineering Laboratories, University of Delaware, A.I. duPont Institute, Wilmington, Delaware, pp67-72, June, 14-16, 1994.
14. Vinet, R., Lozac'h, Y., Beaudry, N., & Drouin, G., "Design Methodology for a Multifunctional Hand Prosthesis." JRR&D, 32(4) 316–324, Nov, 1995.
15. Kyberd PJ & Chappell PH "The Southampton Hand: An Intelligent Myoelectric Prosthesis." JRR&D, 31(4) 326-334, Nov, 1994.
16. Abboudi, R.L., Glass, C.A., Newby, N.A., Flint, J.A., & Craelius, W., "A Biomimetic Controller for a Multifinger Prosthesis." IEEE Trans Rehab Eng, 4(2) 121-129, June, 1999.
17. Schulz, S., Pylatiuk, C., & Reischl, M., "A Lightweight Multifunctional Prosthetic Hand." Submitted to Robotica, Special Review: Advance in External Prosthetic Limbs at the Start of the 21st Century, 2002.
18. Lozac'h, Y., "The Preferred Working Plane for an Active Thumb." Proc. 2nd Conf Rehab Eng (RESNA 84), Ottawa, Canada, RESNA Press, Washington DC, pp24-25, June 17–22, 1984.
19. Heckathorne, C.W., "Components for Adult Externally Powered Systems." Chapter 6C. In: Atlas of Limb Prosthetics, 2nd Ed, Bowker, J.H. & Michael, J.W., (Eds), Mosby-Year Book, St. Louis, MO, pp151-175, 1992.
20. Peizer et al., 1969.
21. Keller, A.D., Taylor, C.L., & Zahn, V., "Studies to Determine the Functional Requirements for Hand and Arm Prostheses." Department of Engineering, University of California at Los Angeles, 1947.
22. Jacobsen, S.C., Smith, F.M., Iversen, E.K., & Backman, D.K. "High Performance, High Dexterity, Force Reflective Teleoperator." Proceedings of the 38th Conference on Remote Systems Technology, American Nuclear Society (ANS), La Grange Park, Illinois, v2, pp. 180-185, 1990.
23. Direct-Link Prehensor, NASA Tech Briefs, Vol. 15, No. 12, pp. 78, December, 1991.
24. Direct-Link Prehensor, Technical Support Package, NASA Tech Briefs, ARC-11666, NASA, Ames Research Center, Moffett Field, CA 94035, 1991.
25. Beasley, R.W., "Surgical Treatment of Hands for C5-C6 Tetraplegia." Symp. on Rehab After Reconstructive Hand Surgery, 14(4) 893-904, Oct, 1983.
26. Triolo, R., Nathan, R., Yasunobu, H., Keith, M., Betz, R.R., Carroll, S., & Kantor, C., "Challenges to Clinical Deployment of Upper Limb Neuroprostheses." JRR&D 33(2) 111-122, April. 1996.
27. Weir, R.F.*ff*., & Grahn, E.G., "A Myoelectrically Controlled Prosthetic Hand for Transmetacarpal Amputations." Proc MEC2002, Fredericton, New Brunswick, Canada, Aug 19-23, 2002.
28. Ajiboye AB Weir RF*ff* Heckathorne CW Childress DS "Neurofuzzy Logic as a Control Algorithm for an Externally Powered Multifunctional Hand Prosthesis" Proc MEC2002, New Brunswick Aug 19-23, 2002.

A MYOELECTRICALLY CONTROLLED PROSTHETIC HAND FOR TRANSMETACARPAL AMPUTATIONS

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INTRODUCTION

We have developed a new externally-powered, myoelectrically controlled partial hand prosthesis that is suitable for fitting those persons with amputations at or more proximal to the level of the metacarpophalangeal (MCP) joint. This hand mechanism is capable of reasonable pinch forces (12 lbs_f) and fast opening and closing (2 rads/sec). In a partial hand mechanism there is very little space available for the drive mechanism if an aesthetic result is to be achieved and any residual motion of the wrist is to be preserved. The challenge is to be able to fit all the requisite mechanisms and electronics in the highly confined volume that remains after accommodating the residual limb and still have reasonable performance.

CURRENT STATUS OF RESEARCH IN THE AREA

Prosthetic solutions for partial hand amputations are considered only after surgical procedures have failed. With an intact thumb, even if the metacarpals have been somewhat shortened with an oblique amputation, an orthotic device can be fitted that will provide a post for the uninjured thumb to oppose. Alternatively a static cosmetic prosthesis can be fitted. A review of the current surgical and prosthetics practice for partial hand amputations can be found in the *Atlas of Limb Prosthetics* [1, 2]. In general, only those cases where all digits (thumb and all four fingers) have been lost at a level equal or proximal to the metacarpophalangeal joint is a functional hand prosthesis recommended. Functional partial hand prostheses may be body-powered or externally-powered.

Body-powered prostheses for persons with partial-hand amputations fall, for the most part, into one of two groups, those devices that are powered by biscapular abduction and/or gleno-humeral flexion, and those devices that are powered by flexion or extension of the wrist. A shoulder-driven device is often the system of choice for the persons with bilateral partial-hand amputations. The control mechanism consists of a figure-of-nine harness that fits about the shoulders and a cable that runs from the harness to an appropriate terminal device. The most usual terminal devices take the form of a hook. The disadvantages are poor cosmesis and the requirement for a harness and control cable to be worn; however, these devices do preserve the residual motion of the wrist and provide feedback to the user about the state of their prosthesis.

One such device used a hook attached to the palmar surface of a partial hand prosthesis socket. The attachment was made through the use of a "Handy Wrist" that used to be available from USMC (United States Manufacturing Corp.). The combination of hook and wrist became known as the "Handy Hook". This was one of the most functional prosthetic fittings for this level of amputation because wrist motion was preserved to assist the prehension process. Another system that attempted to provide both function and cosmesis for body-powered partial hand fittings was the Robin Aids Hand. The Robin Aids Hand consisted of mechanical fingers that had interchangeable components that were mounted on a very short frame. The Robin Aids hand remained the device of choice in the fitting of partial hand amputations for many rehabilitation centers until it was discontinued in 1990.

The second type of body-powered partial-hand prostheses is the wrist flexion and extension device. These devices are functional cosmetic hands that operate in a manner similar to a tenodesis type of hand orthosis ("Tenodesis action" is a method by which prehension of the index finger, middle finger, and thumb is achieved through active wrist extension). Essentially these devices operate by way of a linking mechanism that translates wrist flexion into finger pinch and wrist extension into finger opening. The user can tell where the fingers are and the forces involved through the wrist. This method is used in Europe to fit persons with partial hand amputations. The shortcomings of this type of device are that any remaining motion of the wrist that would normally be used to position a prehensor in space must be sacrificed to drive the opening and closing of the mechanism. Also while these devices are in general statically aesthetically pleasing, their operation can result unnatural appearing motions.

Externally-powered prosthetic devices are few and far between. Only Otto Bock, GmbH., Germany, has a commercially available partial hand mechanism. This device is suitable for persons with wrist disarticulations and short trans-carpal hand amputations. The majority of the other commercially available mechanical hands for persons with trans-radial and trans-humeral amputations are not suitable for persons with wrist disarticulations or partial-hand amputations because the resulting prostheses are too long. A couple of exceptions are the powered hand from Centri System, Sweden, which can be shortened by a creative prosthetist to enable its use in some short trans-carpal cases and Motion Control, Salt Lake City, UT, claims that its new hand can also be shortened for use in short trans-carpal cases. Of the commercially available mechanisms none is short/compact enough to be used in transmetacarpal cases.

In the area of ongoing research, Gow et al. [3] in Scotland developed an externally-powered device based on a wrist flexion/extension type prosthesis. This system evolved into the REACH hand [4] and more recently into a small externally powered partial hand prosthesis, named ProDigits [5]. A custom made cosmetic silicone glove is worn over the mechanism, giving the hand an aesthetically pleasing result. This main limitation of the device is the limited overall pinch force and speed. The actuators can generate gripping forces of about 20 N (4.5 lbs_f), for adult-sized fingers, at speeds of around 1 rad/sec. RSLSteeper, England, has conducted limited clinical trials of the ProDigits system but as yet it is not commercially available.

Another externally powered device for persons with partial hand amputations was developed at the University of New Brunswick, Canada [6]. This mechanism used a single transverse motor in the line of the knuckles to open and close finger armatures. This system had a range of finger motion of approximately 60^o. The opening and closing speed for the unloaded unit was approximately 200 ms, which corresponds to a speed of 5.2 rads/sec. The device had a stall torque of 9 in-lbs, which equates to a pinch force at the finger tip of about 2.5 lbs (11.1 N). Both the ProDigits and the New Brunswick system should be capable of being fit to persons with trans-metacarpal amputations.

OUR MECHANISM

The motivation for our interest in externally-powered partial hand mechanisms stems from a temporary fitting our laboratory performed for a patient of the Rehabilitation Institute of Chicago (RIC). This patient had partial-hand, shoulder disarticulation, and trans-femoral amputations. This fitting was performed to give the patient some hand function for the duration of his stay at the RIC. The prosthesis fitted was self-contained, self-suspended, and used myoelectric control. The muscles of the thenar eminence provided the myoelectric control site. The importance of this fitting was the observation of the exceptionally functional movements, which were made possible by the unrestricted motion of the wrist. Most impressive was that his movements were not "amputee-like", which was attributed to the conservation of nearly normal wrist function.

The availability of small (10mm in diameter) DC motors enabled us to develop and build an initial prototype partial hand with powered fingers [7]. These motors were small enough to be placed within the fingers and thumb and were used in synergistic pairs to achieve reasonable speed and force. When an object is grasped a force is exerted with very little excursion while excursion of the grasping fingers usually occurs in space and requires very little force. In both cases the work involved is minimal. Childress [8] observed that this kind of prehension could be readily implemented using multiple motors that operate in synergy. One motor provides high speed and excursion but little force (fast side), another motor provides high force but little speed and excursion (force side). In this way the motors work to boost overall performance. An alternative to multiple motors is an automatic transmission as used by Otto Bock in their adult hands.

Our next prototype used three motors operating in synergy [9]. There were motors in the index and middle fingers to provide force, and a third transverse motor lying in the line of the knuckles to provide speed of opening of the fingers. The thumb was kinematically linked to move with the "Fast side", or knuckle motor, thus achieving a large width-of-opening. The volume of the ring and little fingers housed the battery and control electronics. As with our original prototype the two motors in the index and middle fingers operate independently. The total force generated by the hand is the vector sum of the force from each finger.

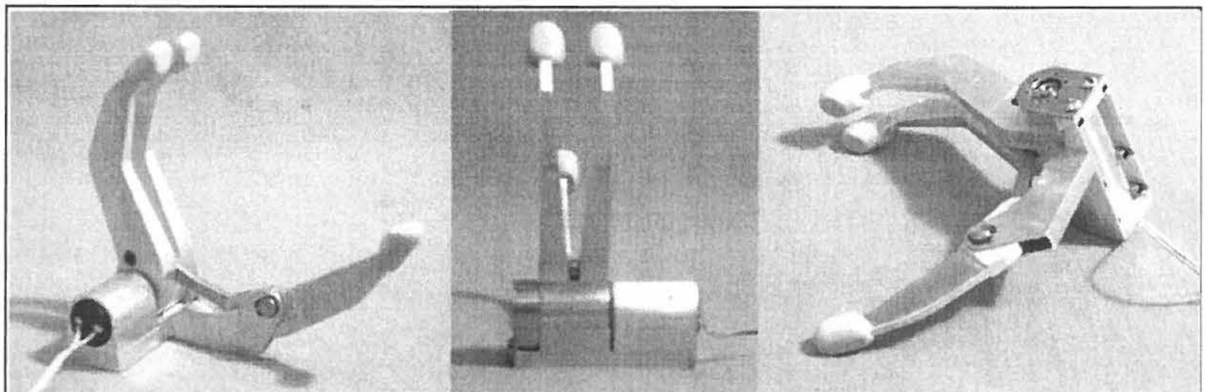

***Figure 1**: Photographs of our final single motor prototype*

The surprisingly good performance of the knuckle drive train prompted us to redesign this prototype. This effort resulted in a new single motor trans-metacarpal hand mechanism (Figure 1). In this mechanism performance is achieved through the use of a transverse knuckle drive that uses a planetary gear train. The novelty of this design lays in the use of the transverse drive housing as the axle upon which the hand's fingers rotate and the use of an entirely planetary based gear train with a 6 planet gear final stage to distribute the output torque. Using the drive housing as the finger axle allows for a hand mechanism with a hand width of only 60mm (2.25").

A planetary gear train was used because they are self-centering, with torque that is distributed equally on the planet gears. The use of a planetary gear train allows for a high gear ratio drive train capable of handling the high speeds and torques required to generate acceptable grip forces and hand opening speeds. As with the previous prototype, the thumb is kinematically linked to the fingers so as to provide a large width-of-opening. The weight of the current mechanism alone is 157g (0.35 lbs) and its length, in the closed position, from back plate to finger tip is 85mm (3.5").

Proportional myoelectric control sites on the residual partial-hand are the preferred mode of control. Ideally there should be two such sites using the intrinsic muscles of the hand. For example, closing could be provided by a site over the thenar eminence and opening by a site over the lateral aspect of the hypo-thenar eminence. Flexing/adducting the thumb would cause the hand to close while abducting the little finger would cause it to open. This kind of control might not always be possible, but even if only one EMG site is available three-state control could be used.

IN CONCLUSION

We have developed a new externally-powered partial hand prosthesis mechanism suitable for persons with amputations at or more proximal to the metacarpophalangeal joint. This single motor mechanism is smaller and lighter than our original multi-motor prototype and is capable of attaining pinch forces in excess of 12 lbs_f and has a speed in excess of 2 radians/sec. This performance is achieved through the use of planetary gearing, which allows for a very compact high gear ratio drive train capable of handling the high torques required to generate acceptable grip forces. Given the compact and lightweight nature of our mechanism we believe this mechanism could also have application in the fitting of persons with high-level upper-extremity amputation where weight considerations are of paramount importance. We also believe that preservation of wrist motion for positioning of the terminal device is necessary to achieve maximum function and cosmesis.

ACKNOWLEDGMENT

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REFERENCES

1. Ouellette, E.A., McAuliffe, J.A., & Carneiro, R.,: Partial-Hand Amputations - Surgical Principles. Chapter 7B. In: Atlas of Limb Prosthetics, 2nd Ed, Bowker, J. H., & Michael, J. W., (Eds), Mosby-Year Book, Inc., St. Louis, Missouri, pp199-216, 1992.
2. Michael, J.W. Partial-Hand Amputations-Prosthetic and Orthotic Management, Chapter 7B. In: Atlas of Limb Prosthetics, 2nd Ed, Bowker, J.H., & Michael, J.W., (Eds), Mosby-Year Book, Inc., St. Louis, Missouri, pp217-226, 1992.
3. Gow, D., Dick, T., Droper, E., London, I., and Smith, P.: The Physiologically Appropriate Control of an electrically Powered Hand Prosthesis. Proc. 4th World Congress of the International Society for Prosthetics and Orthotics, (ISPO), London, England, 1983.
4. Philipson, L., Pettersson, S-O, Waldenlöv, K, Geggie, C., and Gow, D.,: An Exploratory Fitting of a Prototype Transmetacarpal Electric Hand. Proceedings of the Seventh World Congress of the International Society for Prosthetics and Orthotics (ISPO), Chicago, Illinois, June 28 - July 3, 1992.
5. Gow, D.,: Experimental Powered Prosthetic Digit System. Proc. International Society for Prosthetics and Orthotics (ISPO), United Kingdom National Member Society, Quality Scotch Corner Hotel, 23rd - 25th October, 1997.
6. Biden, E., and Bush, G.,: Recent Advances in the Development of Partial Hand Prostheses. Proceedings of MEC'97, MyoElectric Controls/Powered Prosthetics Symposium. Institute of Biomedical Engineering, University of New Brunswick, Fredericton, New Brunswick, Canada, July 23-25, 1997.
7. Weir, R. F. *ff.*, (1989): The Design and Development of a Synergetic Partial Hand Prosthesis with Powered Fingers. Proc. RESNA 12th Annual Conference, New Orleans, Louisiana, June 25th-30th.
8. Childress, D. S. (1973): An Approach to Powered Grasp. Published in Advances in External Control of Human Extremities Proceedings of the Fourth International Symposium on External Control of Human Extremities, Dubrovnik, Yugoslavia, August 28th to September 2nd., 1972.
9. Weir, R. F. *ff.*, Grahn, E. C., and Duff, S. J.,: A New Externally-Powered, Myoelectrically Controlled Prosthesis for Persons with Partial Hand Amputations at the Metacarpals. Journal of Prosthetics and Orthotics. Vol. 12, No. 2, pp. 26 - 31, June 2001.

EVALUATION OF A NEW FLEXION WRIST INTEGRATED WITH ELECTRIC HAND

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SUMMARY

A new electric hand has been developed which integrates flexion/extension into an electric hand, with no increase in length over standard-length electric or body-powered hands. The development* has included a field trial with six wearers of the electric hand. All the wearers were surveyed on the actual tasks they performed with the hand, including those utilizing flexion/extension.

Electric Hands have typically operated with a single gripping mode, usually three-finger tip prehension. Our attempt at this stage is to enhance function in the hand, but without adding complexity to the gripping mechanism.

Passive and electric wrist rotation can add an important degree of freedom (DOF), but still provides only one of the natural wrist's three DOF. Additional degrees of freedom in the hand, however, would offer the amputee broader function of course, but practicality dictates the addition of a DOF without adding to the complexity of the hand.

In the past, integration of a flexion wrist with an electric hand has been accomplished with the addition of a component adding to the length of the hand/wrist combination. Typically, this combination has not been possible without sacrificing the quick disconnect feature of the hand as well, since the flexion components available have not contained the quick disconnect.

The development of the Motion Control Hand was done with a high emphasis placed on reducing the length of the drive mechanism. We sought to integrate a flexion mechanism into the hand, distal to the quick disconnect unit at the base of the hand. The development succeeded in reducing the drive mechanism length by approximately ¾ inch (1.9 cm), relative to earlier electric hands.

The shorter drive length also allows the Motion Control Hand to be configured without the Flexion Wrist mechanism, so that a "Short Hand" can be used for patients who require minimum length of the hand, e.g., wrist disarticulation or long trans-radial amputation length.

A simple flexion mechanism was sought, which would allow the wearer to easily lock or unlock the wrist position, and reposition the wrist in either a flexion or extension position. The mechanism that resulted allows the wrist to be flexed or extended to 30 degrees in either direction, and locked in each of the three positions.

The functional advantage of the flexion/extension joint is generally to allow greater positionability of the hand, for a much wider range of gripping positions. This allows positioning of the finger tips (the main gripping surfaces) closer to a wearer's mid-line, as well has more convenient gripping of an object which must be held level, e.g., trays, plates, pans, etc.

SURVEY RESULTS

Six experienced prosthesis wearers used the hand with flexion. At the time of the field trials, the MC Hand required either a Utah Arm or a ProControl to operate it, so it happened that three subjects used Utah Arms (all transhumeral level amputees), and three used ProControl 2 controllers (all transradial level, including two bilateral amputees).

All of the six subjects were adults, including 5 males, and one female. The average period of usage was 15.5 hours per day, so obviously, the group was composed of very active wearers.

We attempted to collect from the wearers a list of the typical tasks performed during their days, by asking about tasks in the categories of: "Grooming, Daily Activities, In The Kitchen, On The Job, and Hobbies". Out of an average task count of 18 per person, the average percentage of tasks in which the subjects would bend the wrist in flexion or extension was 55%. We felt this was quite a significant number, considering that prior to their use of the test hands, the function was simply not available to them.

Examples of tasks using the flexion feature included: donning/zipping coats, pulling doors open (more conveniently grasping the handle), holding books or reading materials (allowing the book to be held in a convenient position for reading), driving, and many others. The extension was used less frequently, but was specifically mentioned for eating, to grasp and turn a key, holding a nail, and even removing a gas cap!

Problems mentioned in the survey included some difficulty in reaching the release button for the flexion joint, which led us to redesign the button for easier reach. Also, the on/off button for the hand proved difficult for some subjects to reach early in the trial, which also was redesigned for easier access.

CONCLUSION

A great deal has been learned about the ways in which the prosthetic hand wearers utilize their hands for two-handed tasks, and the usefulness of the flexion/extension component. Design changes were made in response to the feedback from the wearers, and the device has evolved to the stage of a commercially released product.

Other versions of the electric hand have also evolved, including three shortened versions, and an integrated controller/hand version.

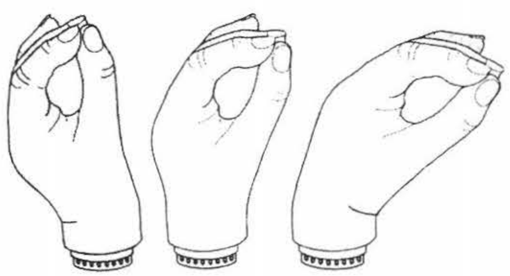

Figure 1. *Three positions of the flexion/extension wrist, which are 30 degrees in each direction.*

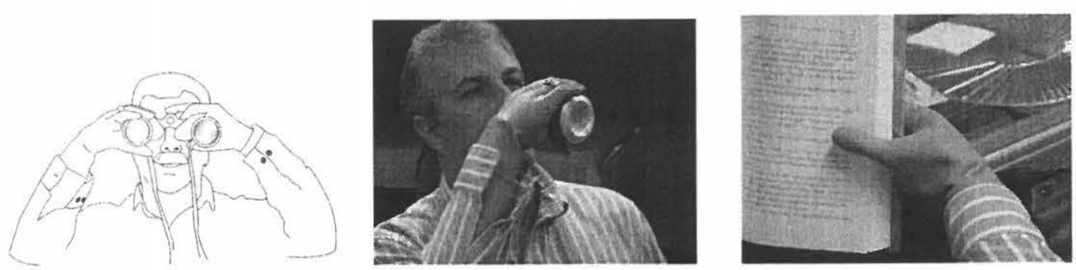

Figure 2. *Three examples of usual tasks made easier by the flexion or extension feature.*

*supported by NIH (NICHD) grant#2R44HD36ll9

OPTIMAL FIXED WRIST ALIGNMENT FOR BELOW-ELBOW, POWERED, PROSTHETIC HANDS

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INTRODUCTION

Generally, a prosthetic wrist has one degree of freedom in a supination/pronation sense. Some do allow bending of the wrist [1-4], however, due to the complexity of integration and operation, they are not as commonly used. The restricted motion of the prosthetic wrist is compensated for by additional motion of the shoulder and elbow. Not only can this additional motion appear awkward making the person self-conscious [5], but it may also increase the risk of joint injury [6].

In an attempt to balance function, cosmesis, and "dynamic cosmesis" (minimal compensatory motion), the prosthetist aligns the wrist with the forearm in some combination of flexion/extension and ulnar/radial deviation. Changing the alignment once it is fixed requires significant effort. Therefore, having a sense of the optimal alignment would be very beneficial. Guidelines for wrist alignment do exist [7 and 8], but while some consider them "very applicable"[9], others feel "there is a need for updated text materials" [10]. These guidelines were written over thirty years ago (1958 and 1971 respectively) and this lag of standards behind technology has left prosthetists to develop their own alignment procedures that sometimes contradict those of their colleagues. Where one prosthetist aligns a wrist at 10 degrees extension, another uses slight ulnar deviation and flexion [11-13]. While the experienced prosthetist has a practical knowledge from which to decide on an individual wrist alignment, the novice practitioner requires guidelines on which to base decisions.

Our goal was to determine which, if any, of the wrist alignment angles commonly used allows elbow and shoulder motions to be closest to "normal", resulting in optimal dynamic cosmesis.

TEST METHODOLOGY

We compared the upper-body motion of normally-limbed people performing activities of daily living (ADLs) under two conditions: 1) the wrist unrestricted (normal, N), and 2) the wrist splinted in five different alignments: 0° Straight (S), 10° Extension (E), 10° Flexion (F), 10° Radial deviation (R), and 10° Ulnar deviation (U). In addition to restricting wrist motion to imitate a prosthetic wrist, the hand was splinted for all tests to imitate the grip of a myoelectric hand. We subsequently repeated the experiments using 20° offsets with a different subject population [14]

For each of the experiments, the ten, adult subjects (five female, five male) sat at a desk, and at their own pace, performed four ADLs: Cup (C) - drink from a cup, Spoon (S) – eat soup with a spoon, Sandwich (W) – bring a sandwich level with the mouth, and Zipper (Z) – zip a jacket. A 3-camera VICONTM 140 motion analysis system was used to record 3D coordinates of reflective markers attached on the trunk, head and right arm of each subject. The marker coordinates were used to calculate upper-body motion (ubm) angles of the head, shoulder girdle, glenohumeral joint, and elbow joint

RESULTS

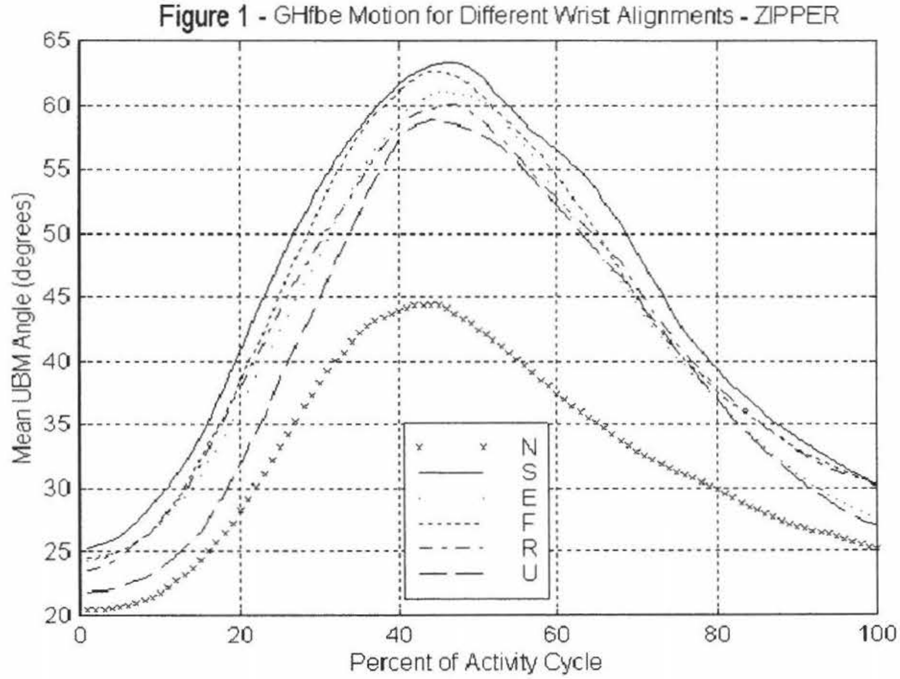
For both 10 degree and 20 degree offset tests, the ubm angle data for the ten subjects was averaged to result in one data set per wrist alignment for each ubm angle and each ADL. As an example of the data, Figure 1 shows forward/backward extension of the glenohumeral joint during the zipper activity. To determine which, if any, of the fixed wrist alignments caused significantly less compensatory motion than the rest, extreme (max/min) mean values were compared. Radial deviation showed a statistically significant ($p<0.05$) advantage over straight, extension, and ulnar deviation alignments, but only in reducing elbow flexion by at most 7°, and only for the cup and spoon ADLs.

Results showed that not only were the extreme values close (less than 8°) among the fixed alignments, but the patterns of motion are also very similar during the entire activity.

Results also provided clear evidence that compensatory motion did occur. While the fixed wrist alignment measures were similar to one another, normal (unrestricted) wrist alignment often caused less motion. Statistically significant differences were seen between normal and fixed alignments in all but the sandwich activity.

Results for the tests with 20° offsets yielded similar results without any consistent differences between the orientations tested.

Figure 1 - GHfbe Motion for Different Wrist Alignments - ZIPPER



CONCLUSION

No optimal fixed wrist alignment was clearly distinguished among the alignment angles tested. It can therefore be concluded that a person will move approximately the same regardless of the prosthetic wrist alignment selected within +/- 20° of flexion/extension or +/- 20° radial/ulnar deviation. Therefore, when fixing a wrist alignment, dynamic cosmesis should be less of a concern than function and cosmesis. Stated another way, the act which changes compensatory motion is fixing the wrist, and the angle at which the wrist is fixed has only a small effect on the compensatory motions.

Recommendations for future work include testing other ADLs, testing alignments that combine flexion/extension with radial/ulnar deviation, modifying the wrist splint to be more rigid, and modifying the splint to control forearm supination/pronation.

REFERENCES

1. Liberty Technology, *Prosthetics and Orthotics*, Liberty Mutual Insurance Group, Boston, 1997.
2. Williams, T.W. Reducing the Houston Flexible Wrist to Practice. Presented at the *MyoElectric Controls/Powered Prosthetics Symposium*, Fredericton, New Brunswick, July, 1993 (Proceedings pp. 68-71).
3. Hosmer Dorance Corporation, *Twelfth Edition Catalogue*, 1992.
4. Otto Bock Orthopedic Industry, *Arm Components*, 1995/1996.
5. Jacobs, R. The Addition of a Flexible Wrist Unit to Enhance Prosthetic Function. Presented at the *MyoElectric Controls/Powered Prosthetics Symposium*, Fredericton, New Brunswick, July, 1993 (Proceedings pp. 63-67).
6. Cooper, J.E., E. Shwedyk, A. Quanbury, and J. Miller. Elbow joint restriction: effect on functional upper limb motion during performance of three feeding activities. *Archives of Physical Medicine and Rehabilitation*, 74(8):805-809, 1993.
7. Santchi, W.R. and M.P. Winston, Editors. *Manual of Upper Extremity Prosthetics, 2nd Edition.* University of California at Los Angeles, 1958.
8. New York University, Post-Graduate Medical School, Prosthetics and Orthotics. *Upper-Limb Prosthetics Supplement to Upper-Extremity Prosthetics.* New York University, New York, New York, 1971.
9. Childs, S. (pacorthoprosser@juno.com). (1997, Oct 17). *Prosthesis wrist alignment.* E-mail to O. Stavdahl (Oyvind.Stravdahl@itk.ntnu.no). (1998, Apr 23). *Prosthetic wrist alignment.* Forward to J. Landry (y2r9@unb.ca).
10. Okumura, R. (okumura@u.washington.edu). (1997, Nov 13). *Summary: Prosthesis wrist alignment.* E-mail to O. Stavdahl (Oyvind.Stravdahl@itk.ntnu.no). (1998, Apr 23). *Prosthetic wrist alignment.* Forward to J. Landry (y2r9@unb.ca).
11. Andrew, J.T. (JTAndrew@aol.com). (1998, April 28). *Prosthetic wrist alignment.* E-mail to J. Landry (y2r9@unb.ca).
12. Andrew, J.T. (JTAndrew@aol.com). (1997, Oct 13). *Prosthesis wrist alignment.* E-mail to O. Stavdahl (Oyvind.Stravdahl@itk.ntnu.no). (1998, Apr 23). *Prosthetic wrist alignment.* Forward to J. Landry (y2r9@unb.ca).
13. Bush, G.: Personal Communication, April 23, 1998.
14. Cormier, R.M. "Optimal Fixed Wrist Alignment for Prosthetic Hands", Senior Project, Mechanical Engineering, University of New Brunswick.

MECHANICAL CHANGES AND NEW CONTROL OPTIONS IMPROVE THE FUNCTION OF THE LTI-COLLIER SHOULDER JOINT

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INTRODUCTION

When a new prosthetic component is introduced to the market, it often needs further improvements or accessory devices that only become apparent when clients and clinicians have had a chance to put the device to all of its possible uses. Since the LTI-Collier Locking Shoulder Joint was introduced, a number of improvements have been made both in the product and its accessories in response to user feedback. In the original concept, the user was assumed to need to unlock only momentarily between activities. For most users this assumption is wrong. The most popular feature of the joint turns out to be its ability to remain in free swing. Improvements in the joint itself have either added strength where required or have increased the number of users who can readily put the joint into free swing. To make free swing accessible to more amputees, an electric elbow lock was developed. There are now many ways to activate this lock

REDUCING FRICTION DURING UNLOCKING

The unlocking lever should move easily when pulled, and then it should snap back into the locked position quickly. A bearing, consisting of six balls placed under the ring, reduced friction by half. It is now easier for the user to activate; but more important, the lock fully engages more reliably, which increases the life of the mechanism.

MAKING THE UNLOCK CABLE EASY TO INSTALL

The lever that unlocks the joint requires a tangential force in the plane of rotation for activation. Prostheses returned for repair showed that people were neither attaching the activation cables to the lever tangentially nor at the right height. Furthermore the cables could not be tensioned properly causing wear inside the joint. A new anchor plate was designed to correct these problems. It mounts under the joint and will clamp any Bowden cable so that it points directly at the operating point on the lever. Tension is adjusted by moving the cable sheath under the clamp. There are three choices for cable operation of the unlocking mechanism. They are built out of Blatchford knee lock mechanisms or the Hosmer Sierra Nudge Control. With all units the standard cable is replaced by a heavy-duty sheath and a Teflon™ liner. The Blatchford units are also supplies with an added mounting plate. Figures 1 and 2 show the unit before and after the change while Figure 3 shows the optional mounting plate supplied with the Blatchford units. This plate makes it easier to mount the mechanism on the outside of a prosthesis.

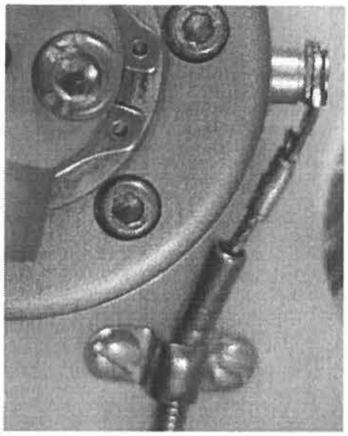


Figure 1. This mounting bends the cable as it attaches to the lever, and the clamp is far below the plane of the lever.

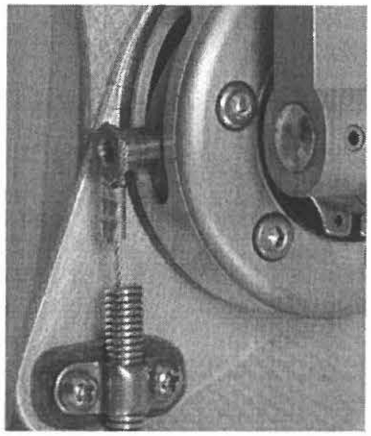


Figure 2. *This new plate lines up the cable sheath perfectly. To adjust tension loosen the clamp and move the sheath.*

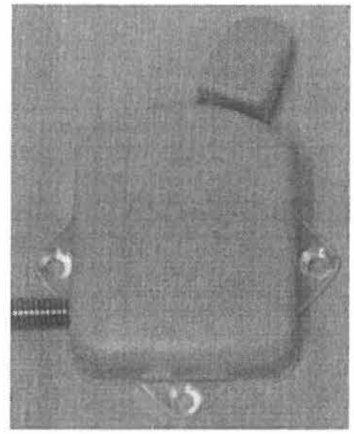
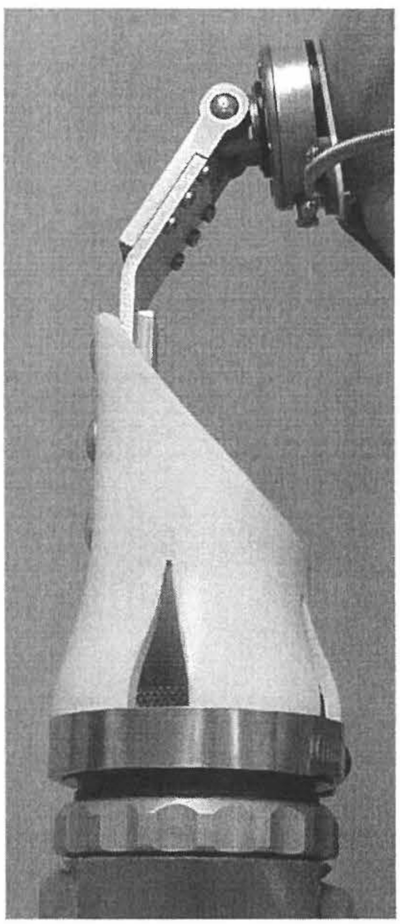


Figure 3. *This optional mounting plate makes the Blatchford unit accessible from the top.*



THE UPPER-ARM ATTACHMENT PLATE

The original joints were supplied with an upper-arm attachment plate that could be bent when an offset was required. Because the alloy had a low yield, it also bent under normal usage, and failed on heavy users. However, the ability to provide an offset was a popular feature. Redesign of the plate was the answer. The new high-strength plate has a bend on one end and is 50% thicker than before. Strength has been doubled. The bend is permanent because the new alloy cannot be bent by ordinary means. Since not everyone wants an offset, there are attachment holes on both ends. To choose between an offset and a straight plate technicians simply cut off the end they do not need. The piece that holds the plate to the joint has also been made thicker, and the friction screw is easier to turn. These changes add a little weight but greatly improve reliability. There have been no failures since they were made.

Figure 4. *The heavy-duty bent plate is used here to attach a temporary upper arm fitting to a Boston Elbow.*

PROVIDING HANDS-FREE UNLOCKING

The people who need a locking shoulder most are those with bilateral involvement. They cannot simply reach over and push a lever. At first the only choice was to use the Hosmer Sierra Nudge Control. However, many users already had too many controls around their chins, so an electric unlock mechanism seemed in order. Several units were jury rigged out of available cable activation units, and proved that a custom design was needed. [1] Figure 5 shows the present Electric Lock/Unlock for an LTI joint. The mounting bracket is a thin plate that goes under the joint. It holds a motor, reduction gear, and screw-activated sliding block. The block engages the unlock lever. A 6V battery will drive the block back and forth to lock and unlock the joint. Note that an offset upper-arm plate may be used to keep the motor and upper arm from interfering.

Activating The Electric Lock

Motor activation is simple in principle, but the details can cause difficulty. The motor was selected to operate from 6 to 9V, but at the high end of this range it can be damaged if run continuously. When the lock is to be switch operated, it is supplied with a series Polyfuse™. With a control circuit, motor stall current can be sampled to stop the motor, so a fuse is not required. The clinical challenge with the lock is minimizing battery weight. Whenever another battery-operated device is present, the two units should share a common battery if possible. As for controllers, a spare hand-control circuit will work well if it incorporates the battery saving feature. Otherwise, one of LTI's many VariGrip II controllers will do. The lock works best when it is teamed up with the LTI Boston Digital Arm. The arm program offers the clinician more control choices than any other system, but there are almost as many choices with the VariGrip II.

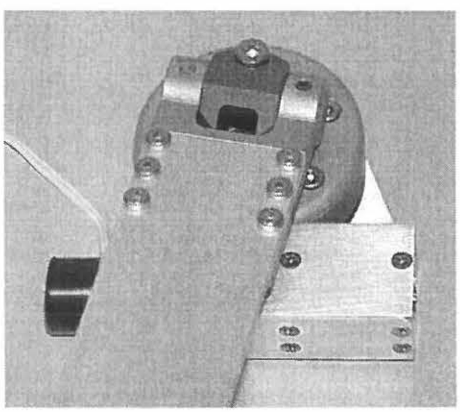


Figure 5. *The Electric Lock/Unlock is shown here with a straight plate. The offset would be better and prevent the plate from hitting the drive mechanism.*

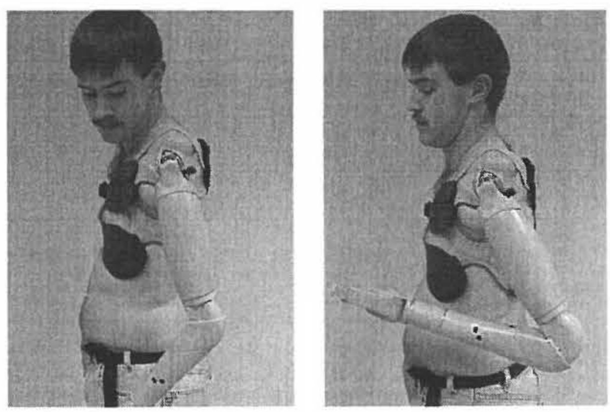


Figure 6. *Here the shoulder joint is in free swing. Note how the elbow moves backward with respect to the shoulder as the user flexes the elbow. This motion is in important form of dynamic cosmesis.*

Control Choices

While the lock functions well when operated by a current-carrying switch, this setup forces the user to remain in contact with the switch until the lock or unlock operation is complete. Users much prefer to simply touch a switch with the circuit taking over completion of the task. With the Boston Arm or VariGrip II controllers there are many ways to activate the Lock. They all can be set up so the user triggers the operation, which is then completed automatically. A single switch closure, push on a Touch Pad™, or cocontraction can trigger the circuit to alternately lock and unlock the joint. Since the user can hear the motor activate, no user feedback is required. Users can even be trained to use more than one cocontraction strategy, so that one type shifts myoelectric control between devices, while a second triggers the alternate lock-unlock action.

User Convenience

It is worth mentioning some of the choices that users have requested. A simple chin-activated switch is popular, but other locations often work better. One unilateral user needed to place his arm out front for work tasks. For him a switch was placed on the forearm, where it could be activated by the same hand motion being used to reposition the arm. For other users an extra Touch Pad has been used as a trigger. Simple body motions are best. Another motion is that of a short humeral neck. It can push out against a Touch Pad or switch.

The Triggering Spike

Sometimes a user will have a weak muscle that cannot be used for proportional control. If the muscle can generate a momentary spike of activity, however, it may be ideal for operating the lock. In fact any momentary signal can be used. Recently LTI has reexamined the problem of triggering the selection of multiple devices. While the rapid cocontraction is the least likely trigger to be inadvertently activated, there are other unique patterns. For instance a momentary strong spike of a single muscle can be characterized such that it becomes another trigger choice. Use of existing myoelectric sites avoids adding switches and other extra hardware.

REFERENCES

1. Anderson, C., Farnsworth, T. & Stevens, G. "Adaptation of locking shoulder joints to increase functional range of motion for bilateral upper limb deficiencies", Proceedings, Myoelectric Controls/powered Prosthetics Symposium, University of New Brunswick, 146-148 1999.

HYBRID APPROACH TO BILATERAL UE PROSTHETIC REHABILITATION

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ABSTRACT

Patients with "high" level bilateral upper extremity deficiencies require maximum functional rehabilitation to increase independence and self-care skills. Traditional prosthetic rehabilitation for these individuals utilizes various control mechanisms including body power, electric, and hybrid systems.

Rehabilitation teams rarely gain experience with multiple cases using varied control methods. In most cases systems are recommended and fit based on the limited past experiences and training of the rehabilitation team members and the local prosthetist.

Body powered control provides the user with fine motor skills and excellent proprioceptive feedback through Extended Physiological Proprioception (EPP). Traditional body powered hooks provide the user a superior line of sight and grasping patterns that allow for easy manipulation of small objects. Limitations to body powered bilateral systems include limited grip strength (for voluntary opening devices), limited ability to easily preposition components, contralateral controls interference, and minimal requirements of strength and excursion.

Electric powered systems provide the user with excellent control (i.e. using proportional control systems). Current electric powered terminal devices provide the user with superior grip strength and varied grasping patterns that easily handle larger and heavier objects. The user can achieve operation and preposition of components with little effort.

Hybrid designs traditionally combine body powered and electric control within a single prosthesis. This combination can provide the user with the best functional components from each system to improve overall functional outcome.

This presentation will look at bilateral prosthetic wearers using combinations of body powered systems on one limb and electric powered on the contralateral side. Theory of prosthetic fitting, controls design, and operation will be discussed. Multiple case studies will be utilized as examples for all levels including transradial, transhumeral, shoulder, and combined levels of bilateral upper limb absence.

FINAL PAPER NOT RECEIVED AT PRESS TIME

NOTES

August 21, 2002 (Wednesday)
1:30pm – 3:00pm and 3:30pm – 4:30pm

Theme: Prosthetics

Invited Speaker:
1:30 – 2:00
John M. Miguelez
Enhancing function: the glimcher test at its best (hybridization of servo and myoelectric control for the transhumeral level limb deficient individual)
Experiences with the Otto Bock ergo elbow

2:00 – 2:15
Christian Hell
Solutions for ergonomics in high level amputation treatment

2:15 – 2:30
Randall D. Alley
Advancement of upper extremity prosthetic interface and frame design

2:30 – 2:45
Thomas Bertels
Results of a study with the goal to optimize harnesses by means of up-to-date biomechanical engineering

2:45 – 3:00
Donald L. Russell
Progress towards a biomimetic prosthetic arm

3:00 – 3:30 Refreshment Break / Vendor Displays

3:30 – 3:45
Gerald Haslinger
EMG electrodes for myoprosthetics : design principals and application tips

3:45 – 4:00
Wayne Daly
Electrodes installed in roll-on suspension sleeves

4:00 – 4:15
Ian Whatmough
Custom silicone liners for upper extremity prosthetics

4:15 – 4:30
Jack Uellendahl
Experience with silicone suction sockets using myoelectric control

ENHANCING FUNCTION: THE GLIMCHER TEST AT ITS BEST (HYBRIDIZATION OF SERVO AND MYOELECTRIC CONTROL FOR THE TRANSHUMERAL LEVEL LIMB DEFICIENT INDIVIDUAL)

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ABSTRACT

In the late 1960s, Dr. Melvin Glimcher, was Professor of Orthopedic Surgery and was responsible for amputee rehabilitation at Harvard Medical School. He became aware that the transhumeral level amputee population had low prosthetic success rates. In an effort to improve the success rate, Dr. Glimcher analyzed prosthetic function of transhumeral level amputees. Dr. Glimcher observed that positioning of the mechanical elbow, locking the elbow, and finally opening and closing the terminal device was very inefficient. He concluded that one cause for diminished success rates was the necessity to lock the mechanical elbow prior to operating the terminal device. He implemented a challenge to the prosthetic team to design a control mechanism that allowed for simultaneous control of the elbow and terminal device. Over time, the requirement for a transhumeral level amputee to simultaneous control the elbow and terminal device has become known as the "Glimcher Test". The most popular method to obtain independent control of elbow and terminal device is the hybrid prosthesis. While there are many combinations possible when designing a hybrid prosthesis, most often they involve the marriage of body-powered control and electronic control. One disadvantage of using body-power control is the force that the harness exerts on the contralateral shoulder and axilla. This force can often lead to axilla tissue irritation, breakdown and in many cases peripheral neuropathy. This paper will examine the use of an electrically-powered prosthesis with independent control of the electric elbow and electric terminal device by using myoelectric and servo control schemes.

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EXPERIENCES WITH THE OTTO BOCK ERGO ELBOW

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ABSTRACT

The patient population that elect hybridization of their prosthesis to include an electrical terminal device and body-powered mechanical elbow is limited. The population typically is comprised of transhumeral level amputees. Elbow disarticulation level amputees typically do not possess sufficient space for traditional mechanical elbow units to be utilized without a contralateral limb length discrepancy. Shoulder disarticulation level amputees often have difficulty producing sufficient excursion and/or force to position the elbow or engage the locking mechanism. Other factors that limit utilization of a hybridized approach can be attributed to the lack of integrity of the electrical cable as it crosses the axis of flexion, extension and humeral rotation at the elbow center, and diminished cosmesis that occurs when the electrical cable exits the humeral section, crosses the elbow joint center, and enters the radial section of a prosthesis. Cable damage is quite frequent secondary to exceeding the flexion cycle characteristics of the cable or inadvertent torque placed on the cable.

The Otto Bock ERGO Elbow System offers several enhancements over traditional mechanical or body-powered elbows that address these limitations. A thorough understanding of the design features will allow the rehabilitation team to consider a hybridized approach on a larger patient population while increasing the success rate of those individuals who have already elected this method of prosthetic intervention. The purpose of this paper is to detail the design enhancements of the ERGO Elbow that address many of the challenges that surface when using traditional mechanical elbows for hybrid prostheses.

FINAL PAPER NOT RECEIVED AT PRESS TIME

SOLUTIONS FOR ERGONOMICS IN HIGH LEVEL AMPUTATION TREATMENT

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ABSTRACT & INTRODUCTION

The needs of above elbow amputees vary a lot depending on the level of amputation and the patient's physical and mental abilities. Eliminating the hassle of a harness for a body powered elbow lock has always been a very tempting idea. Especially if the price you have to pay for this is not as high as the weight of a fully electrified elbow joint. Yet the success of such a fitting depends very much on the controllability of the lock. Furthermore a low level of noise is as essential as an instantaneous activation of the lock.

This presentation will give you an idea of the possibilities which a new high-tech electronically controlled lock offers to optimize high-level myoelectric fittings. Furthermore, it will explain different control options which allow an adaptation to the individual needs and capabilities of the patient

ERGOARM ELECTRONIC PLUS 12K50

The ErgoArm Electronic plus provides additional functional possibilities, enabling the prosthetist to do high level fittings in some cases even without the need for a harness. Its unique electronically controlled lock is the heart of this high-tech device. Controlled by signals coming from a myo electrode or a switch, a tiny but powerful drive unit locks or unlocks the elbow joint instantly within milliseconds. An important development goal was to make the locking / unlocking process as quiet as possible. The unique rigid but light weight design eliminates unwanted noise successfully, so that in everyday environment the locking and unlocking can hardly be heard.

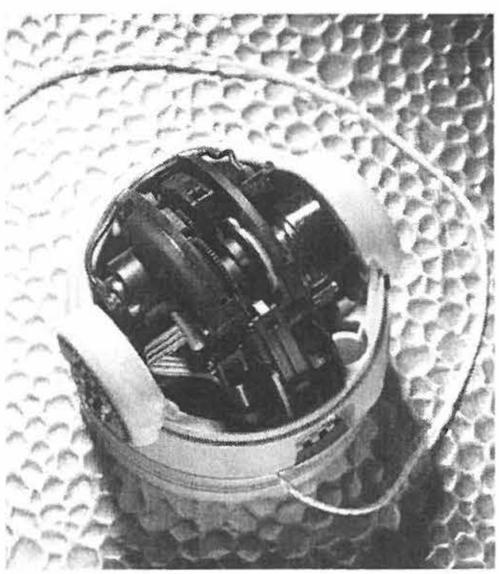

Figure 1: *Electronic controlled elbow lock*

Control Options

Seven different control schemes allow the choice of an optimal solution for the individual amputee. These schemes are easily activated by means of small colored coding plugs. No computer is needed, as thanks to microprocessor technology the "Patient-Machine interface" has been optimized to meet the demands of most patients. In addition, this saves the prosthetist's time as there is no need for time consuming PC adjustments.

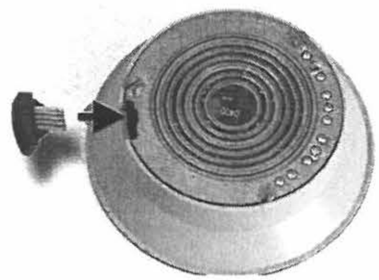

Figure 2: *Electronic Housing with integrated slot for coding plugs*

After the battery is inserted into the prosthesis the current switching mode will be indicated by a vibration feedback signal. For the safety of the patient the ErgoArm Electronic plus also includes a mechanical (un)locking possibility if the battery runs empty: A pull cable which can be attached to the socket.

The ErgoArm Electronic plus may be controlled by means of electrodes, switch or a combination of electrode or switch. These are the available control schemes:

Switch:

1. Actuating the switch alternately locks or unlocks the elbow joint

Electrodes:

2. A Co-contraction alternately locks or unlocks the elbow joint
3. Make a Co-contraction to get into elbow mode. Now signals from one electrode lock the elbow joint as the other one unlocks it. Co-contract again to leave elbow mode
4. As 3. , but leaves elbow mode automatically if there is no signal within 10 seconds.

Switch and Electrodes:

5. As long as the switch is actuated the control is in elbow mode. Now signals from one electrode lock the elbow joint as the other one unlocks it. Release the switch to leave the elbow mode.
6. Actuated the switch to get into elbow mode. Now signals from one electrode lock the elbow joint as the other one unlocks it. Actuate the switch again to leave elbow mode.
7. As 6. , but leaves elbow mode automatically if there is no signal within 10 seconds.

Slip-Stop

A unique feature of the ErgoArm, the Slip-Stop function for a controlled lowering of the forearm, has also been integrated into the electronically controlled lock: If the control is in elbow mode (3., 4., 5., 6., 7.) a small signal on the opening electrode releases the lock temporarily.

Once the signal drops, the elbow locks instantly. Slip-Stop makes it easier for the patient to exactly position the forearm.

ErgoArm Family of Elbows

Together with the ErgoArm Electronic plus there is now a family of 4 different ErgoArms. Every ErgoArm version fits the demand of a certain fitting option.

	Lock mechanical	Lock electronical	Easy Plug	AFB	Slip & Stop
12K41 ErgoArm: for body-powered prostheses	X				X
12K42 ErgoArm plus: for body-powered prostheses	X			X	X
12K44 ErgoArm Hybrid plus for hybrid prostheses	X		X	X	X
12K50 ErgoArm Electronic plus for myoelectrical prostheses	X	X	X	X	X

Figure 3: *ErgoArms*

All of them are based on the same sturdy and light weight construction. This improves reliability and makes it easy to handle them.

ADAPTER 13Z68

The Adapter allows the easy connection of any ErgoArm to a prosthesis which was built for use with a Hosmer elbow joint. After screwing the Adapter on, the ErgoArm is ready to be connected for evaluation or replacement.

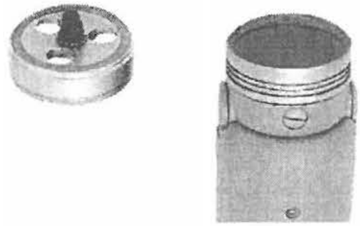

Figure 4: *Adapter*

LINEAR TRANSDUCER

Another important solution for high level amputation treatment is the Linear Transducer. If integrated into a harness system, this device transforms the patient's body movement into proportional electric signals. This signals can be used to proportionally control an electric device such as an electric hand, an electric Greifer or an electric elbow joint.

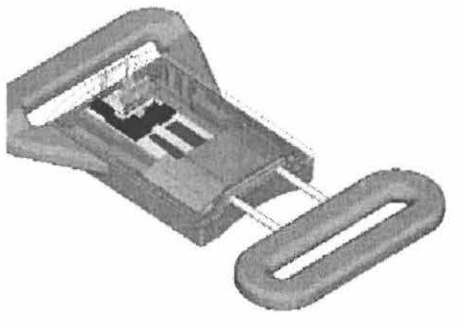

Figure 5: *Linear Transducer integrated in a harness*

Figure 6: *Linear Transducer*

CONCLUSION

Higher amputation levels have less possibilities to control a prosthesis whilst the need for additional functions is increasing. Therefor the treatment of high level amputations has always been a special challenge. Technological progress such as the introduction of microprocessors in the field of upper extremity prosthetics lead to new innovative components. These provide new solutions for ergonomics in high level amputation treatment.

ADVANCEMENT OF UPPER EXTREMITY PROSTHETIC INTERFACE AND FRAME DESIGN

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ABSTRACT

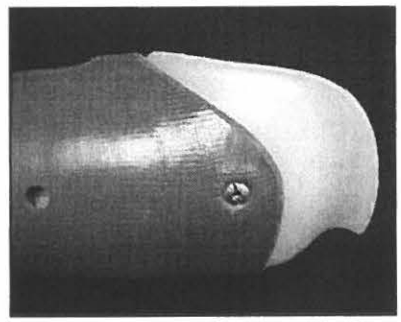
Although traditional upper extremity prosthetic interface and frame (collectively referred to here as "interface") designs have enabled many individuals to integrate prostheses into their rehabilitation plan, the biomechanical attributes and other parameters of these designs have not been significantly reviewed and improved upon until recently. In the last decade, a multitude of design innovations have been incorporated, which have resulted in wearers reporting superior comfort, suspension, stability, and range of motion, among other advantages. In most cases, when paired with a variety of control systems, the new designs appear to be inherently more efficient in terms of force transmission and motion capture, and more functionally consistent than traditional types of "sockets". It is the intention of this paper to highlight these novel design elements, as well as to discuss the biomechanical principles involved, to enable prosthetic users and other individuals to better understand these advanced interfaces.

INTERFACE DESIGN CRITERIA

The initial step in the improvement of an interface design is to first gain an understanding not only of the underlying functional intent of its constituent structures, but the resultant effect of these structures upon the critical elements responsible for successful prosthetic utilization. The interface designs for three levels or degrees of absence will be discussed. These are the radioulnar (transradial), humeral (transhumeral) and the glenohumeral disarticulation and interscapulothoracic (shoulder disartic and forequarter, respectively) levels, the latter two incorporating essentially the same design, albeit with minor changes to auxiliary suspension. A brief comparison and contrast of both the physical and biomechanical properties of the traditional interface model with its more contemporary associate will be given.

RADIOULNAR

Traditional self-suspending interface designs at the radioulnar level typically focus on three distinct areas: 1) the supracondylar component; 2) the cubital component; and 3) the olecranonal component. The soft tissue and skeletal anatomy distal to these regions are generally enclosed with a simple circumferential shape for volume containment. The ACC interface, in addition to these elements, adds a relief and modification to the antecubital region.



Northwestern style interface

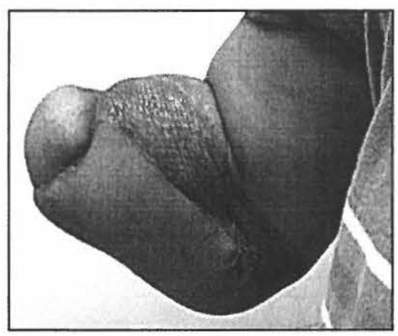
The Muenster design focuses on increasing anterior-posterior (AP) compression in the cubital fold. The Northwestern design focuses on increasing mediolateral (ML) compression proximal to the epicondyles. The Anatomically Contoured and Controlled (ACC) interface increases AP and ML compression but focuses them in different areas. In addition to the antecubital region described above, consideration is also given in the ACC interface to the distal humerus and the proximal portion of the extensor carpi ulnaris.

Comfort

The supracondylar component as addressed by the Northwestern design results in added suspension throughout the full range of motion, but also compresses the sensitive area over both the medial and lateral intermuscular septum, as well as the ulnar nerve and the medial antebrachial cutaneous nerve, hence negatively affecting comfort. The Muenster design utilizes AP compression primarily, but inherent instability can cause slight compression discomfort at the sensitive regions described above in dynamic situations. In both designs, a lowered anterior brim minimizes the effective lifting area of the forearm, and hence may increase lifting discomfort. The ACCI reduces compression in sensitive areas while concurrently increasing compression posteroproximal to the cubital fossa, and on either side of the antecubital relief, maintaining comfort while increasing suspension throughout the range of motion. In both the Muenster and Northwestern designs, the shape of the olecranonal modification often creates shear stress and compression problems as the olecranon begins to displace toward the posteroproximal brim under flexion. The ACC interface modifies the shape of the olecranonal relief to minimize or eliminate olecranon contact during ranging as well as donning and removal.

Cosmesis

Both the Muenster and Northwestern designs reduce the proximal extent of the anterior brim, and hence create a buildup of redundant tissue in the antecubital region with prolonged use. The ACC interface's antecubital region prevents significant redundant build-up and its greater ML compression tends to more readily conceal its more extensive medial and lateral stabilizers.

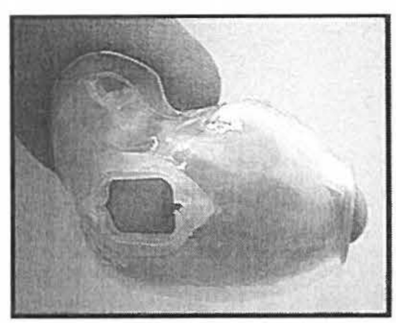


Redundant antecubital tissue caused by reduced anterior trim line

Stability

In the Muenster and Northwestern designs, rotational stability is minimally provided due to the general absence of medial and lateral stabilizers. AP stability is often reduced during ranging and is often lost at full flexion as the olecranon displaces posteroproximally.

In the ACC interface, the medial and lateral stabilizers significantly improve rotational stability. In addition, ancillary rotational resistance is provided by the modification of the antecubital area and displacement of the olecranon throughout the flexion range is minimized by the vertical segment of the interface proximal to the olecranon.



ACC interface in diagnostic phase

Suspension

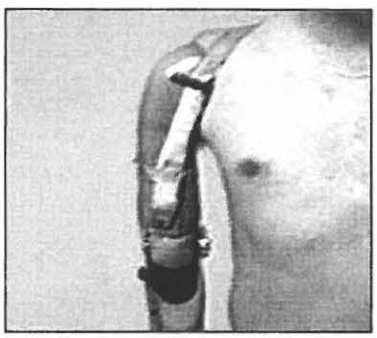
Difficulties in achieving adequate suspension with the traditional designs arise in terms of either the discomfort of applied compression in sensitive areas, or the lack of adequate compression in non-sensitive areas. Additionally, the inclined olecranonal relief typically permits translation of the residual limb out of the interface with the arm held over the head and the elbow joint is fully flexed. The ACCI focuses on maximizing compression in areas deemed suspensory and minimally sensitive. The olecranonal modification inhibits displacement of the olecranon posteroproximally.

Range of motion

Both the Muenster and Northwestern interfaces address elbow flexion requirements by lowering (reducing the proximal extent of) the anterior brim. This increases the build-up of redundant tissue just proximal to the brim, which in itself is a primary inhibitor of full flexion. The ACC interface extends deeply into the cubital fold, while allowing a relief for the biceps tendon as well as the antecubital tissue to expand into during flexion. In addition, by significantly reducing the AP dimension, flexion is increased.

HUMERAL

The more traditional style focuses on basic volume containment along the humeral shaft and vertical loading over the shoulder complex via the proximal portion of the socket.



Traditional humeral interface

Tom Andrew C.P.'s Dynamic Socket applies significant ML compression along the length of the humerus while lowering the lateral brim of the interface to a point between the axilla level and the acromion. In addition, anterior and posterior stabilizers are utilized. A derivative of the Dynamic Socket provides minimal ML compression distally while maximizing it proximally, and concentrates AP compression into a smaller area than can be found in the Andrews design.

Comfort

The most common complaint is this design's over-the-shoulder style suspension, in which the socket rests on the superior surface of the humeral complex, and discomfort caused by this interface's inherent instability. Both the Dynamic Socket and its hybrid cousin eliminate the interface contact on the proximal surface of the shoulder complex resulting in greater comfort during abduction and with increased heat dissipation.

Cosmesis

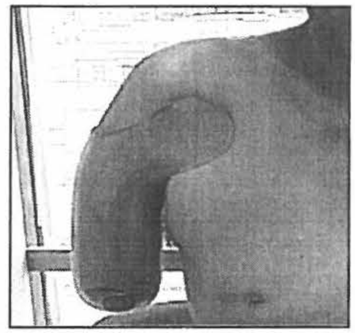
Inherent instability and proximal interface coverage reduce cosmesis in the traditional designs when the arm abducts or the ipsilateral scapula is elevated. Inherent stability in dynamic situations and reduced trim lines improve cosmesis in newer interface designs.

Stability

The newer styles utilize anterior and posterior stabilizers as well as ML compression, unlike traditional designs, to limit rotation in both the axial and sagittal planes.

Suspension

As previously stated, traditional humeral interfaces rest on the shoulder complex and occasionally on the trapezius in order to reduce vertical displacement under load. The newer designs utilize a shoulder saddle in most cases in order to achieve primary suspension. In addition, a secondary suspensory component is provided by AP compression anteriorly at the proximal overlap of the sternocostal and clavicular heads of the Pectoralis major, medial to the delto-pectoral groove, and posteriorly at the level of the Infraspinatus.



Humeral interface with anterior stabilizer shown

Range of motion

In the traditional interface, range of motion is significant; however, stability at the extremes is precarious. The Dynamic Socket and its derivative simply reduce the medial projection of their anterior and posterior stabilizers to optimize the functional envelope while retaining a significant degree of stability in all planes and at the extremes of available glenohumeral motion.

GLENOHUMERAL DISARTICULATION AND INTERSCAPULOTHORACIC

A thoracic socket design that encloses and rests on the shoulder girdle is still the most common interface style used today. Also referred to as "basket-type", "bucket" or "encapsulatory", this design has been used for decades. Newer interface designs rely on a frame-style approach, where only enough proximal, distal and medial projection is utilized to achieve adequate stability.

Comfort

Poor heat dissipation and discomfort from instability are the most obvious and negative aspect of traditional thoracic interfaces. Frame style interfaces such as the XFrame cover minimal tissue, and are significantly more stable. Similar to the humeral interfaces, the XFrame utilizes a discontinuous design which eliminates interface loading over the shoulder. The XFrame utilizes a large projection of thermoplastic distal to the laminated border than is currently used in other designs that greatly increases distal weight bearing comfort.

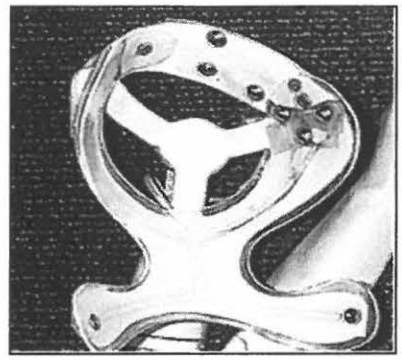
Cosmesis

Encapsulatory interfaces by nature are fairly cosmetic in the static anatomical position, yet become readily observable in the dynamic phase, where even subtle body motion typically results in frame displacements of large magnitude away from the body. The discontinuous feature of the XFrame inherently prevents vertical displacement above the shoulder during these movements.

Stability

Traditional interface stability is nebulous due to its significant coverage of a dynamically changing thorax and shoulder complex. Actions and reactions that occur within the frame secondary to gross body movement tend to leverage the socket off the body in a multitude of locations. This occurs because of the varying shape of the cross-sectional area defined by shifting contact points between the anatomy and the socket. In the newer frame-style interfaces, full volume containment is neither achieved nor desired, and as such stability is increased during gross body movement.

The XFrame is inherently more stable than other interface and frame designs in large part due to its geometry. Contact points provide rotational resistance in all planes, while the absence of a rigid transitional or suspending member over the trapezius prevents leveraging of the system off the body in response to gross body movements. Additionally, both superior and distal AP compression aids in restricting erratic motion of the frame in response to applied force.



Medial view of XFrame with neoprene electrode membrane and shoulder saddle suspension

Suspension

As mentioned previously, suspension in both encapsulatory and most frame-style designs is primarily achieved via a rigid socket or frame member extended over the shoulder complex or the trapezius. In the inherently unstable encapsulatory design, it tends to be its Achilles' heel in regards to comfort. In addition to a soft shoulder saddle, three additional suspensory elements are utilized in the XFrame when possible: 1) superior AP compression similar to that that utilized in the newer humeral interfaces described previously; 2) distal "hydrostatic" suspension over soft tissues (this also is utilized in some other frame-style interfaces); and 3) an additional scapulospinal element involving compression of the supraspinatus and the distal portion of the middle fibers of the trapezius.

Range of motion

It is important to discern between total range of motion, prosthetic range of motion and functional range of motion. A complete discussion of this is outside the scope of this paper. Inherently unstable designs may have excellent total range but are too unstable at the extremes to provide adequate function. The XFrame is stable at any and all ranges and therefore functional as well. The XFrame is the culmination of a long line of high-level frame designs that have been utilized for individuals requiring the use of a prosthesis. It was developed as a direct response to significant weaknesses inherent in encapsulatory and other interface and frame styles, hence it borrowed from their basic design elements and from many of the individuals over the years whose work has preceded it. In clinical observations, the XFrame provides greater comfort, cosmesis, stability, and suspension while maintaining a smaller footprint and hence covering less surface area than more traditional designs.

RESULTS OF A STUDY WITH THE GOAL TO OPTIMISE HARNESSES BY MEANS OF UP-TO-DATE BIOMECHANICAL ENGINEERING

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ABSTRACT

Body harnesses control body-powered and hybrid prostheses of the upper extremities. They transfer muscular forces from the amputee through motion of the shoulder girdle or stump directly to the artificial limb. The development of body harnesses requires not only to look at technical possibilities, but even more important, to take the biomechanical abilities of the patient into consideration. Different tests were performed to identify typical motion patterns of the patients and to measure the forces produced thereby.

Founded on the results of this biomechanical study and the comparison between different harnesses, a new above elbow body harness has been developed. It was ergonomically optimised to provide a more efficient force transmission. Modern materials and easily replaceable axillar pads complete the advantages and support today's hygienic aspects. There are new accessories available allowing easy and fast adaptation of the harness to patient and prostheses without any need for sewing.

INTRODUCTION

Body powered and hybrid prostheses use the patient's own body power to control functions of the artificial limb. Therefore the harness is an important component in this kind of prosthesis. The harness takes motion and forces from the residual limb, shoulder girdle and trunk and transfers them directly to the prosthesis. Natural muscle motion helps the patient perform his activities smoothly. Furthermore, the pressure of the harness on the body gives the patient sensory feedback. In cases of above elbow amputations, the harness provides supplementary fixation of the prosthetic socket to the residual limb. The higher the amputation level, the more difficult it becomes to control and fix the prosthesis. Nevertheless, it is possible to also fit patients with shoulder-level amputations. Besides the gripping process, the harness can be used to control elbow flexion and locking. In cases of below elbow fittings, the harness activates only the terminal device. There is a variety of harness systems for different amputation levels.

For above elbow prostheses two very different harness systems have been proven in practice. In Central Europe, the *triple-pull harness* after Kuhn and Hepp is preferred. Using this harness, all three required functions (i.e. activating the terminal device, below elbow flexion and locking or releasing the elbow joint) can be performed individually. On the American market *double-control harnesses* are usually fitted. This system is different from triple-pull harnesses because prehensile control and elbow flexion are activated with one cable and controlled with one motion. Therefore, when the elbow is flexed, the path of this harness required for opening the terminal device is longer. Because prehensile function as well as flexion of the elbow depend on whether the elbow is locked or released, an additional motion always has to be performed between these actions. These harness systems are also differentiated by the fact that they are controlled by different motions.

METHODS AND SURVEY

This survey was done to find out with which harness system the patient can be fitted most effectively. An important part of the survey deals with measuring the mobility of the shoulder joint using the shoulder joint simulator IKARUS, from the company Biofeedback Motor Control GmbH in Leipzig, Germany. This device represents a three-dimensional measuring and motion simulation system for testing and training the biofeedback of shoulder and joint. Through use of the shoulder simulator the mobility of the shoulder joints and force activation at any measuring point of the anatomic plane can be shown. The tests were performed with and without above elbow socket according to the neutral-zero method.

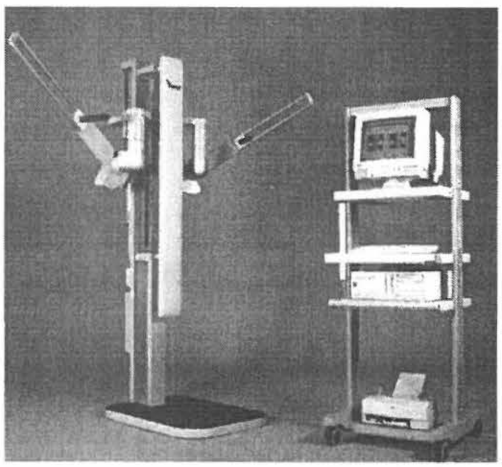

Figure 1: *Shoulder joint simulator IKARUS*

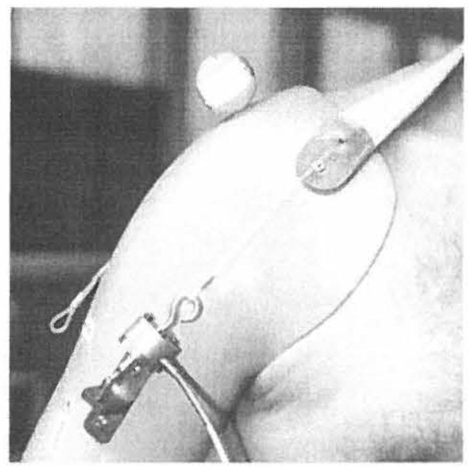

Figure 2: *Motion analysis*

To be able to assess the motion patterns with fitted body harness more detailed, measurements of the upper body were carried out in a motion analysis laboratory. For this purpose reflectors were attached to prominent points of the body and the prosthesis of a test patient wearing a prosthetic socket with different body harnesses. Using the optical-electronic measuring system PRIMAS, all typical motion patterns were recorded. At the same time, the body harness was equipped with a compression-tension sensor. In addition, the produced forces enabled the evaluation of motion efficiency.

Furthermore, it is important for optimal function of the harness to make most effective use of movements. The prosthetist has to assess how to fit the harness most appropriate for the patient. For this reason, measurements were taken describing the maximum possible paths for different motion patterns of several harnesses.

RESULTS

The results clearly show that there is sufficient force to operate the different cable-activated prostheses. Concerning the required paths, however, there are significant problems which become obvious when comparing different harnesses. Harnesses with ring have the problem that the ring slips upwards after a few movements. The tests with such a harness were performed at first with the ring placed at optimal position and then with the ring placed in the neck region.

Looking at the paths there are clear differences. Therefore, it is of high importance to adjust the cable-activated components of a prosthesis most effectively for the patient.

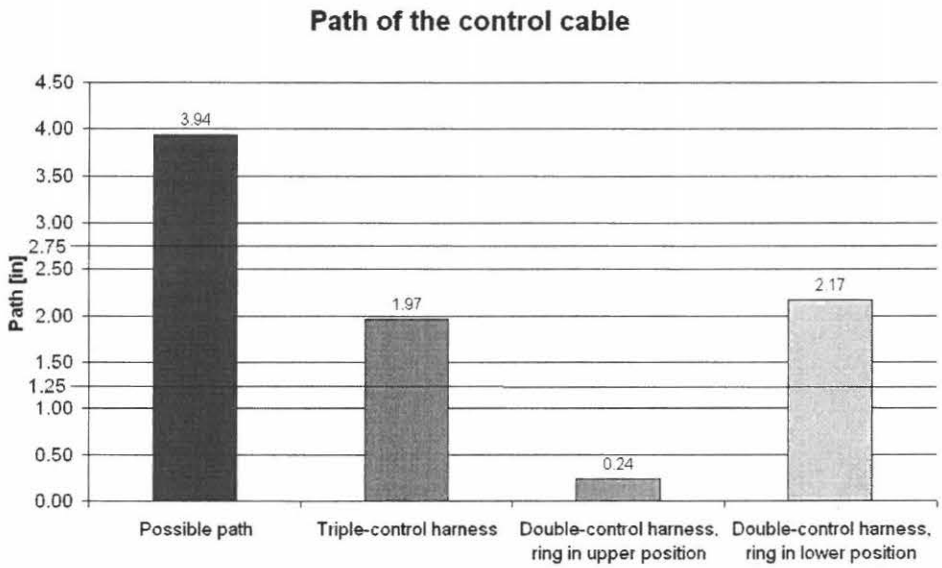

Figure 3: *Path of the control cable of different harnesses*

NEW HARNESS

Founded on the results of the study a new triple-pull harness was developed. It allows to perform each single function of the prosthesis individually. The cables for suspension and elastic straps are flexible and run from the harness ring to the prosthetic socket. Inner perlon cables transfer the forces from the patient to the prosthesis. These perlon cables are connected with the harness ring.

Common body harnesses with ring show the problem that the ring slips upwards on the back of the patient after few motions. The optimised ring is furnished with integrated cable guides preventing the perlon cables from moving on the ring. They block each other and thus avoid slipping of the whole harness system. The straps always remain in optimal position. The path and the forces of the patient are transferred to the prosthesis with high efficiency. The patient is able to control his body powered prosthesis by unobtrusive motions. In addition, optimised harness fit offers increased wearing comfort. Replaceable and washable axillar pads as well as washable harness straps support hygienic needs. A new strap connector allows to produce the harness without need for sewing.

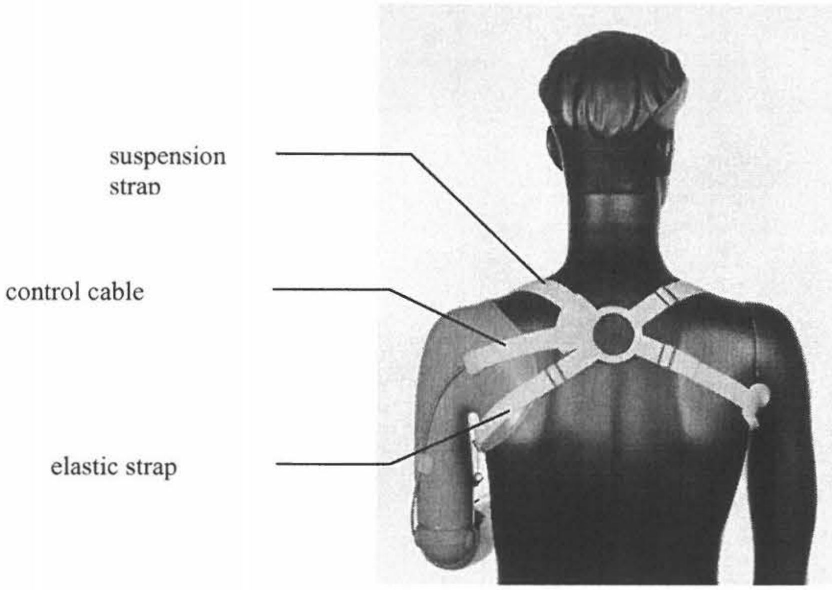

Figure 4: *New Harness*

CONCLUSION

Because of the individual needs of the patients, there is a variety of body harness systems which can be cut-to-size. But normally, ready-made harness systems are used. The triple-pull harness and the American double-control harness represent basic models for fitting above elbow amputees. It is highly important that the harness is carefully fitted to the patient and meets his requirements. Sensory feedback of performed activities and effective control of functions depend on an optimal fit of the body harness.

The new harness system was developed for the patient to make operation with the artificial limb easier.

REFERENCES

1. Bertels, Th., „Funktionen einer Kraftzugbandage für Prothesen der oberen Extremität", Med. Orth. Tech. 121 (2001) 13-17, Gentner Verlag Stuttgart

PROGRESS TOWARDS A BIOMIMETIC PROSTHETIC ARM

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ABSTRACT

Detailed design of a prototype prosthetic limb based on biomimetic principles has been completed. This paper will update the progress that has been made toward the creation of a new, high-performance limb. The limb uses antagonistic actuators with low and variable stiffness to create dynamics and interaction properties similar to those of a healthy arm. Theoretical examination of the mechanical design has yielded several interesting results and an accurate estimate of the performance and improved efficiency levels of the limb. The results have been used to understand several fundamental issues regarding the design of such a limb. Prototype construction is underway and reflects overcoming several design challenges by careful use of standard components.

BACKGROUND

Biomimetic design of a prosthetic limb involves the use of natural limb properties to inspire the configuration and design of the prosthesis. One of the major long term hypotheses to be tested with this design approach is that both ease of use and ease of training will be greatly improved due to the inherent similarities between a biomimetic prosthesis and a natural limb. The basic concept [1,2] behind this design is to use to antagonistic, low and variable stiffness actuators in the prosthetic limb (See Figure 1). Once built, the limb is to be controlled using myoelectric inputs from two antagonistic muscles, nominally, the biceps and triceps muscles. However, many challenges concerning the mechanical design of the limb have been discovered. This paper summarizes a number of these challenges and their solution.

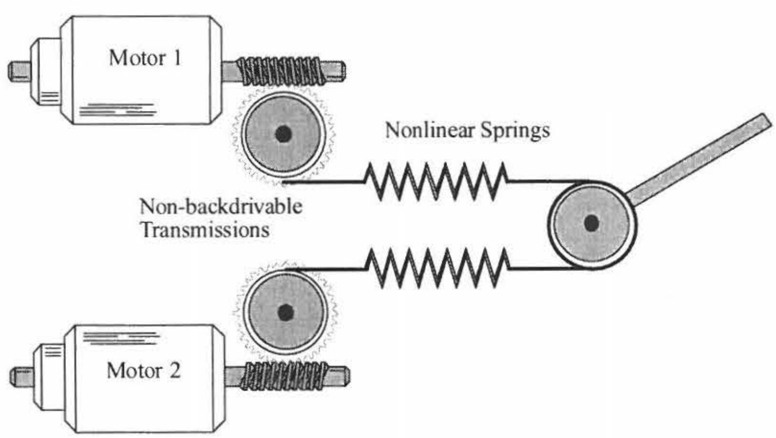

***Figure 1:** Limb Configuration*

Basic Function

Before describing the details of the design issues, a basic description of the operation of the limb is in order. To flex the limb, both motors move so that the limb rotates and that the lengths of the springs remain unchanged. The springs allow the limb to absorb impacts. Static loads can be supported without energy input since when the motors are not active the gears cannot turn (non-backdrivable). Any load applied to the limb in this situation causes the limb to move from its equilibrium position by an amount dependent on the total joint stiffness. To change the net stiffness of the limb, the actuators both turn in a direction that lengthens or shortens both springs without changing the equilibrium position of the limb.

ACTUATOR SYSTEM DESIGN ISSUES

Motor/ Gear Selection

As this design approach makes use of a self-locking or non-backdrivable transmission, new techniques are required to understand and optimize the efficiency of the motor and transmission. A graphical technique has been developed [3] that not only allows a clear description of the motor dynamics when the transmission is locked or unlocked but also clarifies the transition between these two states.

The main limitation on the selection of the motor is shown on the figure as the Maximum Thermal Current Range. Also clearly shown is the operating condition of the motor, that is, whether it is performing as a motor (delivering power) or as a generator (absorbing power).

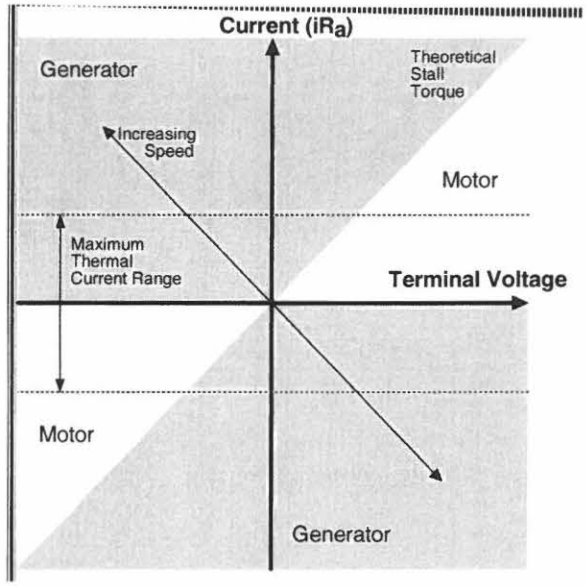

***Figure 2:** Motor / Transmission Selection*

Efficiency

In previous work [3] rough estimates of improvements in efficiency that result from allowing the amputee to choose a stiffness that allows improved interaction with constraints were found to approximately 20 fold in extreme cases. In recent work [4,5] a figure of merit has been developed to clarify certain tradeoffs that appear when a non-backdrivable transmission is used. The principle tradeoff arises from the fact that non-backdrivability arises from increased friction in the gear train. This means that there are more losses during movement despite the improvements in supporting a static load. The figure of merit (Equation 1) is the ratio of the time spent in performing static tasks to the time spent in dynamic movements.

$$\frac{\Delta t_s}{\Delta t_k} = \frac{\omega_w K_t^2 g_h^2 G e_w\big|^2_{Static}}{\tau_G R_a}\left(\frac{1}{e_w} - 1\right) \tag{1}$$

In this equation, R_a is the resistance in the moor armature, $\square_G$ is the torque required to support the load, K_t is the motor torque constant, g_h is the motor gearhead ratio, G is the gear ratio of the worm and gear set, $e_w|_{static}$ is the static efficiency of the gear set and defined based on the gear tooth geometry and the coefficient of static friction between the gear teeth, e_w is the dynamic efficiency of the gear set and w_w is the speed of the worm gear.

For the design under consideration and typical working conditions, $\square_W \sim 275$ rad/s, $K_t = 0.0075$ Nm/A, $g_h = 5.75$, $G = 20$, $R_a = 0.19$ □e□|$_{Static}$ = 0.4193 and e□= 0.4911, this ratio is

$$\frac{\Delta t_s}{\Delta t_k} = 1.93 \qquad (2)$$

Therefore in this analysis, if the arm spends 1.93 times more time in static load support than in motion, the non-backdrivable approach, and other energetic issues are not considered (such as the ability to efficiently interact with constraints) the non-backdrivable approach should be used.

CONTROLLER ISSUES

Stiffness Description

The ability to specify the stiffness of the limb assumes that the behaviour of the stiffness is well understood. [6,7] In a biomimetic design the input signals are assumed to be as they would have been for a healthy limb. Figure 3 shows a schematic, simplified representation of the stiffnesses present in such a limb. We have developed techniques to assess the impact of the missing two-joint muscles in that are missing after an above elbow amputation. Techniques for describing the limitations due to this absence are developed and various possibilities of compensating for the missing stiffness are being investigated. One important impact of the missing stiffness is the change in stiffness that occurs during interaction with a constraint.

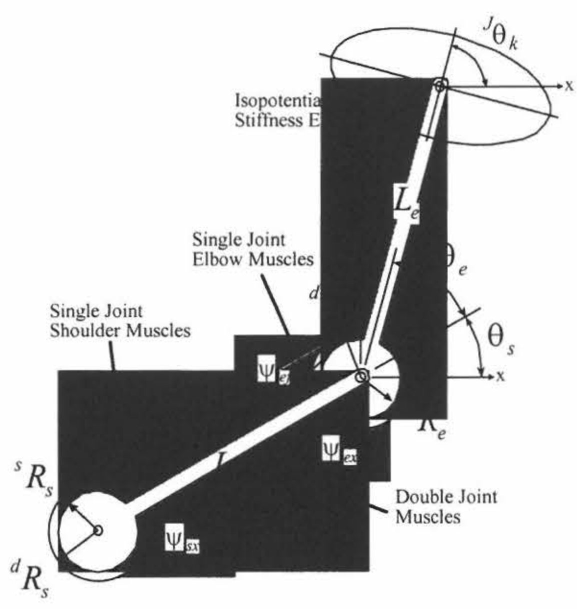

Figure 3: Stiffness Relationships

CONCLUSIONS – REMAINING WORK

A prototype is being designed and constructed based on the techniques and approaches outlined in this paper. Several control schemes are under consideration and on completion of the prototype it will be possible to assess the impact of this design on issues such as ease of interaction, the benefits of controlling stiffness and simplified training and ease of use requirements.

REFERENCES

1. C. English and D. Russell, *Mechanics and Stiffness Limitations of a Variable Stiffness Actuator for Use in Prosthetic Limbs*, in Mechanism and Machine Theory, 1999 Jan. Vol.34 No.1 Pages 7 – 25
2. C. English, *Mechanics and Design of a Variable Stiffness Actuator for Prosthetic Limbs*, MastersThesis, Carleton University, Ottawa, Canada, 1986.
3. D. L. Russell, *Energy Expenditure in Upper Limb Prostheses Performing Constrained Movements*, in Proceedings of the Myo-Electric Control Symposium, pgs 98-104, 1993.
4. T. Burke, *Modelling and Evaluation of Nonbackdrivable Transmissions for Variable Stiffness Prostheses*, Masters Thesis, Carleton University, Ottawa, Canada, 2000.
5. T. Burke and D. L. Russell, *Benefits of Using Passive Stiffness Elements in Interacting Prosthetic Limbs - A Figure of Merit* , in Proceedings of CMBEC 26, October 2000, 2 pages.
6. C. English, *Stiffness Behaviour in Two Degree of Freedom Mechanisms*, Ph.D. Thesis, Carleton University, Ottawa, Canada, 2000.
7. C. English and D. Russell, *The Influence of Double Joint Muscles on Interactive Stiffness*, in proceedings of CMBEC 25, June 1999, pgs 98-99.

EMG-ELECTODES FOR MYOPROSTHETICS: DESIGN PRINCIPLES AND APPLICATION TIPS

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SUMMARY

The myoelectrodes for the detection of myoelectric activity play a paramount role in myoelectric upper extremity prosthetics. This presentation explains the procedures which should be employed for the optimal use of myoelectrodes and introduces a new myoelectrode.

INTRODUCTION

Myoelectric control of externally-powered prostheses has proven to be state-of-the-art technology in upper extremity fitting. In contrast to body-powered prostheses, the control of myoelectric prostheses generally does not interfere with other motions, since the muscles of the residual limb can be used to the control the prosthesis. In addition, the maximum grip-force is only limited by the external power source. Myoelectric (externally-powered) prostheses are more comfortable and cosmetically attractive than body-powered ones. Another positive note is that reduced phantom limb pain has also been reported [1].

The proportional control scheme improved the rehabilitation with myoelectric prostheses using the strength of the muscle signal for precise control of grip-speed and grip-force [3]. In particular, proportional control makes high demands on the quality and stability of the control signals.

With all the positive attributes of myoelectric control, myoelectrodes for the acquisition of the myoelectric signal are the "weak link" in myoelectric fittings. On the one hand, they reduce the variety of muscle signals in their pickup area to a single control signal and therefore limit the possibility of multi-degree of freedom control of prosthetic hands or arms. On the other hand, they are very sensitive and therefore prone to disturbances. The performance of the myoelectrodes is essential for patient satisfaction.

The basic function of myoelectrodes is the reliable acquisition of the myoelectric signal and the rejection of interference. In this presentation we will disclose the optimal use of myoelectrodes while also introducing a new myoelectrode which is currently in development.

TECHNICAL PRINCIPLES OF ELECTROMYOGRAPHY

The myoelectric signal is a very low-level signal with high resistance and therefore very sensitive to disturbances [2].

Disturbances

The disturbances can be classified in the following two groups:

<u>Low frequency disturbances</u>

In definition low frequency disturbances reflect disturbances in the frequency range of the myoelectric signal. They originate from motion artefacts and electrical interference e.g. from power lines. As they are in the same frequency range as the myoelectric signal they cannot be filtered since that would also eliminate the myoelectric signal. They can only be suppressed by the input characteristics of the amplifier (as explained below) or by notch filters which eliminate just a small band of frequency but also cause a loss of myoelectric signal-power.

High frequency disturbances

High frequency disturbances have recently become more problematic by the ever increasing number of mobile phones and other devices. They can be suppressed by low-pass filtering which cuts off the high frequency band. Additional shielding of the myoelectrodes may be necessary.

Amplifier technology

Ranging from 5 to 300 μV_{RMS} the EMG is a very low level signal superimposed by disturbances that are up to 100,000 times stronger. The need for high amplification as well as simultaneous rejection of interference demands sophisticated electronic amplifier technology. The differential amplifier concept helps to reject interference while it amplifies the myoelectric signal.

Low frequency interference can be regarded as a constant potential in the pickup area of the myoelectrode (common mode interference – n in figure 1). The amplifier eliminates this constant potential by subtracting one input from the other. In contrast, the myoelectric signal produces different potentials (m1 and m2) that are superimposed on the constant interference potential. These differences remain after subtraction by the amplifier. The *differential amplifier* is therefore able to separate the EMG from the common mode interference.

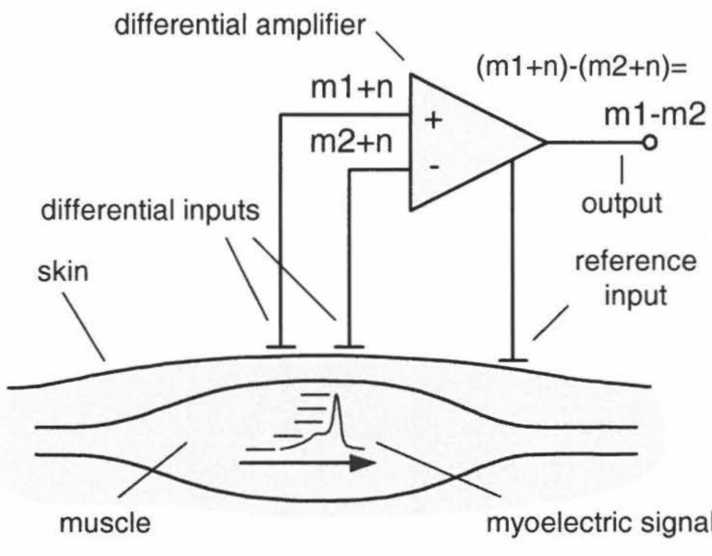

Figure 1: *Differential amplifier (schematic drawing)*

In practice some of the interference will always remain after subtraction since the two differential inputs are never perfectly equal. To reject interference that is up to 100.000 times stronger than the myoelectric signal, the gains of the two differential inputs must not differ by 1/100.000th. The ability of a differential amplifier to reject common mode interference is called common mode rejection ratio (CMRR).

Skin electrode interface

Unfortunately the CMRR is also influenced by asymmetries in skin resistance of the differential inputs, due to the fact that the two input signals are affected by the skin resistance.

This is because the signals are divided by the ratio between skin interface resistance and the input resistance of the amplifier. The effect of the skin interface resistance lessens with higher input resistances. Therefore a high CMRR and a high input resistance is required.

One should be aware that while exhausting technical possibilities of high input resistances, the skin interface will always have a negative effect. For this reason it is vital to keep a balanced resistance of the skin interfaces at the two differential inputs.

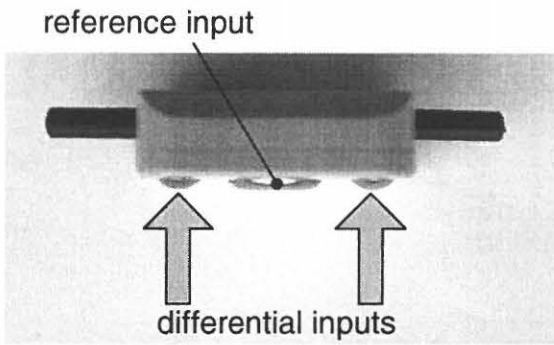

Figure 2: *Inputs of a myoelectrode*

That can be achieved by:
- equal pressure on the two differential inputs and
- a skin patch with uniform characteristics.

Therefore the myoelectrodes have to be mounted flat onto the skin and the differential inputs must avoid any scarred areas or amputation flaps as much as possible.

Gain adjustment

Another important detail regarding the application of the myoelectrode is the gain adjustment. It should be adjusted so that the patient can maintain the signal for approx. two seconds over the maximum control value of his prosthetic device.

Setting the gain too high will result in three drawbacks:
1. a loss in control-efficiency as only a small part of the signals' dynamic range is actually used for controlling
2. It causes higher disturbance susceptibility when the disturbances are over-amplified and
3. The training-effect of the muscles in the residual limb is also reduced.

RECOMMENDATIONS FOR FITTINGS

For the placement of the electrode one has to find *suitable muscles* for the control of the myoelectric prosthesis.

The muscles should:
- produce a strong signal, i.e. should be big enough and located just beneath the skins surface,
- be voluntarily contractable by the patient,
- be selected so as not to be influenced by myoelectric signals from other motions,

- be used as per their original intention and
- be located in an area of the residual limb that offers good contact with the myoelectrodes.
- If two or more myoelectrodes are used, the involved muscles should be apart far enough to be able to produce distinguishable signals.

In order to find a good *myoelectrode position*, the use of an EMG monitor is recommended. The myoelectrode should remain in one position for a longer period of time until its contact has stabilised. To create good conditions for fitting, it is helpful to clean and/or to moisten the skin prior to attaching the myoelectrodes (e.g. with saliva of the patient, disinfectant [clear Kodan spray], or Dermaclean from Otto Bock). The skin should not be cleaned with fluids like alcohol which dry out the skin too much and therefore impair conductivity.

A *well-fitted socket* avoids movements between the skin and the myoelectrodes. Loose fittings result in motion artefacts that can by far exceed the EMG signal strength. *Elastic myoelectrode suspensions* also hinder motion artefacts by making up for volume differences during movement.

Therapeutic care

To achieve the optimum of the patients' potential an experienced therapist will ensure that breaks during the process of fitting are strictly upheld. This will avoid muscle fatigue which leads to severe fading of muscle signals. It is therefore strongly advised that short sessions are spread over the whole day instead of holding a single intensive training session.

INTRODUCING A NEW MYOELECTRODE

Due to our past experiences and the necessities stated in this article we are developing a more sensitive electrode which is less disturbance susceptible due to higher CMRR and higher input resistance. We have improved on gain adjustment with a more precise logarithmic adjustment scale and have upheld patient safety by keeping the myoelectrode DC decoupled. The first results from a field study will be presented during the presentation since these results were not yet completed by submission.

CONCLUSIONS

Myoelectric fitting is state of the art for upper extremity fitting and offers many advantages in rehabilitation. To overcome existing practical needs a new electrode is being developed. Nonetheless, careful application of the myoelectrodes is imperative for achieving optimal patient satisfaction.

1. Flor H, Elbert T, Knecht S, Wienbruch C, Pantev C, Birbaumer N, Larbig W, Taub E, "Phantom-Limb pain as a perceptual correlate of cortical reorganization following arm amputation", Nature, vol. 375, no.8, Jun 1995, 482-484
2. Basmajian JV, De Luca CJ, "Muscles Alive, 5th Ed.", Williams & Wilkins, 1985
3. Kitzenmaier P, Boenick U, "Möglichkeiten der myoelektrischen Ansteuerung von Gliedmaßenprothesen", Biomed. Technik, 37 (1992), 170-180

ELECTRODES INSTALLED IN ROLL-ON SUSPENSION SLEEVES

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ABSTRACT

This presentation describes transhumeral and transradial prostheses which have been successfully fit to patients using electrodes installed in roll-on suspension sleeves. Metal electrodes and modified wiring systems have been developed and tested along with techniques for installing electrodes in roll-on sleeves to provide improved suspension and electrode contact. This series of patients has shown that a Roll-on sleeve is an excellent way to achieve superior suspension and greater range of motion. Information will be presented to document the increase in suspension and range of motion in several patients who were fit with both traditional and roll-on suspension systems. This design maintains consistent electrode contact for patients, especially those who have volume changes or experience difficulty with the traditional suspension methods. This system is adaptable to many prosthetic hand and elbow systems and amputation levels. Future research directions will be suggested to improve this system and make it available to a wider population of users.

DESIGN DEVELOPMENT

I first began experimenting with the attachment of myoelectric electrodes through silicone roll-on sleeves in 1996 with a transhumeral prosthesis. The design separated the EMG (electromyography) pick-up electrodes from the pre-amplification and control electronics. This design preserves the suspension and comfort of the roll-on sleeve and allows the EMG signal to pass through the sleeve without interrupting the suspension, as would cutting holes in the sleeve.[1] The first prototype electrodes were stainless steel bolt heads and had shielded wires leading to the preamp and arm electronics. (Figure 1)

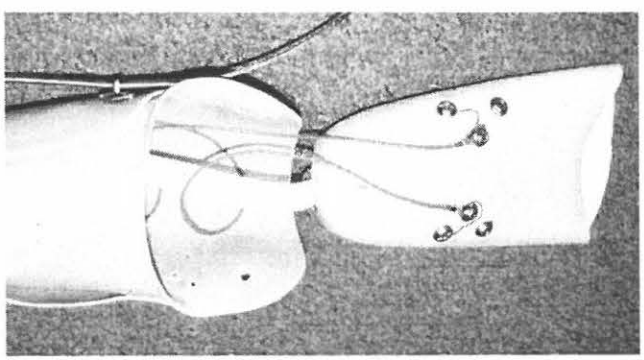

Figure 1: *Initial roll-on design*

The system worked but had severe durability problems since the wires were attached directly to the electrodes and had a plug connector in the wire harness to allow the wearer to detach the sleeve for cleaning.

Later designs used snap connectors at the electrodes with shielded EKG cables attached to the preamps. Several snap designs were tried with significant corrosion problems in the electrodes leading to degradation of the EMG signal. The current design uses custom stainless steel snap head electrodes and wires from Motion Control[2] which transfer the EMG signal through the sleeve and can be attached to many electronics packages for control of the prosthesis. (Figure 2)

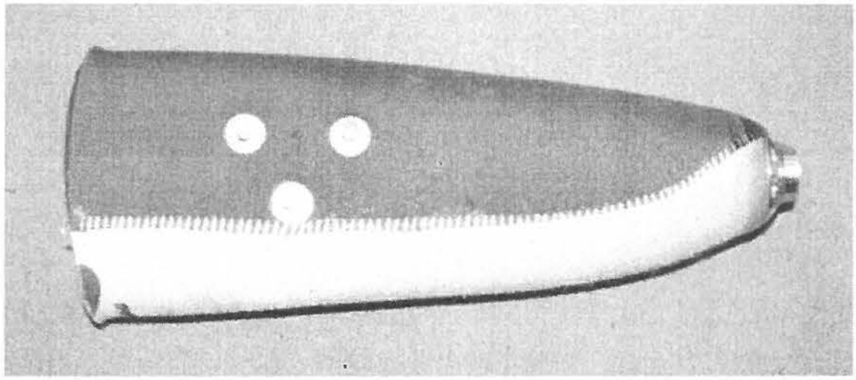

Figure 2: *Stainless steel snap electrodes in a roll-on liner*

The original three transradial cases using this design were replacements of conventional myoelectric socket designs and offered a clear comparison of the suspension and ROM (range of motion) capability of each design. (Table 1)

Table 1

Comparison of ROM and suspension	Conventional supracondular suspension		Roll-on Sleeve Suspension	
Patient	Range of Motion (degrees)	Pull-off force (Lbs)	Range of Motion (degrees)	Pull-off force (Lbs)
R.F.	5-110	30	5-125	50
C.W.	7-90	12	0-110	50
G.J.	30-110	5	5-110	37

This comparison shows a marked improvement in the range of motion and suspension capability of the roll-on sleeve design over conventional socket designs. All of the patients have also commented on improved comfort since the sleeve provides a soft interface between the arm and the hard socket. The full text of the comparison between conventional fittings and the roll-on myoelectric design can be found in the Journal of Prosthetics and Orthotics[3] or at the AAOP web site at http://www.oandp.org/jpo/library/2000_03_088.asp.
Since these first three fittings I have delivered 7 transhumeral and 9 transradial arms using this design. I have had only one transhumeral arm rejected due to discomfort from the sleeve pulling on the distal tissue causing pain.

PRACTICAL CONSIDERATIONS

The design which I have found to work best is as follows. The EMG sites are identified as usual and marked on the appropriate size roll-on sleeve. I have used Ohio Willow Wood[4] Alpha Sleeves on most of the arms since they can be heat molded to the shape of the patients arm and they resist tearing where the electrodes pierce the sleeve. A cast is taken over the sleeve with appropriate landmarks identified. The positive model is modified in the usual manner except for added build-ups over the electrode areas to prevent excessive pressure from the wires and electrodes. A clear check socket is then constructed to check the fit and alignment of the socket. This alignment is then transferred to a removable double wall socket as is customary. The wiring harness is attached to the electronics package of your choice and the wires are passed through slots cut in the socket and the preamps are secured to the socket or the inside wall of the outer socket with double stick foam tape. (Figure 3)

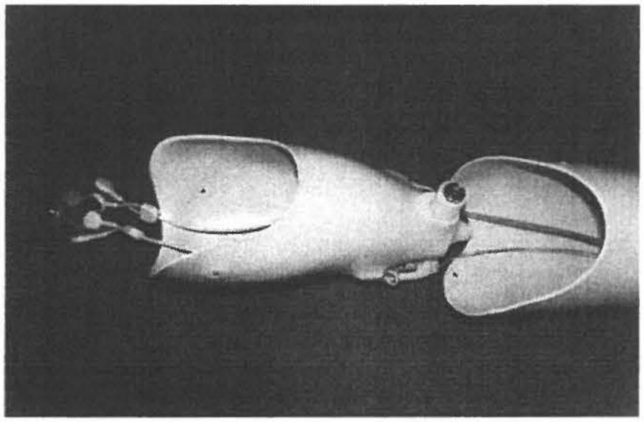

Figure 3: *complete transradial prosthesis*

The wiring after the preamps is the same as a conventional myoelectric arm. Design help can be found at http://www.utaharm.com/srfaq.htm#alt .

I have found this system to be much less sensitive to volume changes of the arm, this allows myo fittings to be made within a few months of an amputation rather than wait the conventional 6-12 months for the limb volume to stabilize. The liner also provides improved electrode contact and much greater comfort.

FUTURE DESIGN IMPROVEMENTS

The main problem with the current system is that the wiring harness is brittle and tends to fail after 3-12 months. Motion Control is working on an improved design which should eliminate this problem. In the future I would like to see the preamps either built into the liner or sealed against moisture so that they could be placed inside of the liner to eliminate the snap connectors completely.

Experimentation with a multi-electrode array seems to indicate that more than one distinct signal can be gained from a muscle group rather than the one signal per group as is used now. An array of electrodes placed in the liner would allow for several simultaneous functions in the prosthesis without the use of switching.

The system as it is currently in use has been very well accepted by patients and has significantly improved their comfort and function.

CUSTOM SILICONE LINERS FOR UPPER EXTREMITY PROSTHETICS

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ABSTRACT

This paper is intended to illustrate some new fitting techniques for the upper extremity amputee using two types of custom silicone liners manufactured at the Bloorview MacMillan Children's Centre in Toronto.

The need for custom silicone liners for upper extremity clients was discovered with the attempted fitting of off-the-shelf silicone liners. These liners were not able to provide a total contact fit. Many of the prosthetic clients at the Centre have various levels of amputation that are congenital in nature. With the wide variety of shapes (and sizes) presented by our client's residual limbs custom silicone liners were indicated over off-the-shelf liners.

Two techniques for fabricating custom silicone liners have been developed at the BMCC. The two fabricating procedures are described as custom injected and custom rolled silicone liners. There are advantages and disadvantages to both techniques. Both of these techniques offer greater accuracy in achieving total contact suction suspension for the upper extremity amputee regardless of the complexity of the limb presented.

INTRODUCTION

Techniques for suspending the upper extremity prosthesis have largely remained unchanged for years. Historically, greater resources have been utilized in improving the functional design of the upper extremity prosthesis, while less attention has been directed to improving the existing suspension techniques. Without sufficient suspension the upper extremity prosthesis will fail regardless of how sophisticated the functional design.

The preferred method of suspension by the amputee is that of a self-suspending device. The self-suspending socket, for the most part, provides the amputee with the freedom from the discomfort of a harness, but at a sacrifice in mobility and ranges in motion. The self-suspending socket design, for users with a transverse deficiency of an upper extremity, is one in which suspension is gained by encapsulating (within the socket) the joint prominences that are more proximal to the actual joint axis in the limb segment. This greatly reduces the ability of transverse rotation of the prosthesis by the amputee. It has been noted that flexion of the prosthesis can also be restricted by the more proximal borders of the socket. Additionally it is recognized that the self-suspending prosthesis is less capable of retaining adequate suspension during situations of heavy perspiration brought on by both physical exertion and elevated climatic temperatures. The temporary loss of suspension can lead to frustration by the wearer and in extreme cases, rejection of the prosthesis. With an appropriately fitted silicone liner and shuttle lock system, these complications can be eliminated.

With the popularity of roll-on-silicone-liner systems available for the lower extremity, the prosthetic community is only now making this technology available for the upper extremity amputee. Despite the numerous options the prosthetist has in selecting lower extremity silicone liners for his or her clients, there are only a few manufacturers supplying upper extremity silicone socket liner systems. Due to the complex nature in fitting the upper extremity amputee one would begin to speculate as to the success rates of fitting the client with so few options with off-the-shelf liners. The Powered Upper Extremity Prosthetics Team at BMCC has discovered few off-the-shelf upper extremity silicone liners are suitable for the types of clients presented. The conical nature of these liners is not compatible with the tapered shapes most often presented. These liners are even more problematic when the pediatric client with a congenital limb deficiency is presented. The attractiveness of a true self-suspended silicone roll-on-liner for all upper extremity clients has led to the exploration of two different techniques for providing services for fabricating custom silicone liners for the clients at BMCC.

METHODS

The two procedures employed at the Centre for fabricating custom roll-on-silicone-liners are described as a custom injected silicone liner technique (figure 1) and a custom rolled-silicone liner technique (figure 2). These techniques have been adapted to allow for the production of the custom silicone liners to be implemented in a cost-effective and timely manner.

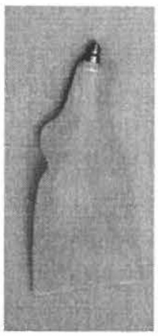

Figure 1: *A custom injected silicone liner with shuttle suspension.*

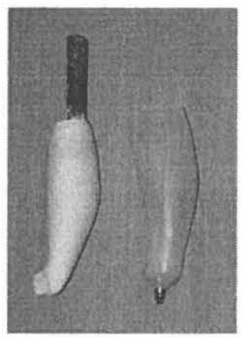

Figure 2: *A porous plaster cast and a custom rolled silicon liner with shuttle suspension.*

Both techniques rely on specific supplies and machinery and in the case of the rolled liner, specific environmental conditions, unique, to even the most modern prosthetic laboratory. The injected silicone liner technique requires a much lower initial investment in equipment and laboratory supplies. A separate dust-free laboratory with reliable environmental controls is essential to produce the custom rolled silicone liners. The machinery involved in the production of rolled silicone liners is also unique, even to the prosthetic industry. Due to the costs involved, it is difficult for the average prosthetic facility to even consider investing in the rolled silicone technology.

Clients assessed to be suitable candidates for a custom silicone liner are measured and casted for the purposes of creating a mold for a silicone suction socket liner. The limb is casted in plaster bandage or alginate, filled with porous plaster, and modified to attain the desired shape and dimensions necessary to achieve a total contact suction socket. It is suggested that the plaster model should be reduced in circumference by approximately 10 to 15 percent, with a greater reduction in circumference proximally as opposed to distally. Any sensitive or scarred areas should be marked as to prevent aggressive reduction in these areas. All measurements should be checked to assure the appropriate reductions have been made to the plaster model prior to fabrication. Liners designed with proximal trimlines which extend past the elbow joint, can be fabricated with pre-flexion to reduce any pinching of the liner or to allow for a greater range in elbow flexion.

The material used in the fabrication of a custom rolled silicone liner is defined as a High Temperature Vulcanization silicone elastomer or HTV silicone. The silicone is processed into sheet form by passing the elastomer between the two stainless steel rollers of a banbury mixer (figure 3). This technique is called Callendaring [1]. The mixer is capable of rolling the silicone into sheets of a desirable thickness. Once in sheet form the silicone is draped over the porous plaster cast. The seams of the silicone sheet are butted together, trimmed, and blended together with specialized tools to form a seamless liner. A shuttle attachment disc can be incorporated into the liner at this stage by sandwiching it in between separate layers of silicone and blending the seams with the first layer. Three wicking layers of stockinette are reflected over the liner. A PVA bag is applied over the model and placed under vacuum for 2 hours. Once the vacuum is removed the wicking layers can be discarded and the silicone can be lightly massaged in order to produce a polished surface. The liner is vulcanized at a temperature of 50 degrees Celsius for 8 hours. Once vulcanized the liner is removed from the cast and any necessary trimming is completed.

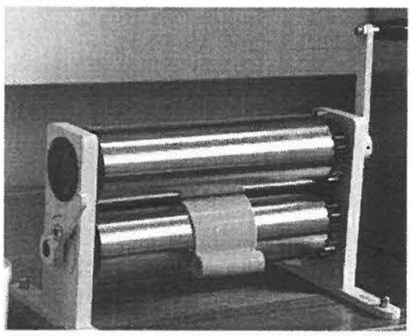

Figure 3: *A manual banbury roller mixing HTV silicone*

Fabricating an injected liner involves the use of Room Temperature Vulcanization silicone gel, or RTV silicone. The fabrication process requires a removable three part mold be designed witch contains a void cavity around the positive cast. A pneumatic injection dispenser forces the silicone gel into this void within the mold. The mold is constructed such that the void area represents the appropriate shape and thickness of the liner after removal from the mold. The void is usually 2 millimeters thick for an upper extremity liner. Thicker or thinner areas can be designed into the liner to accommodate any sensitivity the client may have. The silicone gel is allowed to cure fully before the three-part mold is separated to free the liner from the cast. It is essential that the cast and shuttle disc be secured within the mold to prevent migration. This will facilitate accurate duplication of the liner if the client requests an additional liner. Once the silicone is free from the mold the trim lines can be established and any seams buffed with a silicone-sanding wheel. To facilitate doffing and donning of the roll-on liner, a Lycra cover can be adhered to the exterior surface with a silicone-bonding agent.

RESULTS

Since the custom liner is able to accommodate any shape a true negative atmosphere exists within the liner. This results in a superior suction fit. Only with negative atmosphere is perspiration permitted to transpire through the silicone liner, eliminating loss of suspension with elevated body temperature.

The silicones used in the two techniques differ in physical properties resulting in different characteristics of the liner. A comparison of the properties of gels and elastomers, of the same durometer, reveals elastomers have greater tear resistance and tensile strength due additional cross-linking of the polymer chains [2]. These elastomers are of higher molecular weight than silicone gels. The characteristic of silicone gel is that the polymer has a greater ability to flow reducing the transmission of shear forces to the limb. A wide variety of shore hardnesses exist for both types of silicone allowing customization of the silicone properties.

Duplicating the injected liner is much less involved than producing a second rolled liner. To produce an injected liner the mold is reassembled, injected, after curing the liner is trimmed and the edges finished. An exact duplication is easily produced. To produce a second rolled liner the entire fabrication process is repeated with the exception of preparing the cast. It is much more difficult to duplicate the exact thickness and shuttle alignment with this technique. If corrections to the positive cast are required to achieve an appropriate fit, it is easily accomplished on the rolled liner cast. If errors are made in casting with the injected technique, the entire fabrication process must be repeated.

DISCUSSION/CONCLUSION

The difference in characteristics of the silicone liners in the two techniques provides the prosthetist with flexibility in tailoring the device to the needs of the client. The rolled liner is more durable but is not as effective in transmitting shear forces away from the limb as is the injected liner. The injected liner is more flexible which allows it to be rolled-on with greater ease. These fabrication techniques have proven to be both successful in providing reliable suspension as well as improving the function of the prosthetic device allowing ranges of motion which were lost with conventional suspension techniques.

REFERENCES

1. De Cubber, J., "Introduction of the JDC Manipulation Technique", The Guide, Silicone House Otto Bock Health Care. 5-6
2. The Guide, Silicone House.

EXPERIENCE WITH SILICONE SUCTION SOCKETS USI NG MYOELECTRIC CONTROL

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The use of silicone or similar material with distal attachment as a suspension system has gained great acceptance and usage for lower limb amputees. Since its introduction in the mid 1980's by Kristinsson, the silicone suction socket (3S) has undergone extensive development and the options available have proliferated due to its popularity.

Benefits for the lower limb include: reduced shear forces on the skin, better pressure distribution especially when a gel type liner is employed, and excellent suspension. Shortly after its introduction, 3S technology was applied to upper limb fittings. [2] Some early success was achieved and results were promising. Problems existed related to inappropriate sizing of liners and locking system for utilization in upper limb applications and difficulty interfacing myoelectric control in combination with 3S suspension.

One method for conducting the myoelectric signal through the roll-on-socket, which has gained success, employs the use of steel electrodes threaded to snaps with the silicone sandwiched between [1]. This method has proven very useful for fitting a variety of amputees who have experienced difficulty with conventional socket designs and myoelectric control. As recommended by Daly, the author has used the Ohio Willow Wood Alpha liner exclusively when using the snap electrodes as provided by Motion Control. Types of amputees that have most benefited include: transradial amputees with short residuums, especially those with scar tissue, and transhumeral amputees with short residuums. Electrode contact can be difficult to maintain in both groups when using electrodes fixed to a rigid or semi-rigid interface. In some of these cases, conversion to 3S interface with snap electrodes has meant the difference between marginal, intermittent control and the excellent control reliability attained with the new system.

The Alpha liner now is available with several variation that should be considered when choosing the most appropriate product for a particular application. These include sizing, color, textile backing material (standard or Spirit), and size of the distal umbrella or no distal attachment at all (locking vs cushion liners). When utilizing a distal locking mechanism there are two basic choices, either a clutch lock or a ratchet can be used. The clutch lock has been used when it is desirable to assist the limbs' entry into the socket by "pulling in" using the slotted lock release button. This lock mechanism is longer and larger than the other options and this may preclude its use. There are many ratchet locks commercially available. The ratchet mechanism now favored by the author is the Ossur Upper Extremity ratchet lock which is very low profile and light weight. The Ossur ratchet is also easily assembled, or disassembled, from the inside of the socket. Two other options are a lanyard and no distal attachment in select cases. The lanyard, (often just a Velcro strap), offers the lowest profile distal attachment, allows for "pulling in" and is therefore useful for longer residual limbs.

When donning the Alpha liner it is important to do so with the right orientation in order to best approximate the desired electrode sites. This is perhaps best done by aligning the seam of the liner with some feature of the limb such as a freckle or scar. Electrode orientation using this method has not been a problem. Another consideration is how to address the actual attachment of the snaps, they can be pushed in place either before donning the socket or after the limb is inserted. In most of cases, the snaps have been engaged prior to donning and the wires simply guided into the socket to find there own resting place. (Photo 1) Initially there was concern that the wires, snaps, or the thickened junction points would cause discomfort but this has not been the case with any of the transhumeral cases and has only presented a problem with slender and bony transradials. Therefore, only in the cases where there is insufficient soft tissue padding has the socket been fabricated to align the snaps over the electrode site to be secured after donning. (Photo 2)

Photo 1

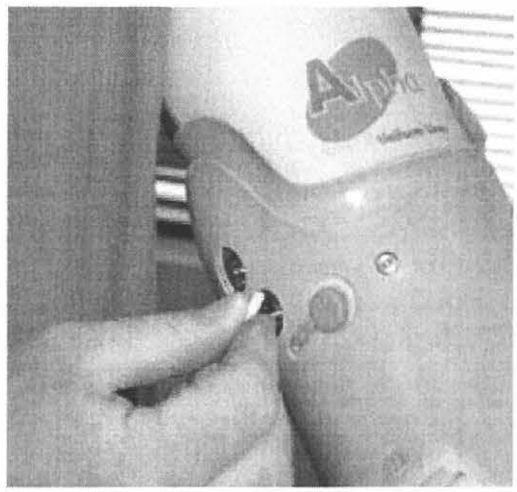

Photo 2

This experience favorably supports that of Daly and suggests that continued research and development of Roll-on-Suction-Sockets for use with myoelectric control should be pursued. The means of conducting the myoelectric signal from the limb to the pre-amps is one issue that deserves attention. Although all of the clients fitted found the new system to be superior to their previous prosthesis, the need to individually attach each electrode and the potential for failure when wires are continually manipulated is not ideal. As suggested by several parties interested in this technology, it should be possible to embed the connection in the liner and bring the signal out through the distal fixation mechanism or through one simple plug. Although the Alpha liner works very well with snap electrodes, it may be advantageous to use a liner that is thinner such as the Ossur UX liner. In any case, 3S systems with myoelectric control are a valuable tool in managing the upper limb deficient client and is a welcome addition to our fitting choices.

REFERENCES

1 Salam, Y., "The use of silicone suspension sleeves with myoelectric fittings", Journal of Prosthetics and Orthotics. 1994;6: 119-120

2 Motion Control, 2401 S. 1070 W. St B, Salt Lake City, UT 84119

3 Daly,W, "Clinical Application of Roll-on Sleeves for Myoelectrically Controled Transradial and Transhumeral Prostheses", Journal of Prosthetics and Orthotics, Vol 12: 3, 88-91

4 Ohio Willow Wood, 15441 Scioto Darby Rd., Mt Sterling, OH 43143

NOTES

August 22, 2002 (Thursday)
10:30am – 12:00pm

Theme: Occupational therapy and case studies

Invited Speaker:
10:30 – 10:50
Margaret Wise
The influence of concomitant diagnoses when treating the older UE amputee

10:50 – 11:05
Bill Limehouse
Prosthetic Management of an individual with "unique" multi-level limb deficiencies: a case study

11:05 – 11:20
Steve Mandacina
Elective amputation of Cerebral Palsy patient successfully wears electric prosthesis

11:20 – 11:35
John M. Miguelez
Electronic prosthesis for the partial hand amputee

11:35 – 11:50
Troy Farnsworth
Clinical trials of the new Boston digital arm system

11:50 – 12:05
Rinchen Dakpa
Fitting of transcarpal myoelectric prosthesis with locking liner

THE INFLUENCE OF CONCOMITANT DIAGNOSIS WHEN TREATING THE OLDER UPPER EXTREMITY AMPUTEE

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Upper Extremity Specialists, Dallas, Texas

When evaluating the older upper extremity amputee, there are often concomitant diagnoses that may influence rehabilitation goals. These conditions may either be ipsilateral or contralateral to the amputation, and significantly influence the amputee's tolerance to prosthetic fitting and training.

Decreasing strength and endurance are normal in the aging process and are usually obvious during an initial evaluation. There are other orthopedic conditions concomitant with amputation, which may be less obvious.

Some of the most common conditions affecting goals and training are (1) osteoarthritis of the thumb, (2) lateral epicondylitis, and (3) shoulder pain. A brief discussion of these conditions will explain their affect on amputee rehabilitation.

According to Alfred B. Swanson, "Almost 100% of the population is susceptible to some form of arthritis, and if an individual lives long enough, he is almost certain to develop one type or another."[1] "Approximately 8% of all adults are estimated to have moderate-to-severe clinical symptoms of osteoarthritis of the hands and feet."[2]

Osteoarthritis of the thumb (also known as CMC arthritis and basilar joint arthritis) generally starts after the age of 40 and is more common in women than men. It is a wearing away of the cartilage between the first metacarpal and the trapezoid. Symptoms occur when performing tight or sustained pinching and gripping. The thumb has a loss of bone contour and develops a "squaring of" appearance. Radiographs show degenerative changes.

Painful grip or pinch associated with CMC arthritis creates significant problems for our upper extremity amputees during a variety of functional tasks. For example, tight pinching is required while using a pull sock and donning a snug anatomical socket. Alternate donning techniques may be required.

Another problem that occurs contralateral to the amputation is lateral epicondylitis, better known as tennis elbow. Priest reports, "the problem frequently occurs between the third and fifth decade of life in the inexperienced [tennis] player."[3]

Tennis is not always the offending factor; lateral epicondylitis can be caused by many activities requiring powerful gripping and repetitive use of the wrist extensors.

Lateral epicondylitis is not well understood, but it is generally agreed that the cause of this pain is inflammation of the wrist extensor tendons and microscopic tears at the insertion of the extensor carpi radialis brevis. Conservative treatment includes anti-inflammatory medication, a counterforce brace, sometimes a static wrist splint, and rest. Clinically, it takes a long time to heal and in the more chronic cases a moderate reoccurrence rate.

As with CMC arthritis, lateral epicondylitis makes donning the prosthesis more difficult. Wrist extension involved in doffing the prosthesis also elicits pain. Alternative methods for donning and doffing the prosthesis may prove helpful in decreasing pain during these tasks.

Shoulder pain often interferes with prosthetic fit and function. The American Academy of Orthopedic Surgeons states that nearly six million people a year seek medical care for shoulder problems.[4] Pain on either the amputated side or contralateral side is equally problematic.

Shoulder pain, with its associated loss of strength and range of motion, makes it difficult to operate many prosthetic control schemes. It dictates prosthetic design, and if ignored, can cause the amputee to abandon the use of the prosthesis.

The most common shoulder challenges encountered with amputees include myofascial trigger points, acromioclavicular arthritis, subacromial impingement, and rotator cuff tears.

Janet G.Travell, M.D. defines, "Active myofascial trigger point: a focus of hyperirritability in a muscle or its fascia that is symptomatic with respect to pain; it refers a pattern of pain at rest and/or on motion that is specific for the muscle. An active trigger point is always tender, prevents full lengthening of the muscle, weakens the muscle, usually refers pain on direct compression,..."[5] In other words, trigger points are tender spots with palpable hardness that refer pain to other areas. The upper trapezius, supraspinatus, and teres minor can refer pain to the shoulder area, causing pain and limiting motion. Trigger points can become more intense during fitting, early periods of training, and after unusually hard periods of work. Therapy to treat trigger points during expedited fitting often facilitates the rehabilitation process.

For people over the age of 50, a common occurrence is acromioclavicular arthritis. This is a degenerative condition that destroys the cartilage between the clavicle and acromion, causing the joint space to narrow. Patients with acromioclavicular pathology have point tenderness over the joint and decreased range of motion, especially in horizontal adduction, and rotation.

And finally, the rotator cuff is another cause of pain and loss of function. The rotator cuff is comprised of four tendons and muscles that surround the head of the humerus and work together to stabilize the glenohumeral joint and allow smooth motion.

The two most frequently noted conditions in the rotator cuff are subacromial impingement and rotator cuff tear. Subacromial impingement is a narrowing of the subacromial space; it is caused by a variety of clinical entities. This narrowing causes compression of the supraspinatus tendon, resulting in pain and loss of shoulder motion. The second problem is rotator cuff tears. While rotator cuff tears can occur with a single event, more often in people over 40, it is a gradual attritional wear of the rotator cuff tendons by impingement. This gradual wear may ultimately result in a full-thickness tear. It is important to note that not all rotator cuff tears require surgery; not all rotator cuff tears can be repaired by surgery.

Patients with impingement or tears of the rotator cuff present with pain. The pain is worse when the arm is dangling at the side, when sitting unsupported, or when trying to sleep. They have loss of motion especially in horizontal adduction, external rotation, and abduction. In severe cases, independent performance of activities of daily living or even self-care is impossible.

Therapy can help. For example, heat modalities, trigger point massage, and stretching help decrease myofascial trigger points. Strengthening posterior shoulder girdle musculature often decreases pain associated with subacromial impingement and rotator cuff tears. And simple things, such as making sure a patient is seated in a chair with arms or providing a pillow for arm support, can increase patient comfort and extend the amount of time the prosthetist can work with the patient. Training in alternate methods of donning and doffing the prosthesis as well as educating the patient in good ergonomic principles for activities of daily living will also improve the amputee's ability to use the prosthesis more comfortably for daily tasks.

Concomitant orthopedic conditions common in the older adult significantly influence our treatment of upper extremity amputees. We should consider the possibility of these problems when evaluating and setting goals for the older amputee. Pain in the hand, elbow, and shoulder can affect fitting, tolerance to the prosthesis, and functional use. With a thorough evaluation, appropriate goal setting and treatment intervention, we can help all of our amputees increase their rehab potential and become as independent as possible.

REFERENCES

1. James Hunter, Evelyn Mackin, and Anne Callahan, Rehabilitation Of The Hand: Surgery And Therapy, Fourth Edition (St. Louis: Mosby-Year Book, Inc., 1995), II, 1307.
2. James Hunter, Evelyn Mackin, and Anne Callahan, Rehabilitation Of The Hand: Surgery And Therapy, Fourth Edition (St. Louis: Mosby-Year Book, Inc., 1995), II, 1308.
3. James Hunter, Evelyn Mackin, and Anne Callahan, Rehabilitation Of The Hand: Surgery And Therapy, Fourth Edition (St. Louis: Mosby-Year Book, Inc., 1995), II, 1816.
4. American Association of Orthopedic Surgeons. March 2000 American Academy of Orthopedic Surgeons. 23 June 2002 <http://www.aaos.org./>
5. Janet Travell, David Simons, Myofascial Pain and Dysfunction The Trigger Point Manual, I (Baltimore: Williams & Wilkins, 1983), p.1.

PROSTHETIC MANAGEMENT OF AN INDIVIUAL WITH "UNIQUE" MULTI-LEVEL LIMB DIFICIENCIES; A CASE STUDY

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ABSTRACT

A 23-year old worker presented with multi-level amputations of all extremities following electrical injury, complicated by burns and skin grafts to over 45% of his body. Amputations of both legs with bilateral partial hemipelvectomies were required along with disarticulation of the right upper extremity at the shoulder and amputation of the left index finger.

After wound healing was complete, the patient was fitted with a hinged clamshell prosthetic socket to attain vertical sitting posture that allowed his body weight to be born over his thorax in pressure tolerant areas of his torso. Urethane gel padding was used to line the socket, and a cold water cooling system was designed and incorporated into the prosthetic socket. Next, a removable myoelectric shoulder disarticulation prosthesis was designed with dual attachments for the thoracic socket and the bed frame. This attachment feature removed the weight of the prosthesis from the patient's pressure sensitive areas and enabled the patient to engage in bimanual activities whether upright in his wheelchair or supine in bed. Removable long-term adhesive electrodes were used to obtain control through the pectoralis and infraspinatus muscle groups.

A Boston 3 electronic elbow with 4 motor controls was provided. Control options for the prosthesis are hand (open-close), wrist (supination-pronation), elbow (flexion-extension), and a power locking shoulder joint (flexion). Humeral internal and external rotation will be added in the future. Detachable lower extremity prostheses have been provided to improve the cosmetic appearance. This complex prosthetic system has provided restoration of functional abilities and overall quality of life for this individual.

Specifics as to why certain componetry was used will be discussed.

FINAL PAPER NOT RECEIVED AT PRESS TIME

ELECTIVE AMPUTATION OF CEREBRAL PALSY PATIENT SUCCESSFULLY WEARS ELECTRIC PROSTHESIS

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Cerebral Palsy affects 15-20 new borns in every 10,000. Currently, near a half million people in the U.S. are effected with C.P. Diagnosing the condition is done clinically, with lab test only ruling out other diseases. The tests performed, such as manual muscle testing, ROM, physical and emotional development, are all compared to normal childhood outcomes. Therefore, many infants are not diagnosed with C.P. until the age of two or three. Symptoms are normally weak or tight muscles, poor balance and gait, along with seizures in approximately half of the effected people. Over time, weak muscles can often develop severe joint contractures, classified as spastic Cerebral Palsy. (1,2,3)

In the fall of 2000, 48-year-old Mr. Cline presented with spastic hemiplegic Cerebral Palsy affecting the arm and leg of the left side. He was also diagnosed with Polio at the age of four, but the Medical team now believes that his limb condition is secondary to C.P. Mr. Cline showed some difficulty ambulating, secondary to poor musculature in his left leg. His left arm was in 60° flexion contracture, 80° abduction contracture, forearm at 135° flexion contracture, with wrist, hand and fingers all flexed; All conditions present since four years of age. ROM at gleno-humeral was passive only, with full active scapular-thoracic motions. Flexion and extension of the elbow along with the wrist and hand were passive. The elbow only allowed 20° degrees of extension from the flexed position. EMG testing showed 100^+ microvolts of the triceps and less than five microvolts of the biceps. No EMG signals were found distal to the elbow. His right arm and right leg were not effected and normal with full ROM and 5/5 manual muscle test.

Mr. Cline is employed as a delivery driver for an auto parts store. Duties include packaging and lifting heavy boxes and transporting them to car dealerships and other parts stores. In his spare time he enjoys fishing and yard work, along with daily chores and activities.

Mr. Cline presented with extreme frustration performing his daily activities with respect to the condition of his left arm. He believed he could perform better at work if the arm was not an obstacle getting in the way. Mr. Cline and the surgery staff decided to amputate his arm even if the prosthesis would be of no benefit. The rehab team consisting of an orthopedic surgeon, physiatrist, occupational therapist, physical therapist, and two certified prosthetists decided to not make conclusive decisions on prosthetic components until after the amputation knowing that ROM may drastically change.

Mr. Cline commenced occupational therapy three weeks prior to amputation to improve ROM and EMG signals. Duration was three times a week for 30-minute visits. Unfortunately, the only benefit was a marginal 10° of passive extension at the elbow.

In December of 2000, amputation occurred at the humerus, three inches proximal to elbow center. Aggressive measures for edema control, healing, and OT immediately took place. Two weeks after amputation we reevaluated for prosthetic components. Unfortunately, gleno-humeral ROM did not improve, nor was there any improvement of bicep EMG. The decision was made for a full electric system of elbow, wrist rotator, and terminal device. For optimal control, two myosites with adequate signals and separation were desired. However, upon evaluation the triceps always interfered with the biceps signal, and the biceps could never achieve a higher signal than the triceps. The decision was made to have a servo controlled elbow, single site control of the TD, and to delay the wrist rotator control. The single site electrode was placed on the biceps, in hopes to increase stamina and isolate the biceps-triceps antagonistic signals.

This prosthesis was fit when the limb was completely healed at four weeks post surgery. Using an expedited fitting protocol, it was fit in a temporary setup with a Boston Elbow 3 with single site hand threshold-servo elbow strategy, shoulder saddle and chest strap, flexible inner liner with rigid frame, and Otto Bock hand and Greifer. On the same day we cast the limb, Mr. Cline was able to function with the temporary setup. He wore the arm for six weeks, receiving therapy three times a week for the first four weeks, then once a week for two more visits. By this time Mr. Cline was able to, for the first time ever, do bimanual activities such as hold on to a cup while he pours, hold on to a fork while he cuts, grooming, and bimanual activities at work.

After six weeks of wearing the arm, reevaluation of the prosthetic components took place. The flexion and abduction contractures of the shoulder were still present, yet lessened to 30° flexion and 40° abduction. EMG signals of the triceps were still more than adequate at >100 microvolts and the biceps improved to near 40 microvolts. More importantly, there was only a 15 microvolt cocontraction of the triceps when the biceps fired. Mr. Cline stated difficulty at work controlling the elbow with a servo because of the body motions necessary along with inadvertent triggering. Therefore, combined with his personal preference, the elbow was switched with a Utah 2, dual site myo elbow and hand, using rate cocontraction to unlock the elbow and a switch to toggle between TD and wrist.

Currently, Mr. Cline is wearing the arm 10-12 hours a day and uses it very well. He is able to do everything he wishes with the arm and enjoys the benefits and ease of activities with bimanual grasps. He wears the arm most of the time, and the hand while at social activities or dinning. He stated he's noticed the ability to do more with his sound right side hand, possibly from a decrease of fatigue and overuse syndrome. Mr. Cline's life has completely transformed from a person with a limb disfigurement to a natural, two handed person able to do whatever he wants, and have a much better outlook on life.

REFERENCES

1. Peacock, J., Goldstein, M., "Cerebral Palsy", Capstone Press, September 1999.
2. Miller, F., Bachrach, S.J., "Cerebral Palsy: A Complete Guide for Caregiving",Johns Hopkins University Press, 1998.
3. Pincus, D., "Everything You Need to Know About Cerebral Palsy", Rosen Publishing Group, Inc., August 1999.

ELECTRONIC PROSTHESIS FOR THE PARTIAL HAND AMPUTEE

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ABSTRACT

There exists a proportionally large upper extremity patient population that presents an absence at the carpal level. Until recently, prosthetic intervention was limited to a passive restoration, a rigid opposition post, or a body-powered hook style terminal device with control harness. The carpal level amputation challenges the prosthetist because the use of most existing functional terminal devices results in a significant contralateral limb length discrepancy. A small group of carpal level amputees have been fit over the last several years with a highly modified Otto Bock electric hand that in effect decreases the overall length of the terminal device by rotating the transmission and motor assembly. These modified electric hands compromised cosmesis for function as the rotated transmission resulted in loss of the hand's natural appearance. The modification also voided the manufacturer's warranty and led to an increased component maintenance and failure. Even with these limitations, many of the patients fit with modified electric hands continued to utilize them because of the increase in grip force and the independence from using gross body movement to control the terminal device.

Responding to the challenges of the carpal level amputee population, Otto Bock has introduced an electric hand with a shortened chassis and overall length. This paper will examine the functional benefits of a production version carpal level electric hand, control scheme options, and interface design considerations

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Clinical Trials of the new Boston Digital™ Arm System

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ABSTRACT

Microprocessor-based controllers for upper-limb powered prostheses have made significant advances over the past few years. These devices allow prosthetists to evaluate the patient and set-up prosthetic controls to optimize performance for the user. This enables the user to obtain a controller that is customized to suit their specific needs and capabilities.

The new Boston Digital™ Arm System is the first powered elbow prosthesis to offer this advanced technology. This System serves as a "platform" for upper-limb prosthetic control. With five motor controllers, the Boston Digital Arm System can operate hands grippers, wrist rotators, shoulder lock actuators and more. The System is universal – it works with prosthetic devices from any manufacturer, allowing prosthetist to create the optimal prosthesis for their client.

Rather than requiring users to adapt to a pre-defined control strategy, this system is adaptable and can be set-up so that the user can select and control the prosthetic devices using motions that are relatively easy for them to do. Since it accepts input signals from many transducers such as; myoelectrodes, force-sensitive resistors, positional-servo sensors and switches, the control options are practically limitless.

This presentation will describe the results of recent field trials for the new Boston Digital™ Arm System. This clinical experience has let to improvements in the computer-generated graphical interface screens as well as better client training techniques. Control strategies will be discussed and examples of simple and complex strategies will be reviewed. Prosthetists will gain an understanding of the versatility of this system through these case reviews.

FINAL PAPER NOT RECEIVED AT PRESS TIME

FITTING OF A TRANSCARPAL MYOELECTRIC HAND WITH A LOCKING LINER

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West Park Health Care Centre, Toronto, Ontario, Canada

INTRODUCTION

This is a case presentation of the fitting of a transcarpal myoelectric hand with a locking liner for a patient with a wrist disarticulation amputation.

Fitting patients with long upper extremity amputations has been a challenge in the past due to the resulting limb length discrepancy between the prosthetic and the sound sides. However, since the introduction of the new Otto Bock 8E44 Transcarpal myoelectric hand, it is now possible to fit wrist disarticulation amputees with a locking liner for suspension.

PATIENT HISTORY

JW is a fifty year old maintenance worker in a school board in a small town in Ontario. He has a wrist disarticulation amputation as a result of a wood chipper accident in September 2000. Since then he has been fit with a standard body powered conventional prosthesis.

JW was referred to West Park Health Care Centre early in 2002 to be fitted with an externally powered prosthesis.

GUIDING PRINCIPALS

When an amputee who currently uses a prosthesis is referred to be fitted with a different prosthetic device, it is often very helpful to jot down the main improvements in function, comfort, cosmesis and any other features that are to be gained in the new device. The patient expressed the following wish list.

1. A prosthesis free of harness
2. Superior function in comparison to his conventional body powered prosthesis
3. Increased grip force
4. A prosthesis that looks like a real hand and one that is functional
5. A light weight prosthesis

Successful outcome is achieved when the gap between the patient's expectations and the availability of current technology in devices fittings are clarified. For example clarification of bullet point 2, above, was discussed in great detail since the conventional prosthesis with a 5X hook could also be more appropriate as a tool in certain situations where the myoelectric device would be inappropriate.

SOCKET DESIGN

It was important for the patient to be free of prosthetic suspension harness and he wanted to maintain normal range of motion of the elbow. Since the Otto Bock 8E44 Transcarpal hand does not allow passive pronation and supination of the wrist, it was important that the socket design maintains the forearm rotation present in the residual limb. Patient's bony distal end of stump with thin tissue coverage and low tolerance to pressure contraindicated for standard socket designs such as the expandable wall socket, partial silicone socket or socket designs with window openings. Supracondylar socket or the Munster socket would have restricted the natural pronation and supination of the forearm.

It was decided to use an Ossur Upper-X locking liner with an Icelock UX721 ratchet lock for suspension. An Ossur reusable fabrication tooling; UX 780 for the Icelock ratchet made it easy to set it up for trial fitting with a thermoplastic socket and save the ratchet lock for the fabrication of the definitive device.

POWERED ELECTRIC HAND

Due to the use of the Ossur Upper-X locking liner and the Icelock ratchet locking mechanism, the selection of the hand had to be such that there was no arm length discrepancy between the prosthetic side and the contralateral sound side. The new Otto Bock 8E44 Transcarpal digital twin hand was recommended. The benefits of this prosthetic design are:

1. Symmetry of arm lengths achieved
2. Light weight in comparison to standard wrist disarticulation myoelectric hands.
3. Acceptable cosmesis

CONCLUSION

The use of the Otto Bock Transcarpal hand and Ossur Upper-X locking liner with an Icelock ratchet made it possible to achieve the primary objectives. Since the Otto Bock Transcarpal hand does not allow passive pronation and supination, it is essential to position the lamination plate to optimize pronation and supination for best functional usage of the prosthesis.

The length of the prosthetic device from the medial epicondyle of the elbow to the thumb tip of the Transcarpal hand was equal to the contralateral sound side. Comfort and cosmesis were improved and the patient did not think the total weight of the finished prosthesis was of any concern.

NOTES

August 22, 2002 (Thursday)
1:30pm – 3:00pm and 3:30pm – 4:30pm

Theme – Hardware and controllers

Invited Speaker:
1:30 – 2:00
Dick H. Plettenburg
Prosthetic actuation: a case for pneumatics
Prosthetic control: a case for extended physiological proprioception

2:00 – 2:15
Todd R. Farrell
Real time personal computer (pc) modeling of a prosthesis controller based on the concept of extended physiological proprioception (epp)

2:15 – 2:30
Craig Wallace
Unique device-selection strategies for powered elbows

2:30 – 2:45
Bill Limehouse
Clinical experiences with animated prosthetics controller and LIMB LINK ™

2:45 – 3:00
A. Davalli
Mini joystick for upper limbs prostheses

3:00 – 3:30 Refreshment Break

3:30 – 3:45
Gary Sjonnesen
Microprocessor control features and control options

3:45 – 4:00
William J. Hanson
New prosthetic controller expands capabilities

4:00 – 4:15
David Wells
Software/firmware tools to customize controller parameters in upper extremity, powered prosthetic systems

4:15 – 4:30
Christopher Lake
Comparative analysis of microprocessors in upper limb prosthetics

PROSTHETIC ACTUATION: A CASE FOR PNEUMATICS

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ABSTRACT

Electrically actuated hand prostheses have the disadvantage of a high prosthetic mass, a slow cycle time, vulnerability, and an excessive volume. Pneumatical actuation can overcome these disadvantages. To demonstrate the feasibility of pneumatic actuation a pneumatically powered hand prosthesis has been developed. A careful assessment of the system choice, the friction losses, the dead spaces, and the supply pressure level resulted in a low gas consumption, enabling the use of small disposable gas containers. The mass of the hand mechanism is 60 grams, the operating cycle takes less than one second, the hand size is comparable to the hand of a 2.5 – 4 year old child, and the prototype functioned well in the laboratory for over 75000 cycles. These results show that pneumatic actuation of hand prostheses excels electrical actuation.

INTRODUCTION

The myoelectrically controlled, electrically powered hand prosthesis for children has become one of the standard prosthetic devices for children with a unilateral below elbow defect. This type of prosthesis is very well accepted by the children and their parents because of its outward appearance and because of the absence of a control harness. Despite these advantages and the success of fitting children with this myoelectric hand prosthesis, it has several known disadvantages: the prosthetic weight is high, the operating speed is low, the system is vulnerable and its size prohibits fitting it to children with long forearm remnants.

METHODS

Pneumatic power can overcome most of these disadvantages. A pneumatic motor is light in weight, fast, reliable and it can be small [1]. A pneumatically powered hand prosthesis has been developed in an attempt to master the disadvantages of electrically powered hand prostheses [2]. Emphasis is on minimal gas consumption, as a pneumatically powered hand prosthesis is only feasible when it can be powered from a single gas container throughout a whole day. In order to minimise the gas consumption the hand operates in a bi-phasic fashion, i.e. the operating cycle of the hand is split into a prehension phase and a pinching phase. In the prehension phase the hand can be opened and closed. As soon as the thumb of the hand touches an object the mechanism is automatically switched to the pinching phase. In this pinching phase a force is exerted between the fingers and the thumb. To resist the resist the reaction forces a locking mechanism is provided. Operation of this system is as follows, taking the rest position of Figure 1 as a start. In this rest position the hand is closed against itself or against an object. A pinching force is exerted between the fingers and the thumb by a spring. All pneumatic motors are at atmospheric pressure.

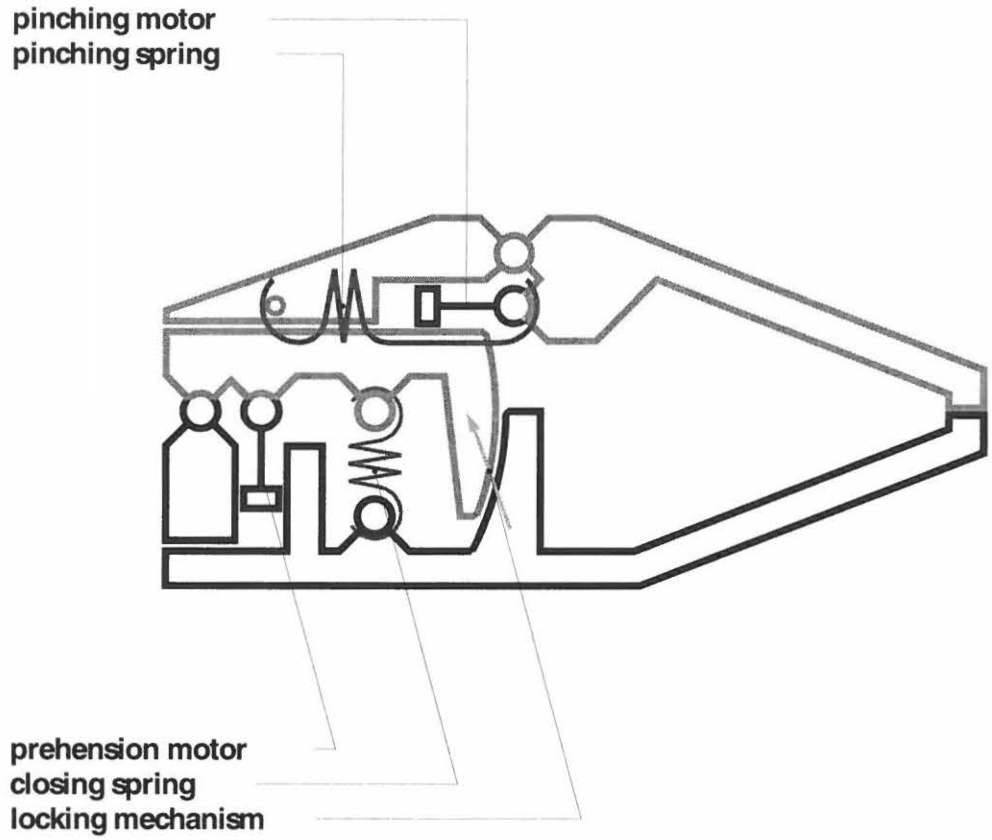


Figure 1: *A schematic drawing of the pneumatic bi-phasic hand prosthesis. The sequence of operation is explained in the text.*

To open the hand the pinching motor is pressurized first. It neutralises the action of the pinching spring. Next the locking mechanism unlocks the thumb and the hand opens as the prehension motor is pressurized. Closing the hand is in reversed order. First the prehension motor is blown off. A weak spring closes the hand. Upon thumb contact the locking mechanism locks the thumb and the pinching motor is blown off, permitting the pinching spring to exert its force between the fingers and the thumb. This system requires little energy to operate: the thumb displacement is at a very low force level [the weak closing spring] and the high pinching force is exerted at almost no displacement. To ensure the desired operation sequence of the three pneumatic motors a logical circuit has been designed. It contains several especially designed pneumatical/mechanical logical elements. The system is powered by pressurized CO_2 from commercially available disposable cartridges. The desired supply pressure is obtained from a small pressure-reducing valve, designed especially for this purpose, Figure 2.

Figure 2: *The bi-phasic hand prosthesis is powered by compressed CO_2 obtained from disposable cartridges. The desired supply pressure level is attained through a pressure-reducing valve, shown here at the left in the figure.*

A further reduction of the amount of gas needed to operate the hand mechanism is achieved by a careful assessment of the friction losses in the different pivot points and in the seals of the pneumatic actuators. Ball bearings were chosen in the pivot points. The pneumatic actuators are all of the piston type, with the piston sealed against the cylinder by an O-ring seal, mounted by special guidelines in order to minimise friction losses. Moreover, special emphasis is placed upon the avoidance of dead spaces within the construction of the mechanism. Finally the supply pressure level has been chosen at the optimum, i.e. the level that operates the hand at the minimum cost in gas. The optimum supply pressure level was found at $p = 1.2$ MPa; and it is shown to be invariable for changes in the energy output required, the timing of the hand cycle, and the length of the gas ducts.

RESULTS

A technical prototype of the pneumatic hand mechanism is designed, built and tested, Figure 3. This prototype validated the concept of a pneumatically powered hand prosthesis, Table 1. The hand mechanism is light, small, and reliable in the sense that it functions well for over 75000 cycles in the laboratory, and it is fast.

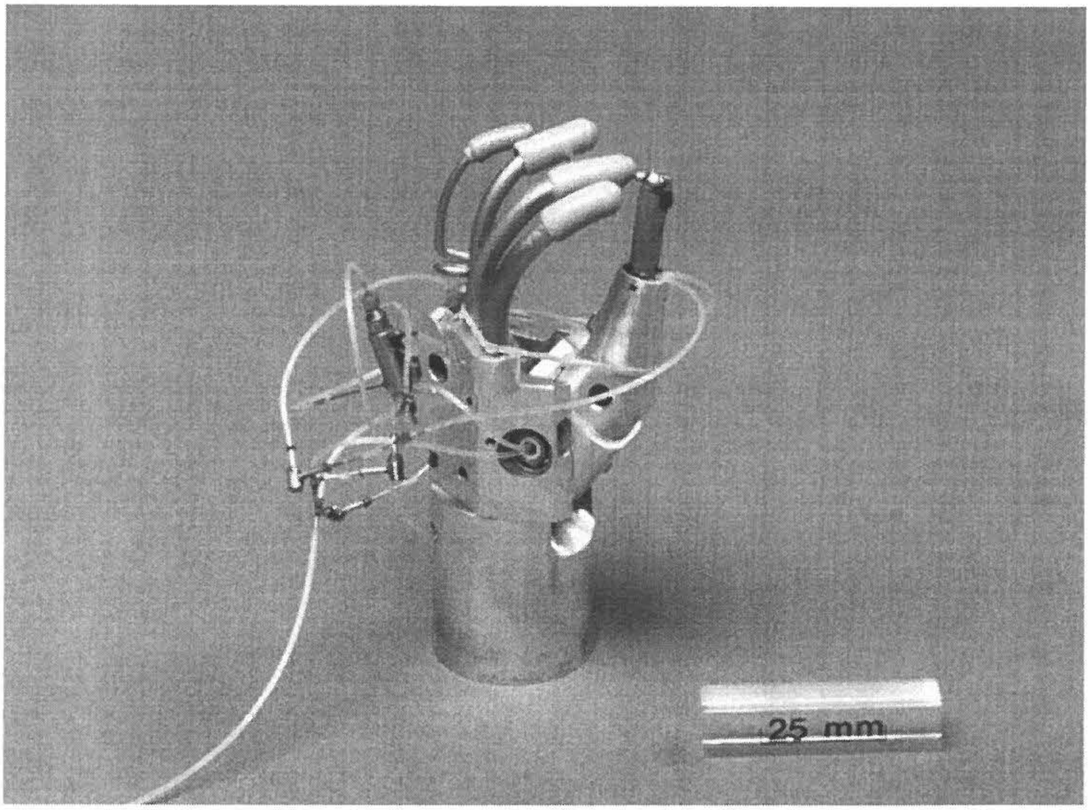


Figure 3: *The WILMER pneumatically powered bi-phasic hand prosthesis for children.*

Encouraged by these results a prototype for clinical evaluation has been designed and constructed. It differs from the technical prototype only in constructive details, not in its principles of operation. The clinical prototype design shows a further improvement over the technical prototype in many of its specifications, Table 1.

Table 1: *Comparison of the technical and the clinical prototype of the pneumatically powered hand prostheses. The data for the electrical powered hand prostheses are listed as well.*

	MYOELECTRIC STEEPER	MYOELECTRIC OTTO BOCK	PNEUMATIC TECHNICAL PROTOTYPE WILMER	PNEUMATIC CLINICAL PROTOTYPE WILMER
MASS OF THE HAND [grams]	230	130	128*	60
MASS OF THE ENERGY STORAGE SYSTEM [grams]	75	60	60	36***
MASS OF THE COMPLETE PROSTHESIS [grams]	550	340	300**	250**

ELBOW TORQUE [Nmm]	760	470	400**	290**
ENERGY CONSUMPTION [per day]	1 BATTERY	1 BATTERY	0.5 GAS CONTAINER	< 0.5 GAS CONTAINER
OPERATING CYCLE [seconds]	2.5	>2.5	<1	<1

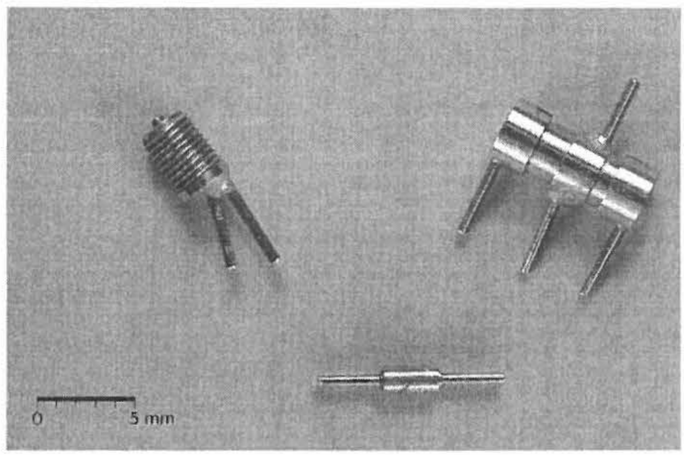
* without the frame & fingers: 41 grams

** estimated figure

*** estimated figure, based upon a mass for the pressure reducing valve of 4 grams after a redesign

The logical elements needed are designed, constructed, and built, Figure 4. They can be characterised by their small overall dimensions, a low mass, very low operating forces, small dead space volumes, and reliability. Within the flow range, these logical elements stand out over commercially available logical elements.

Figure 4: *These logical elements ensure the desired operation sequence of the bi-phasic pneumatic hand mechanism. Top left: the pneumatic switch, a mechanically operated normally closed valve. Top right: the pneumatic relay. Bottom: the check valve.*



CONCLUDING REMARKS

With this study, it is shown possible to improve upon the electrically powered hand prosthesis considerably by the use of pneumatic power. Clinical evaluation should endorse this statement under practical conditions. We believe pneumatic power to be the better alternative in externally powered prosthesis.

ACKNOWLEDGEMENTS

The contribution of present and former members of the WILMER group is greatly acknowledged.

REFERENCES

[1]Plettenburg D.H., "Electric versus pneumatic power in hand prostheses for children", Journal of Medical Engineering & Technology,13, 124-128, 1989

[2]Plettenburg D.H., "A sizzling hand prosthesis. On the design and development of a pneumatically powered hand prosthesis for children", PhD-thesis, Delft University of Technology, The Netherlands, ISBN 90-370-0196-3, 2002

PROSTHETIC CONTROL:
A CASE FOR EXTENDED PHYSIOLOGICAL PROPRIOCEPTION

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ABSTRACT

To achieve subconscious prosthetic control the patient feedback present must be employed as completely as possible. This implies the use of control methods based upon the principles of extended physiological proprioception.

The harnessing of body movements has the inherent ability to fully employ the principles of extended physiological proprioception. However, the present harnessing techniques often fail to do so and are generally of a dreadful engineering quality. Myoelectrical control must be considered as an open loop system. It lacks by principle any useful feedback.

The challenge for the prosthetic profession is to focus research on [improvement of] control options that comply with the rules of extended physiological proprioception. Promising future control options may result from the research into miniature cineplasties, in combination with neuro-muscular reorganization, and from the research into neuro-electrodes.

INTRODUCTION

Many prostheses are not being used. Numerous surveys of the actual use of prostheses have been made [a.o. 2, 7, 13, 14, 19]. Although the outcome of these surveys must be considered with care, as most of them are not based on a sound methodological approach, most are retrospective and do not use a control-group [8], they all indicate a 40 - 60 % of non-users for activities of daily living, vocational activities and leisure activities. Moreover, most studies indicate a 20 - 40 % of non-wearers. People with an arm defect get very frustrated about the performance of their prosthesis shortly after being supplied with one. So after a while the prosthesis finishes up in the closet. The frustration is caused by the discrepancy between the expectations and the reality. This discrepancy is due to inadequate information and inadequate education of the patient, to incompetent professionals, and to inadequate equipment [15].

A patient wants and expects a prosthesis that looks naturally beautiful, that is comfortable to wear and that is easy to use. Unfortunately none of the existing prostheses fulfills all these demands. This poses a challenge to the engineers of the WILMER group to try and develop new prosthetic devices that more closely meet the needs of the patients. In our opinion these needs can be summarized as cosmetics, comfort and control. These three demands are the basic requirements for a prosthesis. All three must be fulfilled, otherwise the prosthesis will be abandoned. This paper is focussed on the requirements in the control domain.

PROSTHETIC CONTROL - REQUIREMENTS

In operating a prosthesis we have to distinguish between the actuation of the prosthetic device, and the control over that device. To move a prosthesis and counteract the gravity forces and the friction losses of the prosthetic mechanism, energy is needed. This energy can be drawn from an external source, for example a battery in the case of an electric motor driving the prosthesis. Hence, this type of prostheses is called externally powered. The energy can also originate from the body of the wearer of the prosthesis, i.e. by harnessing body movements muscle energy can be transferred to the prosthesis. Hence, this type of prosthesis is called body powered.

With control the user determines the position and/or the velocity of the prosthesis, the magnitude of the forces exerted on the environment, and, if applicable, the status of the joint or joints, that is locked or unlocked. Several control options are available, either mechanical or electrical. A control signal always originates from the user of the prosthesis. He decides when to contract a muscle in order to pull a cable or to activate an electrode. Ideally the control of a prosthesis does not require a lot of effort of the user; subconscious control is strived for as to keep the mental load as low as possible. Here the role of feedback must be emphasized. Control theory teaches that without feedback the control of systems subject to external disturbances, like prosthetic systems, is very difficult. The controllability of the sound human hand is very good as a result of many feedback paths present. If a hand is missing, many of these feedback paths are lost. Hence, it is of utmost importance to utilize the feedback paths still present. Here the concept of extended physiological proprioception comes to aid. Dr. David Simpson introduced this concept in 1971 [20]. Simpson depicts the prosthesis as a mechanical extension to the natural system of the human body. Common examples of such mechanical extensions of the natural system are the stick of a blind person, the golfer's club or the way we use a hammer. The way we use these devices shows that we shift outwards the points at which we make contact with the things that we observe as objects outside ourselves. "We pour ourselves out into them and assimilate them as parts of our own existence." This assimilation can be achieved through the use of the body's own joints as control inputs. The involvement of proprioception in the control of artificial limbs provides a totally different situation from that of open loop control where the loop is closed by vision. If vision is necessary for the control of the limb, close, careful and continuously monitoring of the mechanical device is needed in order to prevent difficulties. Providing proprioception relieves the user of this mental load.

For the design of a prosthetic device extended physiological proprioception implies that the movements and forces acting upon the prosthesis correspond with the movements and forces generated by the neuro-muscular system on the control site of the assistive device. It has to be build such that a simple and direct relation exists between the position of the control joint and the position of the prosthesis, between the velocity of the joint and the velocity of the prosthesis, and between the joint force and the force of the prosthesis, Figure 1. The fulfillment of these conditions is a necessity for proper control of prostheses. Implementation of these relations only gives optimal controllability if the polarities of movements, velocities, and forces correspond to physiology.

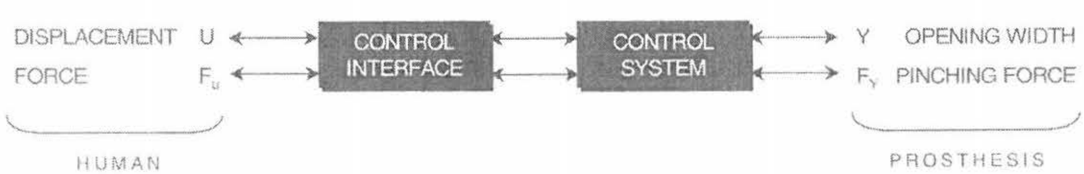

Figure 1: *A diagram depicting the essentials of extended physiological proprioception for the design of a prosthetic device. Extended physiological proprioception implies that the movements and forces acting upon the prosthesis correspond with the movements and forces generated by the neuro-muscular system on the control site of the assistive device.*

PROSTHETIC CONTROL - OPTIONS

The control of a prosthesis by harnessing movements of the body has the inherent ability to fully employ the principles of extended physiological proprioception. Proper use can be made of the feedback paths present. When controlling a prosthesis with myocontrol no use is made of the feedback paths left. Myocontrol must be considered as an open loop system. The actual EMG-signal is considered as a by-product of a muscle contraction in ".......a similar manner to considering the exhaust of a car as a manifestation of the engine's rotation. In both cases the output is approximately proportional to the activity of the motor. There is equally a lack of accurate and suitable afferent communication........" [20]. Although body powered prostheses have the inherent ability to comply with the rules of extended physiological proprioception, most body powered prostheses of today fail to do so because of lousy engineering. Current research efforts focus on the elimination of the complaints associated with the present harnessing techniques [3, 11], and on ways to more fully employ the force feedback by the use of voluntary closing terminal devices [16, 18, 21].

Another control option that complies with extended physiological proprioception is cineplasty. Many decades ago direct muscle attachment or cineplasty has been employed for prosthetic control. Cineplasty offers excellent feedback capabilities, however, it seems incompatible with the cosmetic demands put upon a prosthesis. Today, the cineplasty routine might go through a revival with the modifications as proposed by Dr. Dudley Childress [4, 5]. Childress and his team have experimented with small exteriorized tendon cineplasty to control an externally powered prosthetic servomechanism [9, 22]. Another suggestion made by Childress is the creation of prosthetic control sites by the fusion of small muscle areas with the overlaying skin by removal of the subcutaneous fat tissue. In combination with the neuro-muscular reorganization option, presently investigated by Todd Kuiken [12], the cineplasty becomes a promising option, especially to enable the control of multiple degrees of freedom.

Yet another potential control option may result from the research into neuro-electrodes. If and when it becomes possible to record neuro-activity through implanted electrodes for ultimately a man's lifetime, and if and when it becomes possible to stimulate nerves likewise, a powerful multi-degree of freedom control source will be rendered available [1, 6, 17].

CONCLUDING REMARKS

The challenge for the prosthetic profession is to render available prosthetic devices that do not end up in the closet, but really answer the user's needs. Therefore, to achieve subconscious control, research should focus on [improvement of] control options that comply with the rules of extended physiological proprioception. Promising future control options may result from the research into miniature cineplasties, in combination with neuro-muscular reorganization, and from the research into neuro-electrodes.

REFERENCES

[1] Andrews B., Warwick K., Jamous A., Gasson M., Harwin W., Kyberg P., "Development of an implanted neural control interface for artificial limbs", In: Book of Abstracts of the 10^{th} World Congress of ISPO, July 1 – 6, Glasgow, UK, p. TO8.6, 2001

[2] Atkins D.J., Heard D.C.Y., Donovan W.H., "Epidemiologic overview of individuals with upper-limb loss and their reported research priorities", Journal of Prosthetics and Orthotics, 8, 2 - 11, 1996

[3] Carlson L.E., "Improved control of body-powered prehension", In: Congress Book IXth ISPO World Congress, June 28 – July 3, Amsterdam, ISBN 90-74768-13-x, 102 - 104, 1998

[4] Childress D.S., "Control philosophies for limb prostheses", In: Progress in Bioengineering. Paul J.P., Barbenel J.C., Courtney J.M., Kenedi R.M. [eds.], Adam Hilger, New York, ISBN 0-85274-085-9, 210 - 215, 1989

[5] Childress D.S., Personal communication, 1998

[6] DeLuca C.J., "Control of upper-limb prostheses: a case for neuroelectric control", Journal of Medical Engineering and Technology, 2, 57 - 61, 1978

[7] Hacking H.G.A., Berg J.P. van der, Dahmen K.T., Post M.W.M., "Long-term outcome of upper limb prosthetic use in The Netherlands", European Journal of Physical Medicine and Rehabilitation, 7, 179 - 181, 1997

[8] Hacking H.G.A., "The prosthesis: a clumsy stooge [In Dutch: De prosthese: een onhandige handlanger]". In: Amputation and upper extremity prosthetics [in Dutch: Amputatie en prothesiologie van de bovenste extremiteiten], Linde H. van der, editor, ISBN 90-373-0412-5, 73 - 80, 1998

[9] Heckathorne C.W. et.al., "E.P.P. Control of an electric hand by exteriorized forearm tendons", In: Abstracts 8th ISPO World Congress, April 2 - 7, Melbourne, 101, 1995

[10] Herder J.L., Cool J.C., Plettenburg D.H., "Voluntary pinching", In: Congress Book IXth ISPO World Congress, June 28 – July 3, Amsterdam, ISBN 90-74768-13-x, 466 - 468, 1998

[11] Kruit J., Cool J.C., "Body-powered hand prosthesis with low operating power for children", Journal of Medical Engineering & Technology, 13, 129 - 133, 1989

[12] Kuiken T., "Neuromuscular reorganization to improve the control of myoelectric prostheses", In: Congress Book IXth ISPO World Congress, June 28 – July 3, Amsterdam, ISBN 90-74768-13-x, 659 - 662, 1998

[13] Kuyper M-A., Breedijk M., Muldres A.H.M., Post M.W.M., Prevo A.J.H., "Prosthetic management of children in the Netherlands with upper limb deficiencies", Prosthetics and Orthotics International, 25, 228 - 234, 2001

[14] Lunteren A. van, Lunteren-Gerritsen G.H.M. van, Stassen H.G., Zuithoff M.J., "A field evaluation of arm prostheses for unilateral amputees", Prosthetics and Orthotics International, 7, 141 - 151, 1983

[15] Plettenburg D.H., "Basic requirements for upper extremity prostheses: the WILMER approach", In: Proceedings of the 20th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, October 29 – November 1, Hong Kong SAR, China. Chang H.K., Zhang Y.T., [eds.], Vol.5/6, ISBN 0-7803-5164-9, 2276 - 2281, 1998

[16] Plettenburg D.H., Herder J.L., "Voluntary closing. A promising opening in hand prosthetics." Submitted to Medical Engineering & Physics, 2002

[17] Popovic D.B., Sinkjær T., "Control of movement for the physically disabled", Springer Verlag, ISBN 1-85233-279-4, 2000

[18] Radocy B., "Voluntary closing control: a succesful new approach to an old concept", Clinical Prosthetics and Orthotics, 10, 82 - 86, 1986

[19] Routhier F., Vincent C., Morisette M-J., Desaulniers L., "Clinical results of an investigation of paediatric upper limb myoelectric prosthesis fitting at the Quebec Rehabilitation Institute", Prosthetics and Orthotics International, 25, 119 - 131, 2001

[20] Simpson D.C., "The choice of control system for the multimovement prosthesis: extended physiological proprioception", In: The control of upper-extremity prostheses and orthoses. Herberts P., Kadefors R., Magnusson R, Petersen I. [eds.], Charles C. Thomas, Springfield, Illinois, ISBN 0-398-02869-9, 146 - 150, 1974

[21] Visser H. de, Herder J.L., "Force-directed design of a voluntary closing hand prosthesis", Journal of Rehabilitation Research and Development, 37, 261 - 271, 2000

[22] Weir R.F.*ff*, Childress D.S., "Tunnel cineplasty as a control input for externally powered prothetic components [In German: Die Kineplastik zur Steuerung von Fremdkraftprothesen]", Medizinisch-Orthopädische Technik, 121, 9 - 12, 2001

REAL-TIME COMPUTER MODELING OF A PROSTHESIS CONTROLLER BASED ON EXTENDED PHYSIOLOGICAL PROPRIOCEPTION (EPP)

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INTRODUCTION AND BACKGROUND

Extended Physiological Proprioception (EPP)

Proprioception utilizes the physiological components of the nervous and musculoskeletal systems to allow an individual to sense the position of their limbs subconsciously. By providing a rigid connection to an object this proprioceptive ability can be extended to the object and allow the user to sense the spatial location and orientation of these objects with respect to his or her body. This concept explains how a person can use a tennis racquet to hit a tennis ball without having to observe the position of the racquet during their swing or the way a blind person uses a long cane to 'feel' the location of objects in their surroundings.

Body-powered prostheses take advantage of this proprioceptive ability by relating the motion and position of the prosthesis to the motion and position of an intact joint of the amputee via the control cable. However, most externally powered prostheses do not have any mechanism with which to provide feedback regarding the state of the prosthesis to the proprioceptive system of the amputee. In these cases the amputee must rely on vision and other incidental sources of feedback such as motor whine and socket pressure to control their prostheses and this may place a significant cognitive load on the user.

Simpson suggested that the concept of "extended-physiological proprioception" (EPP) could be applied to externally powered prostheses [1]. He stated that, by mechanically linking the movements of an externally powered prosthetic joint (e.g., an elbow) directly to the movements of a physiological joint (e.g., the shoulder), the amputee could 'feel' the position of the prosthetic joint by using the proprioception inherent in the anatomical joint. Much like a body-powered prosthesis, the linkage between the two joints is able to provide feedback about the position and velocity of the prosthetic joint as well as loads that are applied to the prosthesis. The theory behind EPP control is that by directly relating the movements of the prosthetic joint to those of the anatomical joint, a prosthesis will be able to provide feedback regarding the state of the prosthesis in a manner that is physiologically appropriate in order to utilize subconscious pathways and therefore reduce the mental loading placed on the user.

Uni-directional vs. Bi-directional EPP Configurations

Two configurations exist for the implementation of EPP in an electric powered device. These configurations allow the user to control a single degree of freedom prosthetic component with either one or two control sources. Uni-directional EPP control requires only a single control source but this configuration constrains the prosthetic and anatomical joints to directly follow each other in one direction only (e.g., glenohumeral flexion as seen in figure 1). The two joints are only constrained to move with each other in one direction because the control cable only acts in one direction. It is possible for the anatomical joint to move more quickly than (or to beat) the prosthetic joint in the antagonistic direction (e.g., glenohumeral extension in figure 1). This condition can create slack in the control cable, which creates a situation in which EPP does not exist and feedback is no longer being presented to the user.

Full or bi-directional EPP control requires that the prosthesis is directly linked to the anatomical joint in a manner that produces an unbeatable position servo-mechanism in both directions and thus preserves extended physiological proprioception for all movements of the anatomical joint. An example of a bi-directional EPP system is the Fitch elbow (figure 2). This cable-operated elbow is configured so that humeral flexion results in prosthetic elbow flexion and humeral extension results in prosthetic elbow extension. Due to the fact that bi-directional EPP control preserves EPP at all times and thus creates a more intimate interface with the amputee, it should theoretically be superior to that of uni-directional EPP control. However, the uni-directional configuration has been more frequently utilized because of its simplicity and similarity to body-powered systems.

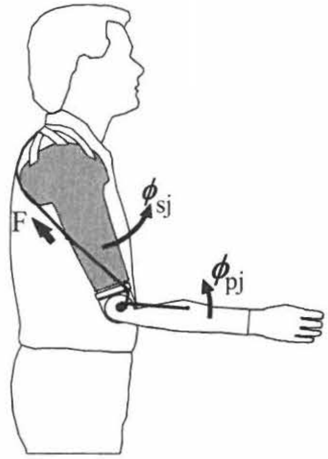

Figure 1*: Uni-directional EPP control of a powered elbow using flexion of the residual*

Weir [2] attributes the current lack of application of EPP in externally powered prostheses to the fact that currently available prosthetic components do not posses a sufficient bandwidth to allow the user to have subconscious control of the device. In other words, due to the slow response of currently available prosthetic components users feel as if they have to constantly pull on the control cables to achieve the position they desire. This results in the operator using the prosthesis in a 'bang-bang' approach. Weir feels that the sluggish response of currently available powered components causes EPP to have the opposite effect than that for which it was intended. Instead of providing sub-conscious control of the prosthesis, the amputee has their attention continuously drawn to the control of their prosthesis.

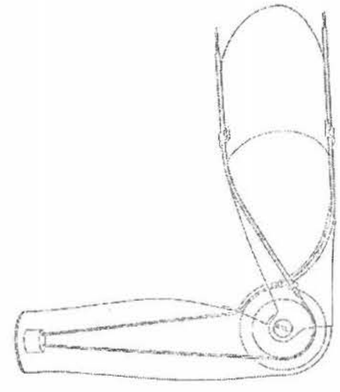

Figure 2*: The Fitch elbow, an example of bi-directional EPP control. (From Orthopaedic Appliances Atlas, Volume 2, 1960, p.51)*

Purpose

While the concept of EPP control seems credible and was demonstrated by Simpson ([3], [4]), implementation of such devices has been problematic. Our laboratory has developed an analog EPP controller [5] and two microprocessor based EPP controllers ([6], [7]), but only the analog controller has been clinically fitted to amputees. When mounted on the laboratory bench and set to high gains, the controllers exhibit smooth operation of an electric elbow in flexion and a 'jerky' behavior in extension. The goal of this study was to attempt to characterize our EPP controller and identify the factors that are contributing to this behavior in order to understand the nuances of EPP control and allow for the development of a clinically viable prosthesis that utilizes EPP principles.

METHODS

To investigate EPP control, a simulator has been developed using Matlab's Real Time and XPC Target toolboxes (Mathworks, Natick, MA). This simulator allows a controller to be designed in Mathworks' Simulink and then converted into C code on what is referred to as a 'host' computer. This code is then downloaded to a separate computer, referred to as the 'target' computer, using the XPC Target toolbox (figure 3). The target computer mimics a hardware controller while allowing the controller parameters to be easily changed from the host. The target computer houses a data acquisition card that enables the target computer to transmit drive signals to the prosthetic joint and allows for data collection.

To examine the capabilities of these controllers, an EPP controlled elbow system has been created using a Hosmer (Hosmer Dorrance Corporation, Campbell, CA) powered elbow. Operational amplifier circuits were constructed to allow for the collection of the elbow's angular position using a Helipot potentiometer (Helipot, Fullerton, CA) as well as collection of the force input control signal from a 25 lb. Flexiforce (Tekscan, S. Boston, MA) force-sensitive resistor (figure 3).

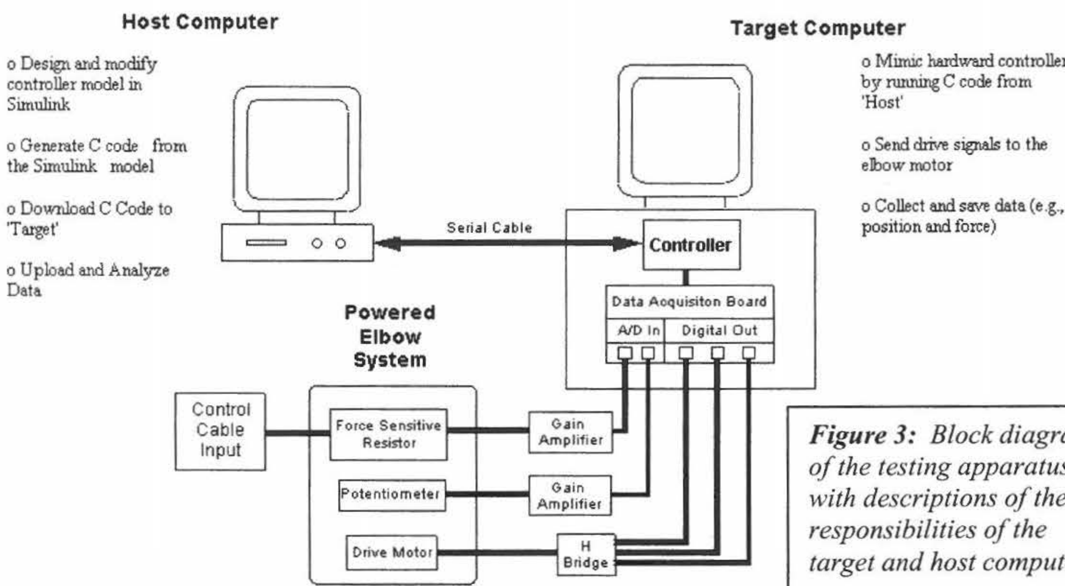

Figure 3: Block diagram of the testing apparatus with descriptions of the responsibilities of the target and host computers.

DISCUSSION

Mathworks' Real Time and XPC Target toolboxes permit considerable system versatility. It has allowed us to easily examine the effects of adding time delays as well as filters of different orders and cutoff frequencies to our system. It has also allowed us to easily assess the effects of varying the update rate and precision of the pulse width modulation of the drive motor signal.

We have spent considerable time using this system to examine the cable interface that exists between the human and the transducer. At present we have limited ourselves to examining a configuration in which a uni-directional EPP controller is implemented on a Hosmer powered elbow. We theorize that the behavior of the EPP system is affected, in part, by non-linearities in the uni-directional EPP configuration related to the control cable. However, the results that have been observed to this point are problematic due to the slow response of the elbow mechanism we are employing.

ACKNOWLEDGEMENTS

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REFERENCES

1. Simpson, D. C., (1974): "The Choice of Control System for the Multimovement Prosthesis: Extended Physiological Proprioception (e.p.p.)" In: The Control of Upper-Extremity Prostheses and Orthoses, Proceedings of the Conference on the Control of Upper-Extremity Prostheses and Orthoses, Göteborg, Sweden, Herberts, P., Kadefors, R., Magnusson, R.I., and Petersén, I., (Editors), Charles C. Thomas, Springfield, Illinois, pp. 146-150, October 6th-8th, 1971.

2. Weir, R.F.*ff.*, (1995): "Direct Muscle Attachment as a Control Input for a Position Servo Prosthesis Controller", Ph. D Dissertation, Northwestern University Evanston, IL.

3. Simpson, D.C., (1969): "An Externally Powered Prosthesis for the Complete Arm," in Basic Problems of Prehension Movement and Control of Artificial Limbs, The Institution of Mechanical Engineers, Proceedings 1968-69, 183(3j), pp. 11-17.

4. Simpson, D.C. and G. Kenworthy, (1973): "The Design of a Complete Arm Prosthesis," Biomedical Engineering, 8(2), pp. 56-59.

5. Heckathorne, C.W., Strysik, J.S., and Grahn, E.C., (1989): "Design of a Modular Extended Physiological Proprioception Controller for Clinical Applications in Prosthesis Control," Proceedings of the 12th Annual RESNA Conference, pp. 226-227.

6. Bertos, Y.A., Heckathorne, C.W., Weir, R.F. and Childress, D.S. (1997) "Microprocessor Based E.P.P. Position Controller for Electric Powered Upper-Limb Prostheses", Proceedings of the 19th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBS), Chicago, Illinois, Oct. 30-Nov. 2, 1997, pp. 2311-2314.

7. Al-Angari, H.M., (2001): "A Design of Two Degrees of Freedom Microprocessor-Based EPP Position Controller for Upper-limb Prostheses," Master of Science Thesis, Department of Biomedical Engineering, Northwestern University, Evanston, Illinois.

UNIQUE DEVICE-SELECTION STRATEGIES FOR POWERED ELBOWS

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BACKGROUND

In an upper limb prosthetic system Mode Selection gives the user the ability to utilize one or two input signals to control more than one prosthetic device. Mode Selection is defined here as the means by which a user shifts control to the different electric devices in their prosthetic system. Mode selection has been done for many years in several forms, each of which has a benefit to the user. The simplest mode selection scheme uses a simple switch. Other types involve the use of myoelectric signals that cross a specific threshold, or a rapid co-activation of two myoelectric signals to perform the mode selection. This latter method of mode selection has become a basic cornerstone for mode selection. However, the performance of the prosthesis must match the desired goals of the user. If such a technique cannot be mastered then other approaches must be made available.

Mode Selection As A Function Of Myoelectric Control Signals

In most situations, the mode selection technique is based on the use of the same inputs that provide control of the selected device. The rapid co-activation is the most common method for mode selection. This technique can be difficult for some users to master, and in some cases, with the upper limb user, it can not be performed while the prosthesis is in certain positions. With this in mind, one of the new capabilities of the Boston Digital Arm System is to provide more than one means of mode selection at a time. With the situation in point, a simple switch can be added so that mode selection is accomplished by activating the switch, rather than rapid co-activation. This method is included in any Boston strategy that uses mode selection, so that should the user not be able to master the mode selection technique, a simple switch can be added at any time to perform the task, or both can be used as required.

If rapid co-activation is not an option for a specific client, then a level (threshold) based approach may be used. This means that both myoelectric signals must exceed set thresholds, to initiate mode selection. With the help of the Boston interface software, the change from rapid co-activation mode selection to level sensitive can be applied through a simple menu click, allowing the user to experiment with the different options available. This ability to quickly change mode selection technique means the user will not have to wait for the prosthetist to change plugs or settings to potentiometers.

Mode Selection With Other Myoelectric Inputs

Some cases have been found where a third myoelectric site is available. This site can be utilized as a third input, providing the user with the ability to mode select without having to interrupt the signals providing control to the system. As above, the myoelectric signal can be sampled for rate of change, or for exceeding a specific level. The signal from this site can have its' noise threshold and gain adjusted just like the primary sites.

The use of the third muscle opens new possibilities for mode selection. Rather than cycling through available devices in sequence, the user can now select which device to operate. This is similar to the operation of an Otto Bock Double Channel hand. Should the signal reach a specific threshold, one device is selected. If the signal does not reach the threshold, another device is selected. In this case, a default device must be selected, so that when a selected device is not operated, a revert time brings control back to the default device.

Selecting the Default Device

In the Boston interface software, the default device can be selected from a drop down menu, depending on the type of mode selection desired. If cycling through devices is preferred, then the drop down box contains all possible combinations.
In a 3-device system the possibilities would be;

Table 1

Default	2nd	3rd
1. Hand	Elbow	Wrist
2. Hand	Wrist	Elbow
3. Elbow	Hand	Wrist
4. Elbow	Wrist	Hand
5. Wrist	Hand	Elbow
6. Wrist	Elbow	Hand

Mode Selection in Non-Myoelectric Systems

In upper limb systems where no myoelectric signal is available, other inputs must be used. These currently include, linear transducers, Touch Pads™, and simple switches. The linear transducer is also referred to as a servo. (See Fig. 1)

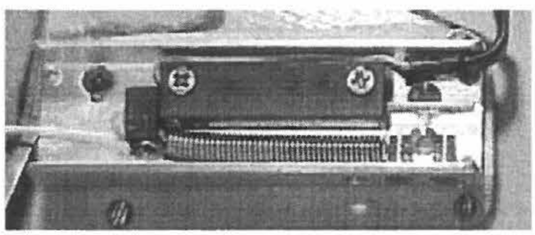

Figure 1

The servo, or linear transducer, generates a voltage signal that increases as the transducer is pulled. This voltage can be used to control the position of the elbow, in which case the system is called a positional servo – transducer position controls elbow flexion angle. The voltage is similar to the output of a Touch Pad or Bock electrode, so it can also replace one of these as a signal source. When the servo is used in conjunction with one or two myoelectric inputs, the user has ability to control two devices simultaneously. As a case in point (Fig 2&3), a recent Boston system utilized three of these linear transducers, one acting as a servo. The other two controlled independent open and close, but the co-activation of these two resulted in mode selection. The two had to reach specific thresholds before mode selection would occur.

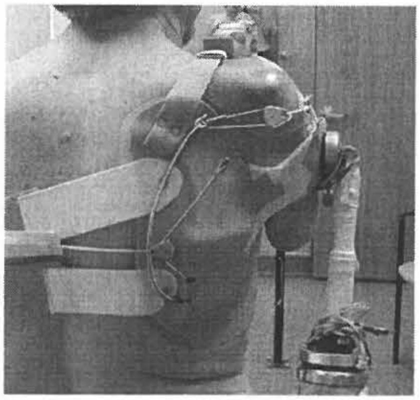

Figure 2 [1]

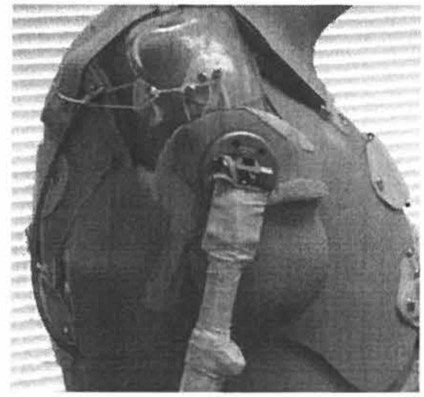

Figure 3 [2]

The use of Touch Pads, force sensitive resistors, is also appealing in upper limb prosthetic systems. The most common systems utilize three individual Touch Pad sites that can be activated by the user. Two of the sites are for function, while the third site is used for mode selection. Three often used sites use the tip of the acromion that is moved forward and back for control and up for mode selection. This mode selection scheme, similar to myoelectric, has adjustments for noise, gain, and the threshold setting for the mode selection site. Similar techniques, as described for myoelectric, can also be applied to the Touch Pad.

Simple switch systems rarely use mode selection, but when it is necessary, the Boston can interpret the signals differently. With switch control a single dual action switch usually controls direction while a second switch may be used to do mode selection. When the second switch is single action, it may be used to cycle through the system devices. With a dual action switch, each position can select a separate, non-default device.

User Feedback During Mode Selection

When changing control from one device to another, it is important to let the user know that the change has occurred. The Boston system does this in a variety of ways. The simplest method is to have the system emit a tone. The pitch, volume, and duration of the tone can be modified through the interface software. Once training is completed, the user may elect to change these parameters, or disable them completely.

This method of feedback can be taken a step further. The Boston is capable of driving up to five devices. When cycling through the different devices, the controller can create sounds that have different pitch, volume, and duration based on the device that has been selected. Elbow selection might produce a long high-pitch sound, while hand selection could be indicated by a short low-pitch sound. The settings for each device are independent and can be set through the interface software.

Some users may find that the sound feedback is too irritating for continued use. In this situation, the system can use vibratory feedback to signal a successful mode change. This can be done whenever one of the five available outputs is not being used to operate a device. The user can then choose how intense and how long the vibration is applied. This vibration feedback can also utilize the same features as described for the sound. A unique intensity and duration of the vibration can be assigned to each device selected. So when the elbow is selected, a short intense vibration might be delivered, while the wrist selection might give a long and low intensity vibration. As with the sound, the vibration may be disabled at any time using the interface software.

When sound and vibration are not convenient for user feedback, light can be used. In particular, when a specified device is selected, an LED can flash red or green. To enhance this feature, an LED can remain lit to indicate, by color, which device is currently active. This insures the user will not "spill the cup of coffee".

CONCLUSION

This paper has shown that the microprocessor in the Boston Digital Arm System now offers a multitude of control and mode selection choices. With the interface software, the system can be tailored to the users needs and capabilities. When user skill increases, further complexity can be added, while some feedback may be removed. This system is typical of what you may expect from LTI and other suppliers in the near future.

REFERENCES

1. Photo appears courtesy of Prosthetic Orthotic Solutions International, Marlton, NJ.
2. Photo appears courtesy of Prosthetic Orthotic Solutions International, Marlton, NJ.

Clinical Experiences with Animated Prosthetics Controller and LIMB LINK™

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ABSTRACT

This presentation will express clinical experience of utilizing the Animated Prosthetics controller for electronic fittings of many trans radial, trans humeral and higher level amputees. Also covered will be aspects of varied control strategies.

The Animated Prosthetics controller and the new LIMB LINK™ prosthesis control unit will be shown and demonstrated. It is a wireless device adapted to a hand held computer. It utilizes radio waves to allow adjustments to the controller without wires or cables. Fine adjustments can be done while the patient is working the prosthesis. The capabilities of the device and control strategies will be discussed and shown. One benefit of the ACS system with LIMB LINK™ is to have the capability to change control strategies with out having to obtain and change various componetry

The average prosthetist may have never had the opportunity to work with this little known control system. The presenter has fit over 70 ACS systems in the past 18 months. Patients fit represent all level from pediatrics to adults, Carpal level to Interscapular-Thoracic this represents a large percentage of all ACS controllers fit during this period.

There will be a live demonstration of the LIMB LINK™ on screen to allow all attendees to fully see the controls, adjustments and strategies available.

FINAL PAPER NOT RECEIVED AT PRESS TIME

MINI JOYSTICK for Upper Limbs Prostheses

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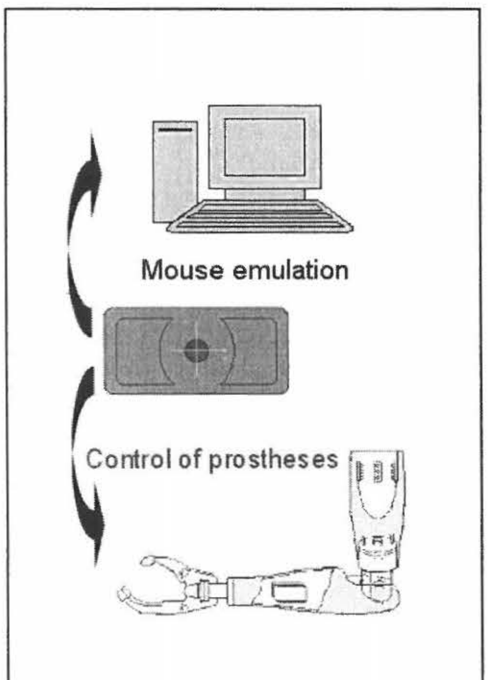
INTRODUCTION

The use of myoelectric command is often difficult to apply to congenital pathologies of the upper limb, which alternative solutions such as pressure sensors or common microswitches are often used. In such cases, prosthesis command entails the use of the same techniques used with the myoelectric method, in other words, the activation of the various hand, wrist and elbow functions by means of a cyclic system or one based on subdividing the signal detected by the sensors into various levels. In movements involving more than one joint, only one function is activated at any one time, which, although extremely safe, often makes these operations lengthier and sometimes unnatural. Therefore using a mini-joystick for prosthesis control can prove extremely convenient as it permits combined control the functions of the artificial limb simultaneously according to the direction in which the joystick is moved.

In today's society, the possibility to access a PC is increasingly important for people with prostheses and the possibility of using the joystick to emulate a PC mouse becomes essential for patients, especially those with proximal pathologies, as using a traditional mouse with an artificial limb is problematic and awkward.

This device permits the setting of two function modes:

a) Prosthesis control mode
b) Mouse emulation mode



The two modes are mutually exclusive; the patient usually uses the joystick to control the prosthesis as described below; for use with a PC, the patient connects a cable to the prosthesis that will redirect the data from the mini-joystick to the PC rather than to the prosthesis. The decision to use a commercial component, on the one hand means having immediate access to a low-cost device with a PC compatible interface and that is therefore able to function directly as a mouse emulator and on the other, requires the implementation of a software interface integrated into the prosthesis control circuit that is able to receive information in mouse format.

Various devices with different characteristics were analysed in order to choose the product that:

- best exploits residual mobility in the remaining body segment;
- supplies a sufficiently accurate output to be processed by the algorithm that controls the limb movement actuators.
- is able to handle the control, event simultaneously, of three degrees of freedom, precisely, the opening/closure of the hand, rotation of the wrist and flexion and extension of the elbow.

Other important characteristics for the choice of the control device are:

- limited dimensions;
- limited weight;
- reliability;
- low consumption

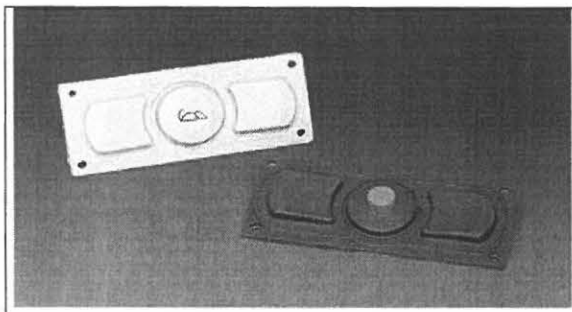
The analysis of the various commercial devices available resulted in the choice of the product manufactured by Interlink. It is composed of a 360° directional control joystick and two buttons, on a support of just 7.5 x 3 cm and supplied with clamping holes.

It is compatible with Microsoft DOS, Windows 9x, 2000, 3.x and OS/2 operating systems and is supplied with RS232C hardware interfaces for PS/2 mouse, USB and direct serial ports with TTL levels.

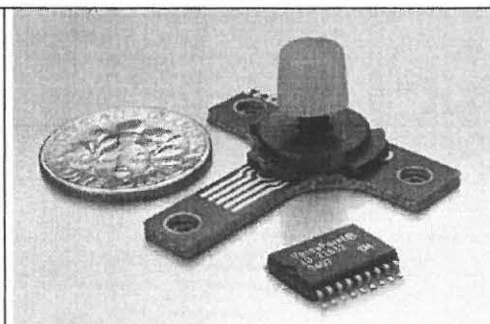
It consumes < 6 mA at 5V DC and < 3 mA at 3,3V DC.

Each action of the device, supplies 3-byte packages that, after suitable processing, provide information on the pressure of one of the two buttons and joystick position, with a variability range of -128 to +127 on both the x and y axes.

The thickness and the weight (1 cm at the highest point and < 15 gr respectively) are almost negligible.



Assembled versions



Version with separate components

In addition to the version of the mini-joystick used for this project, another non-assembled version of the joystick is available, composed of the MicroJoystick, the microcontroller for sensor handling and communication with the PC and buttons, which provide maximum assembly versatility, in order to exploit the extremely limited dimensions of the single components without altering the other functional characteristics.

MOUSE EMULATOR

Communication protocol is in Microsoft mouse serial format. Each time the mini-joystick is moved, three bytes, each of which has 7 bit data, 1 stop bit and no parity bit, are sent at the speed of 1200 bps. Movements on the x-axis are positive towards the right and negative towards the left; movements on the y-axis are positive downwards and negative upwards. The L and R bits are at 1 when the left or right button respectively is pressed. X7-X0 and Y7-Y0 are the X and Y co-ordinates expressed in 2's complement format (range from –128 to 127).

Byte	B6	B5	B4	B3	B2	B1	B0
1	1	L	R	Y7	Y6	X7	X6
2	0	X5	X4	X3	X2	X1	X0
3	0	Y5	Y4	Y3	Y2	Y1	Y0

A driver is supplied with the device in order to use the mini-joystick as a mouse for PC with DOS, Windows 3.x, W9x and W2000 operating systems. Software for handling device function parameters for defining the device's functional characteristics as desired (e.g. the speed and acceleration of the cursor, orientation, etc.) is also supplied. The interface towards the PC is available in serial and USB versions.

PROSTHESIS CONTROL

Prosthesis control must include commanding the three electromechanical joints that can be present: hand, wrist and elbow, for a total of 6 functions. For example, the joystick can be used to control elbow and wrist movements and the 2 buttons can be used for hand control (one for opening and one for closure), even simultaneously. Considering the movement of the joystick on a perpendicular Cartesian plane, each joystick position has a pair of values (x,y), which univocally identifies two variables that will then be processed by the system to provide the desired limb movement.

For example, by associating the hand movement to the x co-ordinate and the elbow movement to the y co-ordinate, according to the position of the joystick (both co-ordinates have a range of –128 to +127) data can be processed and handled with suitable algorithms for controlling the actuators associated to them, thus also providing a proportional command to the motors. In this case, the two lateral buttons are used for wrist movements.

The chart below defines the commands that are given by moving the joystick:

Close hand Flex elbow	Flex elbow	Open hand Flex elbow
Close hand	Stop	Open hand
Close hand Extend elbow	Extend elbow	Open hand Extend elbow

Referring to the chart, it can be observed that by moving on the main axis, the command action refers to one degree of freedom only, whereas by moving diagonally the command action involves 2 degrees of freedom. A 3rd degree of freedom could be activated by the buttons, and this movement would also be simultaneous with the other two degrees of freedom. In order to vary the speed of the actuators and make prosthesis movements even more natural, it is possible to establish thresholds within each sub-quadrant within which the speed of the actuator has a different value: for small movements beyond the speed threshold null the actuators move with low speeds that gradually increase towards the outside of the quadrant.

The association of commands to the actuators, the definition of thresholds and the laws that regulate the speed of the actuators inside the quadrants can be set according to patient requirements. It should also be pointed out that defining the thresholds in an appropriate manner makes it possible to eliminate the simultaneous movement of more than one joint. A software package that permits the technician to calibrate the threshold and parameters according to the patient's abilities and residual functionality has also been developed. It should also be pointed out that the use of the joystick as a mouse emulator has made it immediately possible to use various commercial software packages (video games, etc.) as training software for patients on device use.

Despite the fact that the use of a standard device has a number of advantages, it also requires the implementation of the communication protocol with the mini-joystick module within the control circuit. With the current version, the control circuit must be programmed by indicating that the command is provided by mini-joystick rather than electrodes or microswitches; work is currently in progress on the creation of a system that permits the controller to adapt to the connected sensors automatically.

CONCLUSIONS

The system has now been tested for approximately a year on 4 patients fitted with prostheses with electromechanical elbows and the results would appear to be very good. The aspect that is most appreciated is the possibility to command the computer mouse directly, whereas only two of the patients appreciate using the combined elbow and hand movement. All the patients were already prosthesis wearers and this doubtlessly influenced their opinion. However, the number of cases is still too limited to draw any definite conclusions.

MICROPROCESSOR CONTROL FEATURES

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The very Underlying Principle of Control for the Otto Bock Myoelectric System is "Muscle contraction should lead to function." If you think about picking up an object with your sound hand, you simply pick up the object. An amputee must concentrate on more activities than that of an individual with a sound hand. If an amputee wears a prosthesis they must first open their hand, then position their hand over the object, then close their hand around the object and finally determine how much grip force should be applied to the object. With all of those things to consider, the relationship between their input signal (EMG) and the output of the hand (motor speed or grip force) must remain constant to minimize the learning curve. If the relationship is variable, the patient must relearn how to control the hand depending on the variables and thus control can be very unpredictable. The analogy is getting into someone's car that just had a brake job done. You are used to putting high pressure on the pedal in your car in order to slow the car down so when you touch the pedal in the other car, it brakes very abruptly. You need to relearn the relationship between pedal pressure and braking speed. This is something we need to avoid in myoelectric fittings. Therefore, the microprocessor control in the Otto Bock system contains features to minimize the effects of outside influences. The outside influences include the following:

Opening position:
As the glove is stretched while the hand is opening, resistance to the motor is increased and without intervention would cause the hand to slow down.

Battery Voltage:
In the morning when the battery is fully charged, the hand runs at full speed. During the day the voltage of the battery slowly drops off and without intervention would cause the motor to run slower with the same signal (EMG) from the amputee.

Temperature:
Colder temperatures produce higher resistances to the motor than if the temperature were warmer. Without intervention the cold temperature could make the hand run slower.

Aging glove:
If a glove or inner hand shell is old and stiff, this again will produce high resistance to the motor. Without intervention the motor would run slower with an old stiff glove than with a new flexible glove.

Normal wear and tear:
As mechanical parts wear out they run less efficient putting higher resistance on the motor. Without intervention this would cause the motor to run slower.

Slipping objects:
If an object's center of gravity or mass changes it could potentially slip out of the prosthetic hand. This would be potentially a problem when filling a glass with water where the held glass is in the prosthetic hand for example.

Crossover signal:
With other systems crossover signal, co-contraction is a very difficult problem to overcome. If the two signals necessary to run two site myoelectric systems are continually influencing each other, the patient is constantly seeing different results in function for the same muscle contraction.

Electro-magnetic disturbances:
While working in some environments, Electro-magnetic noise coming from some electrical sources can cause intermittent functional problems within the hand. Without intervention it would be impossible for the amputee to control the hand and have to avoid exposure to these environments.

Weak signals:
Weak signals can jump up and down erratically and cross the "ON" threshold many times while performing a singular function. Without intervention, this would cause the function to be jumpy and erratic as well.

The Otto Bock microprocessor has features that minimize the affects of these variables so the amputee can learn the control relationship once and the hand takes over from there to maintain that relationship. The microprocessor features are as follows:

Closed loop control:
Motor speed is constantly evaluated to make sure that the same input signal (EMG) produces the same output (motor speed or grip force) no matter the variables. Muscle contraction equals function and the function should always be the same. An analogy would be cruise control in a car. Old technology: the cruise control set a fixed pedal position and if the car went up or down a hill it would either slow down or speed up because it didn't vary the power to the motor. New technology: cruise control sets the speed of the car and varies the pedal position to maintain a constant speed whether going up hill or down hill. In myoelectrics, Otto Bock microprocessor technology varies the power to the motor to make the speed constant with the same EMG signal. In proportional DMC control this is especially important. If the amputee generates a high signal the hand should always run at the same high speed. If the amputee generates a small signal the hand should always run at the same slow speed. Patient benefit is that the hand is predictable and consistent.

Proportional Grip Force:
The question is "Proportional to what?" Otto Bock Microprocessor technology produces a unique relationship for grip force and signal strength. The strength of the grip force is proportional to the strength of the muscle contraction. No other manufacturer has done this. Patient benefit is that this relationship is constant and more physiological and therefore easy to learn and understand. A relationship that remains constant can be learned and thereby produce proprioception. Also while holding fragile objects; the grip force is not increased by small inadvertent signals above the on threshold.

Auto Grasp:

This feature is found only in the sensor hand. In the human hand, when an object in the hand gets heavier, the Autonomic nervous system response is to increase grip force to keep the object from slipping away. The Sensor technology in the hand is designed to mimic this response. Patient benefit is that when holding objects that change weight such as the case when filling a glass with water, the hand automatically grips harder to prevent the object from slipping away.

First signal wins:

Two signals are used to open and close a typical myoelectric hand. The first signal to cross the "ON" threshold will win over the other signal. Patient benefit is that even in situations of crossover signal or some refer to it as co-contraction, they can still have smooth proportional control over their hand. The speed is not influenced by the difference between the two contractions.

Virtual ground:

The operational range of the control of the hand is adjusted automatically inside the control circuit in situations where there might be Electro-magnetic disturbances present in the system. Patient benefit is that they can control their hand even in areas where there are Electro-magnetic disturbances such as would be found around some electronic systems like TV's or fluorescent lights.

Different "ON" and "OFF" thresholds:

The "ON" threshold is fixed at .54V output from the electrode and the "OFF" threshold is fixed at .35V output from the electrode. While EMG is an appropriate signal to control a myoelectric device, it can be rather jumpy in nature. To run a hand slowly requires a small EMG signal. If the "ON" and "OFF" thresholds were the same, the EMG might jump up and down crossing the threshold many times causing a pulsation of the hand while it is running slowly. Separating the thresholds benefits the patient by allowing for smooth control over a slowly moving hand.

To summarize, Otto Bock microprocessor technology is appropriate technology to maintain the ideal that muscle contraction should lead to function and therefore produces many patient benefits. In the end the patient is not bothered with the details of how the hand works, they just know that it does.

NEW PROSTHETIC CONTROLLER EXPANDS CAPABILITIES

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INTRODUCTION

The new VariGrip™ III Multi-Device Controller from LTI has expanded the capabilities of microprocessor-based prosthetic controllers. Traditionally, one or two prosthetic devices were controlled by two myoelectric input signals. This worked well for some people, but others could not master it and therefore they needed a different control strategy. The VariGrip solves this problem by offering a variety of pre-programmed control strategies. It allows prosthetists to set up systems to control up to four devices for greater versatility. Rather than restricting the user to myoelectrodes, the VariGrip III accepts inputs from myoelectrodes, Touch Pads™, servo transducers or switches to fully utilize the user's capabilities. The controller is compatible with all manufacturer's single-motor terminal devices which further expands the options available to the prosthetist.

CONTROL STRATEGIES

The new VariGrip III Multi-Device Prosthetic Controller represents the "next generation" of prosthetic control system for externally powered upper-limb prosthetics. Its versatility allows practitioners to customize the controller to suit each user. Not only can the best control strategy be chosen from the pre-programmed selection, but user-specific adjustments can be made to further adapt the system. This generally simplifies the operation and results in improved functionality. Prosthetists can try more than one control strategy to determine the best one for their patient. Strategies are easily down-loaded to the prosthetic controller for these trials. Once the strategy is found, various adjustments can be made to optimize the system's performance.

VARIGRIP CIRCUIT

The VariGrip III is considerably smaller than other microprocessor-based prosthetic controllers (Figure 1). At just 1" x 1½" x $^{3}/_{8}$" (25½ x 38 x 9½ mm) for the two-device controller, it can easily fit in the forearm of most adult and pediatric prostheses. It is also small enough to pass through the opening in a Bock Quick-Disconnect wrist for easy assembly and service. This size allows prosthetists to build the lamination without unnatural bulges and odd protrusions. It weighs just 14.6 grams, 22% lighter than other controllers and compared to prosthetic components such as batteries at 50 to 70 grams or hands at 500 grams, the weight of the controller is insignificant. If more than two devices are to be controlled, a second circuit can be added.

Figure 1: *VariGrip Controller*

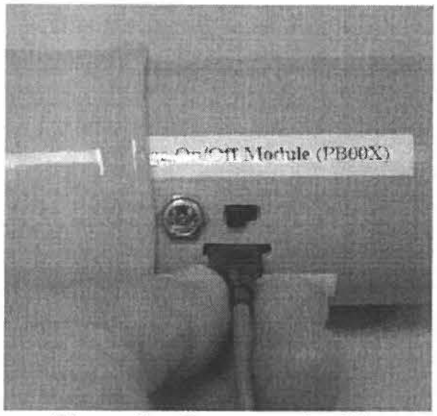

Figure 2: Programming Plug

The VariGrip controller connects to a personal computer through an interface unit that provides optical isolation, protecting the patient from potential shock hazards. A proprietary low-profile connector is used at the controller end. A convenient switch, recharge connector and programming plug is offered as an option with LTI built-in batteries (Figure 2). This provides an accessible connection port without compromising the cosmesis of the system. Once a cosmetic glove is installed, these components are hidden.

The input device(s) chosen depend on the capabilities of the user, and several inputs can be used on a single system. These can also be mixed so that one device is controlled by one input signal and another device by a second. In fact, with this approach, two prosthetic devices can be controlled independently and therefore simultaneously for greater functional efficiency. The system is compatible with most common terminal devices and the current limit for each device is programmed into the software. By selecting the devices to be used in the prosthetic system, the current limit is automatically set to protect these devices and to conserve battery power.

SOFTWARE

The VariGrip III uses software developed by the Bloorview-MacMillan Centre in Canada. This software is also used with the Variety Ability Systems' Single Programming Module (SPM), so users can easily switch back and forth between these systems. **MyoWizard**™ and **MyoAssistant**™ are software packages that enable prosthetists to communicate with the prosthetic controller. Both use convenient graphical user interface screens, enabling the prosthetist to make system adjustments. MyoWizard is used to determine or change the control strategy on an existing system or to load a strategy onto a new system. These strategies are available in a pull-down menu and presently, sixteen control strategies are pre-programmed so the prosthetist can select the optimal control technique for the user. MyoAssistant is designed to "assist" the prosthetist in evaluating the patient, setting parameters such as gains and thresholds and selecting the output devices. In addition, co-contraction thresholds and time windows can be set to match the user's capabilities. This software will be discussed in a separate presentation.

FUTURE ENHANCEMENTS

The hardware has been designed to allow for expanded capabilities as the software develops further. For example, the hardware is capable of accepting up to five input signals on a single prosthetic system and providing user feedback. Future software enhancements will include single-site control strategies and two-motor terminal devices. For users with only one useful muscle site, single-site control is a useful alternative to the traditional two-site control. Both Centri and Otto Bock offer two-motor terminal devices. Centri's UltraLite hand for adults and Otto Bock's 2000 pediatric hands use two motors. These require different strategies however, because one employs two drive motors and the other a drive motor plus a brake motor. As new control strategies and devices are added, they will appear in future releases of the software.

SOFTWARE/FIRMWARE TOOLS TO CUSTOMIZE CONTROLLER PARAMETERS IN UPPER EXTREMITY, POWERED PROSTHETIC SYSTEMS

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INTRODUCTION

Clinician-friendly software/firmware tools were developed which enable prosthetists and therapists to customize quickly and easily the control strategies used in upper-extremity, powered prosthetic systems in order to serve better their client's needs. The tools, MyoWizard and MyoAssistant, are used together to set and document the control strategy relating the client's input to the action of the prosthesis. MyoWizard guides the selection of an appropriate strategy for the client's prosthetic device from a list of pre-defined strategies and then programs the selected strategy into the device's micro-controller. It also allows the micro-controller to be reconfigured for another client. With MyoAssistant, clinicians fine-tune the controller's strategy to the client's specific needs and abilities by adjusting controller parameters, including gain and threshold values and other settings, to optimize performance. The adjustments are aided by graphical display of signals. Changes are immediately written into the controller and can be tested.

BACKGROUND

Clinical fitting of a prosthesis to a client requires customization of the system to the specific needs of each individual because of the uniqueness of each person and his/her situation. This customization applies not only to the physical system of socket, harness, prosthesis type and size, etc., but also applies to the controller which is used to convert the client's desires into a functional motion of the prosthesis.

Continued advances in microcontroller technology have enabled the development of smaller, more flexible systems which prosthetists can use to address their clients' needs. The small size, low power usage, and field-programmable capability of the microcontrollers make them well-suited for use in prostheses, enabling prosthetic controllers that can be adapted for a broad spectrum of amputees. Many manufacturers, such as Variety Ability Systems Incorporated (VASI), Liberating Technologies, Inc. (LTI), Motion Control, Otto Bock, and Animated Prosthetics have developed field programmable controllers. [1-8]

The type of control strategy that is chosen for a client depends upon a number of factors. These include the number of prosthetic components to be controlled, the number of input sites the client can control independently, the characteristics of the signal generated by the muscle(s), switch(es), force-sensitive resistors (FSRs), or other input devices, and the cognitive ability of the user.

The control system can operate a prosthesis in one of two ways, providing digital or proportional control of the prosthesis. In digital control, the speed of the motor will be at a maximum once the input signal exceeds the threshold. In a proportional system, the speed of the motor will change in proportion to the change in magnitude, for example, of the input signal. Proportional control is used for larger hands where finger travel distance is significant, and for people who want finer control of hand positioning.

The controllers used in powered prostheses come in two forms: non-programmable and programmable. With non-programmable controllers, strategy and prosthetic device are fixed. For example, the non-programmable SCAMP hand from RSL Steeper has a single-input voluntary opening strategy, and VASI provides a non-programmable version of its programmable SPM board, which can be factory-converted to become programmable. Programmable boards take advantage of the flexibility provided by their microprocessor 'brains' to provide for selection of the control strategy and to enable different configurations of the selected strategy. These flexible controllers allow the relationship between user input and prosthetic output to be adapted to individual needs.

DISCUSSION

The software/firmware tools developed at Bloorview MacMillan Children's Centre are compatible with the VASI and LTI field programmable controller hardware and enable the controllers for upper extremity, powered prostheses to be fit to a wide range of client needs. The tools have been designed for clinicians who are trained in powered prosthetics and who are familiar with control strategies. The tools allow for input from a variety of input types, such as switches, force sensitive resistors or FSRs, and EMG electrodes. They enable selection of the control strategy from a variety of potential strategies, and enable the parameters within each strategy to be adjusted for optimum use by the client, as well.

The software/firmware tools, MyoWizard and MyoAssistant, are used to set and document the control strategy relating the client's input to the action of the prosthesis. The tools communicate with the microcontroller (a Microchip PIC16F877 microcontroller) to set and to customize the control strategy. The communication occurs through the serial port of a PC, and the cable connector optically isolates the data signals from the PC and converts the RS-232 side from the PC to the TTL side from the controller. Through a user-friendly graphical interface on the PC, the clinician can configure the controller for the client.

About MyoWizard

MyoWizard is used to program a control strategy for a *programmable* controller. It takes the clinician through a step-by-step process of selecting the strategy and programming the microcontroller. It allows the clinician to:

- Select a control strategy from a list of pre-defined strategies, and then write the control strategy into the controller. Some strategies can be implemented in digital or proportional form.
- Re-configure the controller for another client.

About MyoAssistant

MyoAssistant is used to fine-tune the control strategy already existing in the microcontroller. Any changes made are immediately downloaded to the device. This includes changing gain and threshold values, and other settings needed to customize and optimize the control strategy for the client. MyoAssistant incorporates visual feedback to the clinician and user about the relationship between the input signals and the motor outputs. This valuable feedback assists in adjusting the system parameters and in training. MyoAssistant can be used with both *programmable* and *non-programmable* controller boards. It allows the clinician to:

- Read information about the current control strategy from the controller.
- Adjust gain and threshold settings of input channels. (Note that the ability to adjust the gain at the controller allows the gain at the electrode pre-amplifier to be set and kept in an area where its behaviour is linear.)
- Adjust settings that are specific to a strategy.
- Autocalibrate input signals to maximize full dynamic range and minimize the effects of noise or 'cross-talk' between agonist-antagonist muscles, for example.

- Select another type of output device (*programmable* controllers only.) A wide variety of output devices can be selected, including most commercially available hands, powered wrists, and VASI powered elbows.
- Change output device settings such as direction and speed.
- Check battery levels.
- Save client information and controller settings in a database on the computer.
- Recall client information and previous settings from the database, and reload them onto the controller.
- Print out settings.
- Gather and print statistics about a client's usage of the prosthetic device.

As mentioned above, the controller boards that are compatible with MyoWizard and MyoAssistant include: the Single Programmable Module (SPM) from VASI (This board can operate devices that have one motor only), and the VariGrip III Multi-device Programmable Controller from LTI (This board can operate devices that have up to four motors). Table 1 lists the features available to these controllers from MyoWizard and MyoAssistant.

	Tasks	**SPM: Programmable**	**SPM: Non-programmable**	**Varigrip III Controller**
MyoWizard	Change strategy	*		*
	Reconfigure controller for another client	*		*
MyoAssistant	Adjust input gain and threshold	*	*	*
	Adjust strategy-specific settings	*	*	*
	Change output device type	*		*
	Save and recall from database	*	*	*
	Print from database	*	*	*

Table 1: *Features of MyoWizard and MyoAssistant with VASI and LTI controllers.*

SUMMARY

Microprocessor technology advances have enabled reconfigurable controllers which can fit within prostheses allowing the prosthetist to individualize the system to the user's needs and allowing the prosthetist to adapt the system as the physiological or cognitive needs and abilities of the user change. The user-friendly software/firmware tools developed take advantage of these microprocessor features and allow the systems to be modified quickly and easily, better serving the client's needs. MyoWizard and MyoAssistant are used together to set and document a customized control strategy relating the client's input to the action of the prosthesis. An appropriate strategy for the client's prosthesis is selected from a list of pre-defined strategies and then programmed into the microcontroller. Also, the microcontroller can be reconfigured for another client. After programming the controller, the strategy is fine-tuned to the client's specific needs and abilities by immediately adjusting controller

parameters. The adjustments are aided by an autocalibration routine and by graphical display of signals, enabling the client to benefit from a quickly and easily customized prosthetic controller.

ACKNOWLEDGEMENTS

The support of our clinical colleagues is much appreciated. They are Sheila Hubbard, Anne Martin, Winfried Heim, Sol Wierzba, and Shane Glasford. Financial Support for this project was provided by the Ontario Rehabilitation Technology Consortium (ORTC) and Variety – The Children's Charity.

REFERENCES

1. Heim, W, Naumann, S, and Bush, G, "Prosthetic Management of Upper Limb Deficient Children," The 10th World Congress of the International Society for Prosthetics & Orthotics, 1-6 July, 2001.
2. Williams, TW and Wallace, C, "New Control Options for Upper-Limb Prostheses," The 10th World Congress of the International Society for Prosthetics & Orthotics, 1-6 July, 2001.
3. Hanson, W, and Farnsworth, T, "New Digital Control Technology for Powered Prosthetic Elbows," The 10th World Congress of the International Society for Prosthetics & Orthotics, 1-6 July, 2001.
4. Naumann, S, and Heim, W, "Control of Electrically Powered Upper Limb Prosthetic Systems," Association of Children's Prosthetic-Orthotic Clinics, 2002 Annual Meeting, 10-13 April, 2002.
5. Haslinger, G, "A PC Based System for Selecting and Optimizing Myo Controls to the Patient's Needs," MEC '99, 25-27 August, 1999.
6. Kurtz, I, Heim, W, Bauer-Hume, H, Hubbard, S, Ramdial, S, "Programmable Control: Technical Aspects and Clinical Experience at Bloorview MacMillan Centre," MEC '99, 25-27 August, 1999.
7. Wallace, C, and Williams, TW, "New Clinically-useful Control Strategies Made Possible by the Varigrip II™ Multi-Device Controller," MEC '99, 25-27 August, 1999.
8. Hubbard, S, Bush, G, Kurtz, I, and Naumann, S,"Myoelectric Prostheses for the Limb-Deficient Child," Physical Medicine and Rehabilitation Clinics of North America, 2:4, 847-866, November, 1991.

COMPARATIVE ANALYSIS OF MICROPROCESSORS IN UPPER LIMB PROSTHETICS

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The recent emergence of microprocessor based prosthetic control for the individual with upper limb deficiency has significantly expanded the spectrum of treatment options and inclusion criteria for this patient population. Microprocessors can accept a wide variety of input devices and ranges enhancing an individual's prosthetic function allowing control options for individuals who were at one time not candidates for such prosthetic management. Additionally, myoelectric control parameters can be adjusted to optimize function while retaining the flexibility to individualize each prosthesis.

Historical Perspective

Many significant achievements are apparent as one examines the evolution of electronic upper limb prostheses over the last thirty to forty years. The evolution of electronic upper extremity prostheses can be summarized into three distinct generations. First generation electronics, often referred to as digital systems, used an on and off control scheme to actuate electronic terminal devices, wrist rotators, and elbows. These digital systems exhibited a single speed or single rate type of actuation of prosthetic terminal devices. During the first generation there was limited sophistication of input devices. At that time, input devices consisted of myo-electrodes and various switch technology often mounted in the prosthetic interface or attached to a control harness.

Delineation between first and second generation was made at the introduction of the Utah Arm and later the ProControl I prosthetic controller. Both systems allowed for large-scale threshold manipulation, gain or muscle EMG signal amplification as well as adjustment of muscle contraction rate in an attempt to minimize effort required in first generation co-contraction type switching. These systems lower the microvolt requirement (by lowering the muscle thresholds) for terminal device, wrist, or elbow control allowing more individuals with upper limb deficiency to take advantage of myo-electric prosthetic technology. Most importantly, these systems introduced proportional control in a reliable electronics package.

Though more sophisticated than the first generation, second generation electronics exhibited challenges that affected the ability of the prosthetist to provide expeditious prosthetic management and interchangeability. Through the second generation, switch activated, single site or dual site myoelectric control systems required different electronic packages. If, during the rehabilitation of the patient, it was noted that dual site type of control was too difficult and single site would be more appropriate, an entire new electronics package would need to be installed into the prosthesis creating additional expense and fabrication time at the point of rehabilitation where timing and expeditious prosthetic function is so very critical.

The third and most current generation of prosthetic electronics incorporates programmable microprocessors. Third generation electronics are delineated by the acceptance of proportional control as the standard. Microprocessors of the third generation allow a more infinite range of adjustment of myoelectric characteristics for the enhancement as well as ease in prosthetic control.

The significant advantage of microprocessor use in upper limb prosthetics is to provide the ability to modify control options (without purchasing or changing components) and input characteristics during initial fitting quickly and easily within the clinical setting. The use of microprocessors allows more complex filtering of the EMG signal resulting in enhanced terminal device responsiveness. Secondarily, microprocessors provide ease in changing control thresholds and sensitivity of the prosthesis as the users strength and ability evolves. Furthermore, microprocessors utilize algorithms to inherently adjust to various situations unknown to the patient. While many related topics are discussed in the literature, there is a void in the literature when one investigates commercially available microprocessor technology (1-9). A significant portion of this paper is based on discussions with individual manufacturers as well as comparative microprocessor analysis performed by the authors in conjunction with the American Academy of Prosthetics and Orthotics Upper Limb Certificate Lecture Series (10, 11).

Overview of Microprocessor Comparisons

Currently, three programmable microprocessors at the transradial and hybrid type levels were reviewed: Otto Bock – Sensor Hand or DMC plus, Motion Control - ProControl II, and Liberating Technologies Incorporated – Programmable VariGrip III.

With multiple processors available, it is difficult to identify the appropriate component for a particular patient. Variables that should be examined include the amount of clearance available for microprocessor integration, weight of the microprocessor and appropriate power supply, compatibility of existing components, as well as, the patient requirement for such technology. This comparison of the microprocessors provides valuable feedback for the prosthetist as he or she weighs the advantages and disadvantages of each system to optimize both functional and cosmetic requirements for a patient. Fundamental differences were investigated; in particular the characteristics that make each processor unique in clinical application. Objective and subjective comparisons will be explored to discern the fundamental differences between microprocessors.

A rating system approach, commonly utilized in the technology field, provided comparison parameters. A point system was applied to each to provide a means of comparison.

Objective comparisons included:

Provision of written instructions
Weight of the microprocessor
Power supply versatility
Interface Intrusiveness
Control scheme versatility
Warranty guideline of the particular microprocessor

Subjective comparisons included:

Interface method/programming ease
Fabrication specifics

Support Structure
Hybridization – or the ability to incorporate other manufacturers components

Differences noted between the microprocessors indicate that each system possesses unique characteristics that make it attractive or advantageous in specific types of clinical situations. Specific differences will be reviewed at length during the presentation.

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Harold Sears, Ph.D, Motion Control
William Hanson, MS
T. Walley Williams III, MS, Liberating Technology

REFERENCES

1 Heckathorne, CW. "Components for Adult Externally Powered Systems", Atlas of Limb Prosthetics, ed by Bowker and Michael, 2nd Ed, Mosby, Chicago (1992) 151-74.
2 Childress, DS "Control of Limb Prostheses", Atlas of Limb Prosthetics, ed by Bowker and Michael, 2nd Ed, Mosby, Chicago (1992) 175-198.
3 Lamb, D. W. "State of the Art in Upper-limb Prosthetics". J of Hand Therapy January-March (1993) 1-8.
4 Prosthetics & Orthotics Engineering, February 1997:1-3.
5 Kyberd, P.J. and Chappell, P.H. "Object-Slip Detection During Manipulation Using A Derived Force Vector". Mechatronics (1991) 2:1:1-14.
6 Nightingale, J.M. "Microprocessor control of an artificial arm". J of Microcomputer Applications (1985) 8:167-173.
7 Kyberd, P.J., Holland, O.E., Chappell, P.H., Smith, S. Tregidgo, R., Bagwell, P.J., and Snaith, M. "MARCUS: A two Degree of Freedom Hand Prosthesis with Hierarchical Grip Control" IEEE Transactions on Rehab Engineering. (1995) 3:1:70-76.
8 Datta, D. and Brain, N.D. "Clinical Applications of Myoelectrically Controlled Prostheses"
Critical Rev in Phys and Rehab Medicine (1992) 4(3.4): 215-239.
9 Hortensius, P. and Onyshko, S. "A Microcomputer-Based Prosthetic Limb Controller: Design and Implementation". Annals of Biomedical Engineering. (1987) 15:51-65.
10 Research and related work in conjunction with fellowship lecture "Microprocessors in Upper Limb Prosthetics", Chris Lake, CPO, FAAOP, for the American Academy of Orthotists and Prosthetics Certificate Course in Upper Limb Prosthetics. Including the following discussions initiated in November 2000 which are currently ongoing:
Discussions with Hans Dietl, PhD. Otto Bock, Austria and Pat Prigge, C.P., Otto Bock, USA
Discussions with Harold Sears, PhD, Motion Control
Discussions with William Hanson, MS, and T. Walley Williams III, MS, Liberating Technologies
11 Lake, C., Miguelez, JM. "Overview of Microprocessors in Upper Limb Prosthetics," as presented at the ISPO 2001 World Congress in Scotland.

NOTES

August 23, 2002 (Friday)
9:00am – 10:30am

Theme: Research

Invited Speaker:
9:00 – 9:25
P. J. Kyberd
A study of EMG signals from limbs with congenital absence and acquired losses

9:25 – 9:40
A. S. Poulton
Experience with the intelligent hybrid arm systems

9:40 – 9:55
Sam Phillips
Distributed forelimb pressures for prosthetic control

9:55 – 10:10
Marcus Reischl
Control and signal processing concepts for a multifunctional hand prosthesis

10:10 – 10:25
Alvaro Rios
Myoelectric prosthesis with sensorial feedback

10:25 – 10:40
Abidemi Bolu Ajiboye
Neuro-fuzzy logic as a control algorithm for an externally powered multifunctional hand prosthesis

A STUDY OF EMG SIGNALS FROM LIMBS WITH CONGENITAL ABSENCE AND ACQUIRED LOSSES

Sean Taffler, Oxford Orthopaedic Engineering Centre, NOC, Oxford.
Peter Kyberd, Oxford Orthopaedic Engineering Centre, NOC, Oxford, and
Bio-Cybernetics Unit, Reading University, Reading, RG6 6AX.

INTRODUCTION

The basic assumption made by the designers of EMG amplifier systems is that the substantial difference between different person's detected EMG signals is the energy generated, but the spectral content is sufficiently similar to be ignored. Anecdotally, there is little or no expectation that there is any structural difference between the muscles of persons with congenital absence and those with an amputation, although newer evidence suggests this is untrue [1]. This approximation works well for simple amplifier/detector systems that were common a decade past [2], but with the increasing availability of compact on-line processing devices and the increase in differing EMG processing schemes this approximation is likely to prove unreliable [3].

This study was made with persons who attended the Oxford Limb Fitting Centre at the Nuffield Orthopaedic Centre, Hospital in Oxford UK, from 1989 to 2001 as part of a program looking at novel means of processing, and the analysis of EMGs [4,5]. It sought to understand the variability of the results as they were derived from the techniques applied (Fuzzy logic, Neural Networks and Wavelet transforms). While undertaking the study the new information from the structural studies in Texas [1], showed that it was a worthwhile exercise, and one likely to produce unexpected results.

METHOD

Electromyograms from five individuals with congenital absence (CL) of their forearms were compared with those from three persons with acquired losses (AL) and eight without any losses (NL), an additional subject with a loss above the elbow was recorded. The subjects with a limb absence were drawn from the user population of the centre and had a wide range of prosthesis experience, from passive through body powered to active myo-electric users. The signals were detected using a custom-made Delsys Bagnoli-2 system and the conventional electrodes. This system was commissioned especially for the experiment after initial experiments with a Motion Analysis system and a custom-made in-house system showed a need for a wider bandwidth than was usually available. The custom systems bandwidth extended from 11Hz to 3.4kHz. The data was then collected on a PC via a standard data capture techniques.

Each subject was asked to contract a muscle on the forearm. If they used a myoelectric prosthesis, then their control muscles were chosen. Otherwise the muscle they had best control with was used. For comparison, the contra-lateral of the muscle was also recorded.

The signals were processed and analysed using skew and kurtosis of the frequency analysis using Fourier and Hilbert transform based techniques [4,5]. The latter is a technique that allows a greater resolution of the frequencies of the signals than conventional Fourier based methods [5]. The structure of the muscle of one of the CL subjects was observed using conventional ultrasound.

RESULTS

Across all measures there was a detectable difference between the CL group and the AL group, but little difference between the AL subjects and the NL cohort. For example, the measured skew for the frequency analysis of the CLs signals was (51 +/- 20) this was about double the value of the NLs (27 +/- 1), but the skew for the AL group was in the same area (29 +/- 10). This analysis used the Marginal Hilbert spectra rather than the Fourier based Power Spectral Density function, which is less sensitive, but achieves a similar separation, because of this three CLs and one NL had to be excluded

A t-test between the three groups showed a strong separation, with a value of 83% likelihood that the Normal and Acquired populations are the same, while the chances of the CLs and NLs being in the same population was 3% and that the two limb absent groups being the same being as low as 6%.

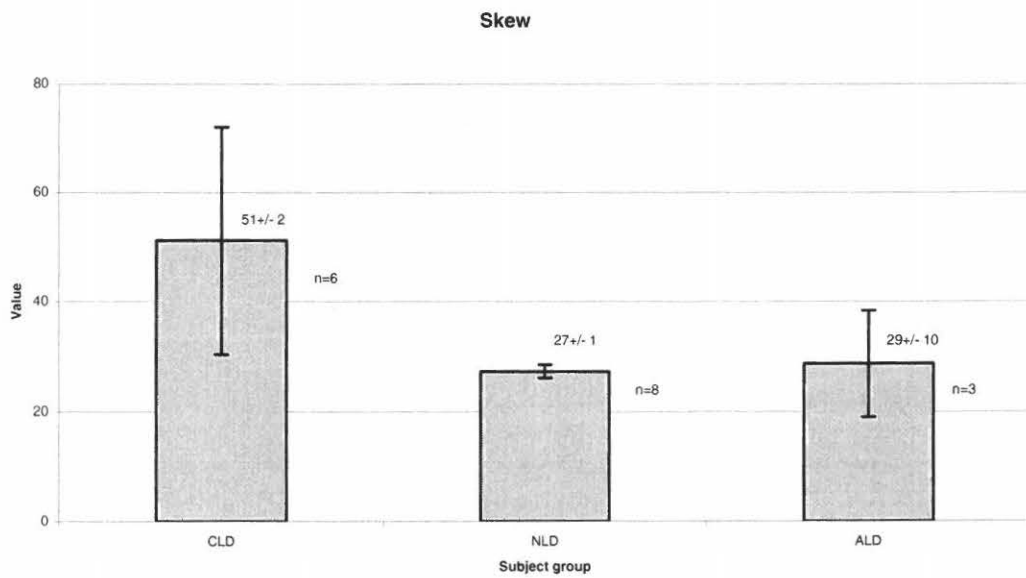

Figure 1: *The skew of the Marginal Hilbert Spectra of the three groups; (CLD) Congenital absence, (NLD) No loss group, (ALD) Acquired Loss.*

The Ultra-sound scans of the CL muscle agreed with the more extensive MRI study by Farry et al. [1], that the fibres were less ordered and the muscle had a more amorphous structure.

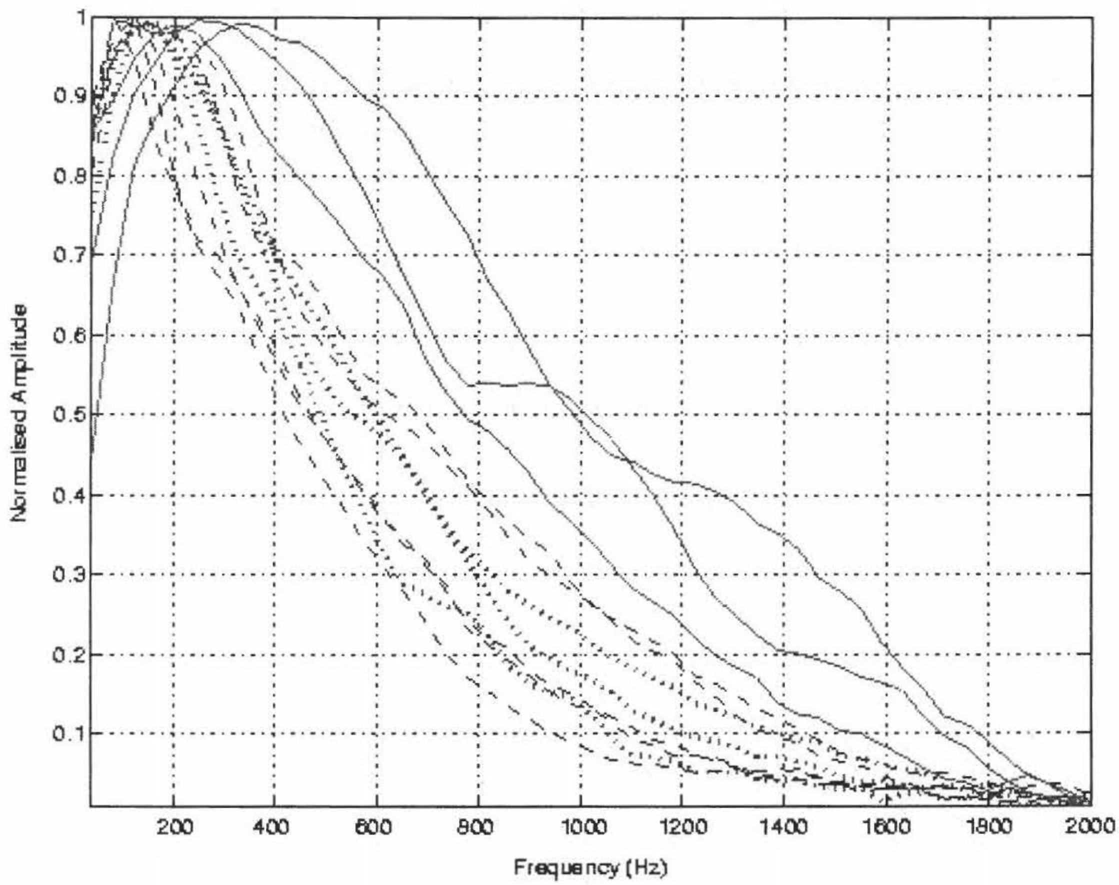

Figure 2: *Fourier transform based Power Density Spectra for the two groups: Dotted lines are CLDs and solid lines are NLD subjects.*

DISCUSSION

From the data, it is clear that there are both structural and electrical differences between muscles in persons with a congenital below elbow loss and those who lost their limb later. There are a number of factors that need to be considered before a possible explanation can be considered

The differences are likely to be developmental and not a matter of usage, since some of the subjects are described by their prosthetists as heavy users of their myoelectric hands and this did not lead to any correlation between the groups. There was no particular variation based on gender, side or age.

Fatigue can be discounted as another study on the effect of measured frequencies using the MHS showed a *lowering* of the frequencies as evidenced by the skew measures[3]. If this is coupled with the knowledge that the myo-arm users would be likely to have a greater endurance than the non-users, it would seem unlikely that the changes were based on fatigue.

There were little difference between measured isometric and isotonic contractions in conventional muscle, thus it seems unlikely that the factors relating to the tethering of the amputated muscle would be the explanation.

The observation of a difference in the structures of the muscle suggests the fibres are not in a parallel arrangement, this difference could be due to the fact that fibres in normal muscle are pulled by the joint and will tend to develop in one direction, growing to respond to the stimulus. What direct effect this might have on the signal can only be speculated at this point.

IMPLICATIONS

Irrespective of the causes, the observation has some implications for myo-processors. That all users have been able to employ the narrower bandwidth in everyday use is likely to be because although the CL users have a wider bandwidth they still produce energy at the lower frequencies, although it might explain the instances when the observed raw signal is looks sufficient, but the output of an amplifier/processor needs more gain than might be expected.

That the bandwidth for some users is wider need not necessarily mean the amplifiers must have a wider bandwidth. Allowing the higher frequencies may make the system more prone to other sorts of interference, thus practically speaking it might be better to continue as before, simply being aware of the differences.

However in recent years there have been a number of proposals for newer and more sophisticated myo-processors based on the underlying structure of the signals, including learning systems. Reported variations in their performance might be explained by the fact that no reference was made to the cause of the limb absence.

CONCLUSION

It is not safe to assume that all users of a myoelectric system are the same. Advanced controllers need to have a wider bandwidth if they are to be as useful for all possible users.

Of course the simplistic response is to simply turn up the amplifier gain.

REFERENCES

1. Farry K.A., Kramer L.A., Atkins D.J. and Donovan W.H. "Magnetic Resonance Imaging of Congenitally deficient Upper Limbs" In: MEC'99 Symposium Upper limb prosthetic research & consumer needs: Narrowing the Gap, Fredericton, NB, Canada, 25-27 August 1999. ISBN1-55131-013-9.

2. Scott R.N., "Biomedical Engineering in Upper Extremity Prosthetics." In: The comprehensive management of the upper-limb amputee. Atkins D.J. and Meier R. H., (editors), New York, Springer-Verlag, 1988, Chapter 12, pp. 173-189.

3. Taffler S.D. "The Use of the Hilbert Spectrum in the Analysis of Electromyographic Signals and its application in the development of myoelectric Prosthesis Controllers." D.Phil Oxford University 2001.

4. Taffler S.D., and Kyberd P.J. "Fuzzy logic in the interpretation of EMG signals for prosthesis control." In: MEC'99 Symposium Upper limb prosthetic research & consumer needs: Narrowing the Gap, Fredericton, NB, Canada, 25-27 August 1999. ISBN1-55131-013-9.

5. Taffler S.D., and Kyberd P.J. "The use of the Hilbert transform in EMG analysis." In: MEC'99 Symposium Upper limb prosthetic research & consumer needs: Narrowing the Gap, Fredericton, NB, Canada, 25-27 August 1999. ISBN1-55131-013-9.

EXPERIENCE WITH THE INTELLIGENT HYBRID ARM SYSTEMS

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THE OXFORD/EDINBURGH HYBRID ARM

Microprocessors are increasingly used in prosthetic applications. The flexibility they provide allows new functions to be added easily, and fitting and maintenance can be simplified [1,2,3]. Prosthetic controllers are available that can be adapted to different needs through field programming, allowing the prosthetist to try different control strategies or even invent completely new ones. The process of setting up the prosthesis is made easier through the use of graphical software programming tools [4]. However, there remains a need for interoperability standards so that complete prostheses can be built up from modular components that are compatible in software terms as well as mechanically and electrically.

Small local area networks, called fieldbuses, have been in use in manufacturing applications for some years. Typically they link together items of production machinery so that they work together automatically in a coordinated sequence. More recently fieldbuses have been developed for automotive use (CAN) [5] and for "intelligent" buildings (LonWorks™) [6]. One element in the success of these products is that technology has progressed to the stage where the network interface can be included on the same chip as a microprocessor. This means that processing capability can be placed physically where it is needed, with the various processors or nodes connected together by a simple cable. In both these markets the fieldbus has been a tidy solution to the problem of connecting increasing numbers of devices. The systems used in vehicles - lighting, security, engine management and so forth - required ever more complicated wiring looms as new features were introduced. These were expensive to produce both in material costs and in assembly time. The CAN bus dramatically reduced the amount of cabling needed. Similarly in buildings the LonWorks bus can link all the devices of the heating and ventilating systems, so that they can operate in a co-ordinated and energy efficient way, and it can also be used in the lighting and security systems. It is also possible to connect a LonWorks bus to the internet via a suitable interface, so that a building may be remotely monitored. Both of these technologies are designed to be used in harsh environments, having good inherent immunity to noise and the capability of recovering from errors.

Many of the benefits that fieldbuses bring to these areas may also be applied with advantage to prosthetics. The ease with which new components can be added to a system facilitates a modular approach. The robustness and reliability of the prosthetic system is improved, because:

- Messages are transmitted around the system in a digital format which is resistant to electrical noise and allows for detection and correction of errors.
- The number of wires connecting components is minimised. Connections are a weak point in any electromechanical system, particularly when they run through moving joints.
- Processing is carried out at the nearest convenient physical location, so that leads from sensors and EMG electrodes can be made as short as possible. This reduces their susceptibility to noise.
- Distribution of tasks over a number of processors gives a degree of graceful degradation,

where if one joint fails the remainder of the system can continue to operate. Further, the use of a standard digital bus allows simple communication with external devices, e.g. when using a PC to set up the prosthesis, or connection to the internet for remote diagnostics.

An experimental system to assess the practical merits of a fieldbus implementation for the control electronics of a prosthetic arm is described in this paper. It uses existing mechanical assemblies, namely the Oxford intelligent hand [7] and the Edinburgh arm [8], which has a powered elbow and rotating wrist. A LonWorks network was chosen as suitable microprocessors (Neuron® devices) were readily available at the time from two sources, and also because it implements a full OSI seven-layer model. Each Neuron device has two processors which are entirely dedicated to the operation of the network, and these are essentially invisible to the applications programmer. A third processor controls the applications. It runs programs that are compiled from a variant of C (Neuron C) that includes instructions to control the I/O features of the Neuron device, and supports multitasking. Thus the programmer deals only with the application and does not need to be concerned with the details of the network communications protocol.

Each Neuron device on the network acts as a local controller for the nearest joint, so that the only wires connecting joints are the network and power lines. Sensors and actuators are physically close to the relevant device. This device acts as an independent closed-loop controller, so that the amount of data that needs to be transmitted along the network is kept to a minimum. Circuit boards were made up for the Oxford hand and the Edinburgh wrist and elbow, using surface-mounted components and four-layer boards to reduce the size. The board for the hand represented an advance on the previous hand controller electronics, as the smaller size meant it could fit inside the hand envelope, instead of in the wrist space. In addition to the Neuron device it has drivers for finger, thumb and brake motors; analogue to digital converter taking signals from position and force sensors; and a signal processing circuit for slip detection. There are also inputs for EMG signals if the hand is used alone and not as part of a larger prosthesis. The boards for the wrist and elbow are similar but have motor drivers with a higher current rating, for the more powerful motors in these joints. As well as these motor controller circuits, another board was made up with a Neuron device and a fast analogue to digital converter with eight channels, referred to as a master or proximal board. It takes inputs from EMG electrodes, force sensing resistors, switches or similar input devices. In a multi-joint prosthesis it can select the joint to be controlled and send the relevant data to the joint's local controller via the network.

The arm uses the network to communicate with external systems. The LonWorks network can be connected to a PC through a transceiver which provides isolation to certified standards. A clinical support system that runs on the PC displays the EMG data graphically in real time, and allows parameters such as operating thresholds to be varied. Radio links have also been investigated, as they have the advantage that the user is not restricted by cables during setup and testing. This can be done with standard low-power transmitters and receivers which handle RS-232 serial data [9], though more advanced technologies such as Bluetooth may be used in the future.

An important power-saving feature of this system is provided by a distributed power management scheme, which relies on the ability of each Neuron device to go into a low power or "sleep" mode when the program has decided that it is not in active use. As a device cannot stay asleep indefinitely, it is woken up periodically by a "heartbeat" signal which originates from a clock in the master controller and propagates through the network to all the other Neuron devices. This mechanism is not used with a single hand - in this case it is woken by detecting activity in the EMG signals.

CASE STUDIES

The electronics boards were tested separately before proceeding to build a complete network. The Oxford hand can adopt two types of grip, precision and power. The user voluntarily opens the hand and then relaxes to allow it to close round an object. Using information from force sensors, the hand automatically adopts the appropriate grip and lightly holds the object. The user can then select a slip reflex where the grip force is automatically increased if the object starts to slip, or else voluntarily squeeze harder, or release the grip. These functions were rewritten in Neuron C to run on the new board. The new version of the Oxford hand was tested with two users.

Complete arm systems composed of hand, wrist and elbow were fitted to two users, one in Göteborg and one in Oxford (Figure 1). Control strategies were developed to suit each user during a training period that covered several visits. The flexibility of the system allowed new strategies to be set up and tested easily. The first user employed ON-OFF operation with an EMG amplifier and used a pull switch to switch between hand, wrist and elbow. The second arm user employed proportional control, also using a pull switch to switch between axes. He could produce myoelectric signals from two muscle sites, but he could not fully separate the two channels, and there was some co-contraction. In this case, a "winner-takes-all" strategy was created so that the signal that was smaller in magnitude was ignored, and the larger magnitude signal was used for control. This continued until muscles relaxed and both signals fell below pre-set thresholds.

User satisfaction is assessed through directed questions based on the findings of surveys [10] as well as the users' informal comments. Functional assessment will be made using a hand function assessment protocol that is based on abstract object handling and simulated activities of daily living [11, 12].

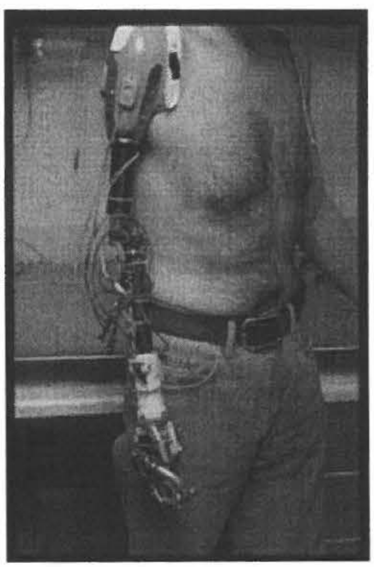

***Figure 1:** Hybrid arm.*

REFERENCES

1. Farnsworth, T. and Mandacina, S., "Clinical experience with programmable controllers in powered upper extremity prosthetics", 10th World Congress of the International Society for Prosthetics and Orthotics, Glasgow, 2001.
2. Heim, W., Kurtz, I., Woo, A., Ramdial, S. and Naumann, S., "Programmable control: clinical experience at Bloorview Macmillan Centre", 10th World Congress of the International Society for Prosthetics and Orthotics, Glasgow, 2001.
3. Lake, C., "Microprocessors in upper-limb prosthetics", 10th World Congress of the International Society for Prosthetics and Orthotics, Glasgow, 2001.
4. Hanson, W. and Farnsworth, T., "New digital control technology for powered prosthetic elbows", 10th World Congress of the International Society for Prosthetics and Orthotics, Glasgow, 2001.
5. Farsi, M., Ratcliff, K. and Barbosa, M., "An overview of controller area network", Computing & Control Engineering Journal, 10 (3), 113 -120, 1999.
6. Sharpe, L., "Setting a standard for building control", IEE Review 47 (4), 35-39, 2001.
7. Kyberd P.J., Evans M and te Winkel S., "An intelligent anthropomorphic hand, with automatic grasp", Robotica 16: 531-536, 1998.
8. Gow D, Douglas W, Geggie C, Monteith E and Stewart D., "The development of the Edinburgh modular arm system", Proc Instn Mech Engrs 215 Part H: 291-298, 2001.
9. Kyberd, P.J., te Winkel S. and Poulton, A.S., "A wireless telemetry system for training users of upper limb prostheses", Prosthetics and Orthotics International, 2002, 26, 78-81, 2002.
10. Kyberd P.J., Gow D, Scott H, Griffiths M, Sperling L, Sandsjo L, Almstrom C, Wartenberg C, Jonsson S., "A comparison of upper limb prostheses users in Europe", MEC '99 Myoelectric Controls Symposium, Fredericton, NB, Canada, 1999.
11. Light C.M., Chappell P.H. and Kyberd P.J., "Quantifying impaired hand function in the clinical environment", MEC '99 Myoelectric Controls Symposium, Fredericton, NB, Canada, 1999.
12. Light C.M., "An Intelligent Hand Prosthesis and Evaluation of Pathological and Prosthetic Hand Function." PhD thesis, University of Southampton, 2000.

RESIDUAL KINETIC VECTORS FOR PROSTHETIC CONTROL

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ABSTRACT

Imagined phantom movements have real physiologic expressions within the residual limb, which we termed residual kinetic activity. In order to exploit the residual functions of trans-radial amputees we developed a system to register dynamic pressure patterns produced by the residual limb during voluntary commands for motions. This pattern was then converted to a multidimensional vector. Specific requested finger motions were thus detected from the forearm as residual kinetic vectors (RKVs). To decode RKVs, we developed a trainable filter derived from the pseudoinverse of the pressure response matrix. The utility of RKVs was tested on several subjects who expressed both afferent and efferent phantom limb activity. Results showed that amputees using the RKV approach could control at least 3 robotic fingers in near-real time.

INTRODUCTION

As anthropomorphic robotic hands become more advanced with multiple degrees of freedom [1], their application as a prosthetic device is limited by the ability to measure control signals from amputees. Intensive efforts are underway to enhance prosthetic function by adding more degrees of freedom (DOF) through advanced signal processing of the myoelectric control signals [2,3]. While this approach is promising, it has not yet had a significant clinical impact. As an alternative approach, we have investigated the use of the kinetic, non-electrical, activity of the residual limb as a control signal for prosthesis control. Registering kinetic activity with an array of pressure sensors results in a residual kinetic vector (RKV) that differs fundamentally from the EMG signal. First, the RKV is a vector that dynamically represents the entire limb, not just individual muscle activity. Second, the RKV represents the end result of neuromuscular action, whereas the EMG represents its initiation, and thereby consists of Gaussian noise.

Translating Finger Volitions

The RKV approach exploits the efferent phantom ability of some below elbow amputees to control extrinsic finger muscles within their residua. This ability represents a phantom efferent phenomenon (PEP) whereby they imagine flexing and extending the metacarpal-phalangeal joints [4]. Their volitions and corresponding muscular activations produce pressure patterns at the residual forearm/socket interface, which are unique for each finger volition. These volitions can be represented as a set of residual kinetic maps.

Filtering RKVs

A below elbow residuum is a complex mechanical system of bone, muscle, skin, adipose, and scar tissue. Volitional activity is degraded through this system when measured as mechanical activity on the residuum surface by a pressure sensor array. Volition degradation and mechanical coupling could be characterized by a degradation function and its inverse, a restoration function, could discriminate specific movements.

Volition degradation is characterized by a system of linear equations

$$\mathbf{g} = \mathbf{Hf} \tag{1}$$

where **f** is a column vector representing intended activity, **H** is a matrix of the impulse responses, and **g** is a column vector of measured responses. Approximations of **f** could be obtained by applying the inverse of the degradation function to the measured responses according to

$$\hat{\mathbf{f}} = \mathbf{H}^{-1}\mathbf{g}. \tag{2}$$

METHODS

Human Subjects

Subjects were tested following informed written consent, after approval from the Rutgers University Institutional Review Board. An initial screening exam was performed by questionnaire and by direct palpation of the forearm during requested finger motions. Criteria for acceptance into the study were (1) a trans-radial amputation, (2) the presence of afferent and efferent phantom activity, (3) palpable soft tissue movement in the limb.

Residual Kinetic Maps

Subjects were tested with a thirty-two-element pneumatic sensor array, spaced around the residual limb to detect the mechanical activity of muscle and tendon as the subject performed a volitional movement. The sensors, coupled to pressure transducers, served as inputs to a desktop data acquisition system. Each subject was requested to flex each finger for which he felt capable to move. The data was then interpolated to form a pressure contour map for each volitional movement.

Residual Kinetic Vectors

The procedure for obtaining the response matrix can be described as follows. Subjects were instructed to perform specific volitions while an 8-element pneumatic sensor array measured mechanical activity on the residuum. A brief phantom finger flexion of maximum intensity was modeled as an impulse to the system, where the component of **f** corresponding to the volition was assigned a magnitude of 1 and all other components were assigned a value of 0. A response vector was constructed from the maximum amplitude of the signal measured by each sensor channel. Alternatively, multiple repetitions of a particular volition could be performed and RMS values of the responses calculated to provide the response vector components [5]. Repeating the process for other independent volitions that the subject was capable of performing provided unique response vectors. The response vectors associated with specific volitions comprised the columns of the response matrix. Generally, the number of sensors in the array and the number of independent motions were different, so the response matrix was not square ($m \neq n$). Therefore, $\mathbf{H}^{-1}$ is the pseudoinverse of **H**, calculated by the singular value decomposition (SVD) of **H** [6].

Determination of the weights for the restoration function, or pseudoinverse (PI) filter, occured during an initial training procedure, which was the most computationally demanding step of the process. In contrast, pressure (input) vector filtering can be done in nearly real-time, with control (output) vectors updated before the next acquisition cycle.

RESULTS

Maximum pressure changes in the socket were approximately 4 kPa. Isobaric contour lines were evenly spaced from the maximum pressure for each volition. Figure 1 shows the residual kinetic maps for subject 1. Each map represents one volitional command over which the subject perceived control. Subject 1 believed he had control over three phantom fingers;

therefore we did not attempt to find additional flexion commands. Subject two had different, but equally unique residual kinetic maps for four fingers.

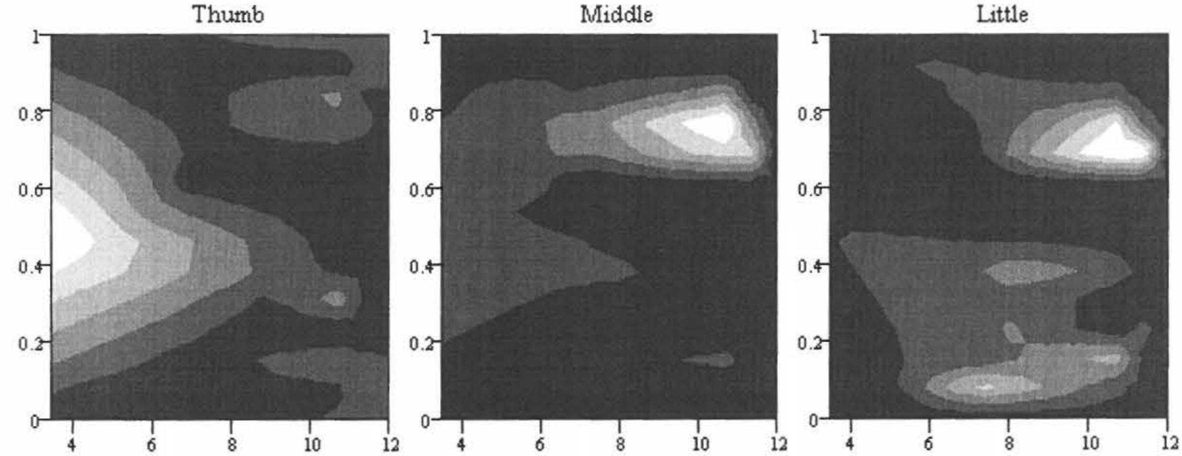

Figure 1: *Residual Kinetic Maps of Requested Finger Movements, Subject 1. These pressure contour maps of the residual limb demonstrate the uniqueness of patterns for different requested movements. The abscissa is the distance from the distal end of the residual limb in cm. The ordinate is the normalized circumference in the transverse plane, with 0 being the most lateral point, moving along the anterior surface.*

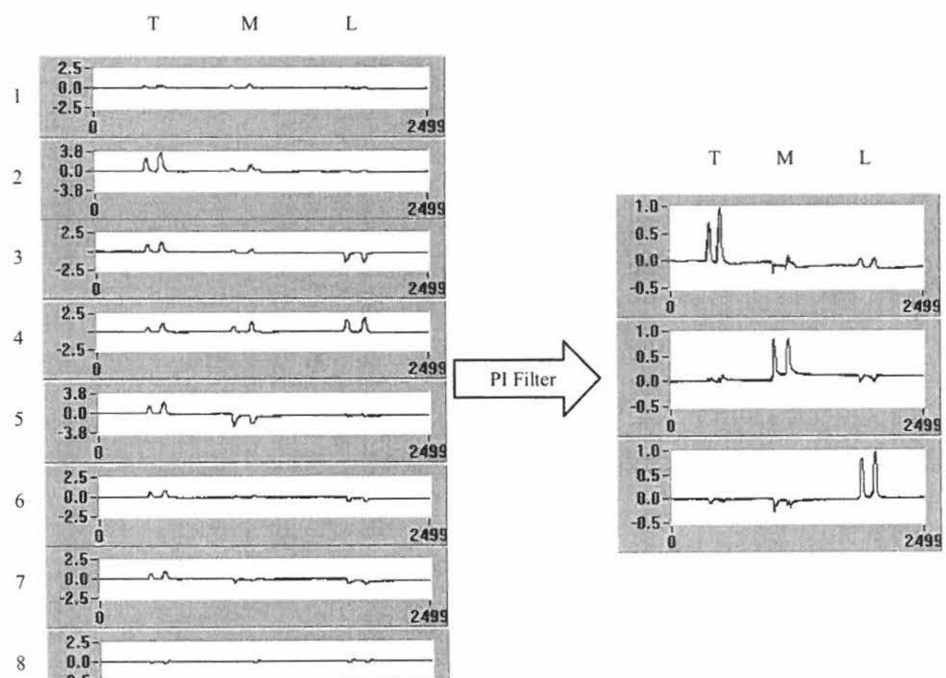

Figure 2: *Filtering Kinetic Signals. Eight channels measure distributed forelimb pressures during flexion of phantom digits, thumb (T), middle (M), and little (L). PI filtering discriminates intended volitions and provides usable control signals.*

Following training of the pseudoinverse filter, signals from the eight-channel array were applied to the filter for two consecutive taps of three different phantom digits. Figure 2 illustrates raw pressure measurements and filtered outputs obtained from subject one. Output signals appear suitable for proportional control of specified digits, with minimal coupling between channels. Both subjects could control three digits on the multi-finger prosthesis independently.

CONCLUSIONS

Phantom efferent phenomena can be represented graphically as a residual kinetic map or as a vector. By extracting the patterns from forelimb movement we have demonstrated that different volitions can be decoded from pressure vectors on the residual limb. Given that phantom sensation is common, we anticipate that a significant portion of trans-radial amputees would be suitable subjects. Furthermore, volitional movements not considered in this study, such as grasping, should also yield unique pressure vectors.

Concerns to be addressed in future work are sensitivity to sensor placement and external loads. Slight differences in donning necessitate the retraining of the algorithm. External loads appeared to be a possible intermittent issue with one subject. To compensate for the effects of external loads and donning imprecision, we are currently modifying the algorithm to include spatial information and developing a new sensor interface with greater spatial resolution.

ACKNOWLEDGEMENTS

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REFERENCES

1. Lin, L., Huang, H. "Mechanism and computer simulation of a new robot hand for potential use as an artificial hand," Artificial Organs, vol. 21., no. 1, 59-69, 1997.
2. Gallant, P.J., E.L.Morin, L.E. Peppard, "Feature-based classification of myoelectric signals using artificial neural networks," Med. Biol. Eng. Comput., vol. 36, 485-489, 1998.
3. Lee, S., Kim, J., Park, S. "An enhanced feature extraction algorithm for EMG pattern classification," IEEE Transactions on Rehabilitation Engineering, vol. 4, no. 4, 439-443, 1996.
4. Scheiber, M.H. "Muscular production of individuated finger movements: the roles of extrinsic finger muscles," The Journal of Neuroscience, vol. 15, no. 1, 284-297, 1995.
5. Curcie, D.J., Flint, J.A., Craelius, W. "Biomimetic finger control by filtering of distributed forelimb pressures," IEEE Transaction on Neural Systems and Rehabilitation Engineering, vol. 9, no. 1, 69-75, March 2001.
6. Strang, G. Linear Algebra and Its Applications, Third Edition. Harcourt Brace Jovanovich College Publishers, Fort Worth, 1988.

CONTROL AND SIGNAL PROCESSING CONCEPTS FOR A MULTIFUNCTIONAL HAND PROSTHESIS

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INTRODUCTION

The loss of an arm means drastic reduction of live quality for affected people. To compensate the lost live quality myoelectric hand prostheses have been developed, that can be controlled by muscle contractions in the patients stump. Besides cosmetical aspects the acceptance of these hand prostheses depends on functionality, weight and user-friendliness. Concerning these items, international studies demonstrated, that about 30% of the users of functional prostheses are not using them regularly [1, 2].

Recently, the FZK (Forschungszentrum Karlsruhe) presented a light-weight artificial hand, that is able to move all finger joints independently [3]. To control this large variety of movement possibilities the principal task is to establish a control that has an extended functionality but is easy to handle.

This paper presents:

- a control concept that is able to perform various grip types using the same sensor configuration like commercial products and
- a software platform to individually adapt necessary parameters to the users needs.

CONTROL CONCEPT

To gain information about muscle contraction activity the surface scan of myoelectric (EMG-) voltage is applied. Basically, a transformation of the scanned signal into a certain amount of grip types for a prosthesis requires the same amount of unique signal patterns. The major problem using myoelectric signal patterns is the patients' deficiency to contract more than two muscles independently. Additionally, the EMG-signal is no pure signal of one specific muscle but contains information about all contracted muscle fibres in the range of the sensor. Thus, signal quality is reduced drastically and the success of a discrimination of different signal patterns decreases. Several groups made promising attempts to increase the corresponding classification rate via preprocessing or sophisticated classification algorithms [4, 5], however no control using more than two to three movements is commercially available.

Thus, we propose a fragmentation of the patient's control signal in two phases. Therefore, the flexor muscle groups and the extensor muscle groups are scanned for myoelectric activation - equivalent to commercial prostheses [6] - and filtered with different IIR-filters to gain significant features [7].

Starting in a neutral state (prosthesis opened) the user may generate a switch signal by contracting one of the two scanned muscle groups. The characteristic of the signal specifies a certain grip type.

As soon as the system detects a known switch signal the prosthesis moves into a preshape state, serving

- to give a feedback to the user, which kind of grip type was identified and
- to move the fingers of the prosthesis into an optimal position concerning movement possibilities associated with the identified grip type.

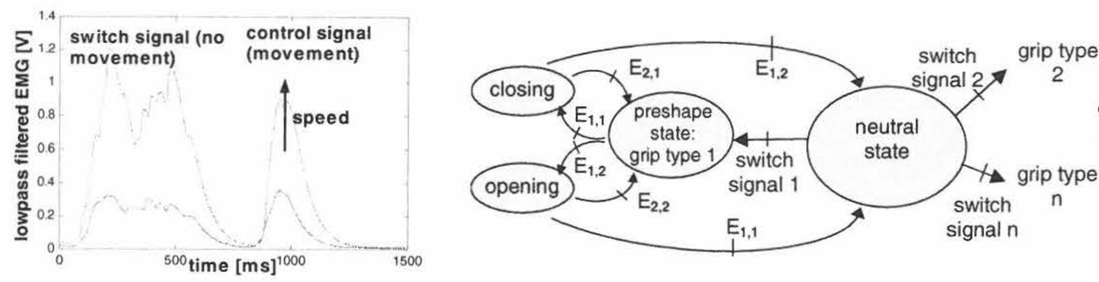

Figure 1: *left: EMG-signal for execution of a prosthesis movement, right: control scheme with control signals derived from two muscles: $E_{1,1}$: contracted flexor muscle group; $E_{2,1}$: relaxed flexor muscle group; $E_{1,2}$: contracted extensor muscle group; $E_{2,2}$: relaxed extensor muscle group.*

The muscle contraction following the switch signal represents the control signal for the movement and leads to a closing or opening speed of involved finger joints proportional to the signal's amplitude (proportional control). The direction of the movement (closing / opening) is specified by the contracting muscle group (flexor/extensor), see Fig. 1.

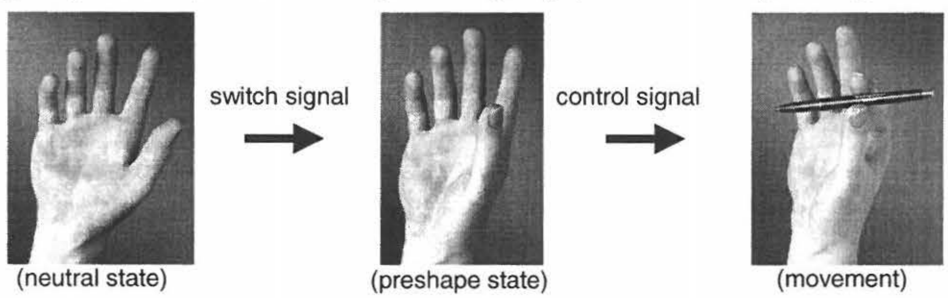

Figure 2: *FZK-prosthesis: Starting from a neutral state, a switch signal moves the joint angles in a preshape state. Depending on a control signal the chosen grip type may be opened or closed proportionally.*

A user-reset may be executed all the time by co-contracting both muscle groups and leads to the neutral state, opening all fingers.

At the moment, switch signals are taught to system by defining a membership between the value of several features and a certain grip type. For instance, an easy possibility to define switch signals would associate one muscle impulse with grip type 1, two impulses with grip type 2, etc. Using this scheme, the following grip types are implemented and can be performed by the patient:

- Cylindrical grasp,
- lateral grasp,
- bracing the pointing finger,
- hook grasp, and
- pincer grasp.

ADAPTATION

A negotiation of the presented algorithm with different subjects shows, that

- scanned signals differ in time and amplitude,
- locations of sensor positions influence signal amplitude and quality, and
- different muscles may have to be scanned with respect to the length of the arm stump.

To use a unified control scheme that is independent from the mentioned differences the algorithm has to be designed adaptable. Therefore a software platform was developed to validate the algorithm and its parameters, see Fig. 3. The aim was

- to create an easy and precise way to adjust parameters (e.g. filter constants) while the patient is practising,
- to give the patient the possibility to simulate the prosthesis, and
- to practise new parameters by controlling a video game (TETRIS).

When proper parameters are found, the control algorithm with adapted parameters is flashed on a microcontroller to process data online. Additionally, to record microcontroller input (sensory data) and output (control signal for prosthesis) an interface was integrated to gain information about signal differences between simulation and operation of the prosthesis.

The presented procedure was first implemented and tested with patients in May 2002 and was well accepted by the patient.

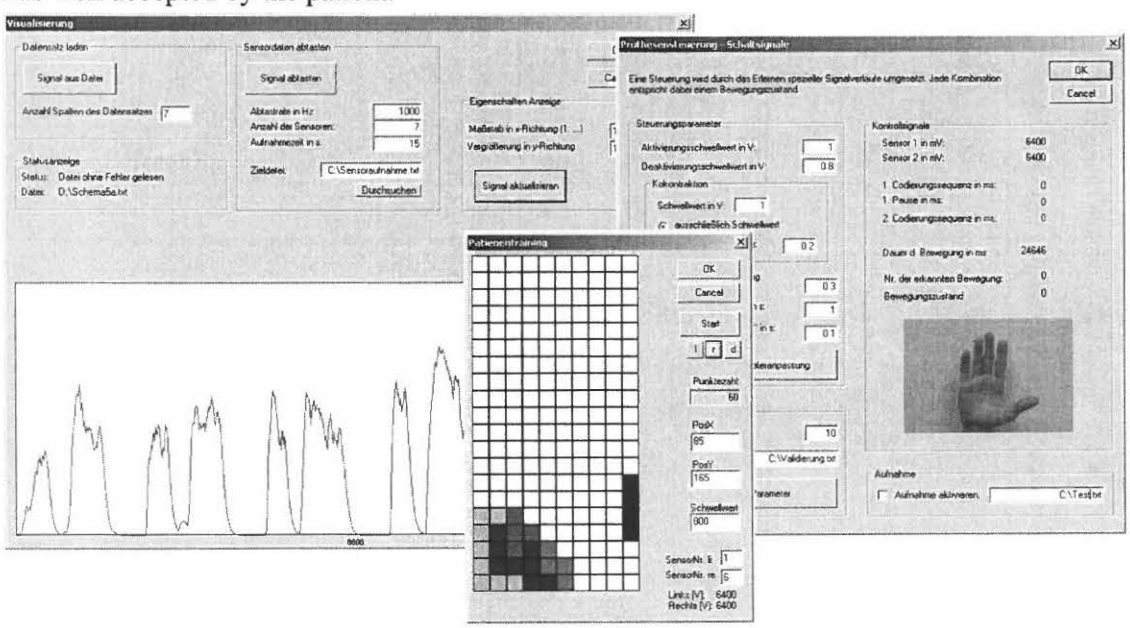

Figure 3: *software platform to adapt control parameters individually and to validate and test corresponding control strategies.*

DISCUSSION

The quality of a control scheme depends on the quality of the data it is processing. To rise data quality there is effort to adapt myoelectric sensors especially to the problems occurring with this control scheme and automatically detect optimal locations on the arm stump to scan for myoelectric data. Additionally, works including a teaching of switch signals to the prosthesis have started.

An upgrade of the presented control scheme is the use of fuzzy rulebases, automatically built out of relevant features of the myoelectric signal [8].

To gain improved signals and in this way to approach to the functionality of the human hand different groups are investigating brain and nerve activity [9, 10, 11].

CONCLUSION

This paper presents a new approach to multifunctional control of hand prostheses. The user has to generate switch signals in order to specify a desired grip type. The following muscle contraction serves to move the fingers proportional to the EMG-signal amplitude. This control scheme has already been tested with patients. To implement this strategy independent from diversities in anatomy and signal quality, a software platform was developed to adapt, validate and train the control strategy.

REFERENCES

1. Silcox, D.H., Rooks, M.D. et. al., "Myoelectric Prostheses", The Journal of Joint and Bone Surgery 75A(12), 1781-1791, 1993
2. Atkins, D.J., Heard, D.C.Y. & Donovan, W.H., "Epidemiologic Overview of Individuals with Upper-Limb Loss and Their Reported Research Priorities", Journal of Prosthetics and Orthotics 8(1), 2-11, 1996
3. Schulz, S., Pylatiuk, C. & Bretthauer, G., "A New Class of Flexible Fluidic Actuators and their Applications in Medical Engineering", Automatisierungstechnik 47(8), 390-395, 1999
4. Hudgins, B., Parker, P. & Scott, R., "A new strategy for multifunction myoelectric control", Proc. IEEE Transactions on Biomedical Engineering, 40(1), 82-94; 1993
5. Nishikawa, D., Yu, W., Yokoi, H. & Kakazu, Y., "Analyzing and discriminating EMG signals using wavelet transform and real-time learning method", C. H. Dagli, et al. (Eds.), vol. 9, 281-286, ASME Press, 1999
6. N.N.: Technische Daten Otto Bock Hand, tech. rep., Otto Bock, 2000
7. Reischl, M., Mikut, R., Pylatiuk, C., Schulz, S., Beck, S. & Bretthauer, G., „Steuerungs- und Signalverarbeitungskonzepte für eine multifunktionale Handprothese", Automatisierungs-technik 50(6), pp. 279-286, 2002
8. Reischl, M., Mikut, R., Pylatiuk, C. & Schulz, S., „Control strategies for hand prostheses using myoelectric patterns", Proc. 9th Zittau Fuzzy Colloquium, 168-174, 2001
9. Stieglitz,T., "Flexible biomedical microdevices with double-sided electrode arrangements for neural applications", Sensors and Actuators, vol. A90, 203-211, 2001
10. Pfurtscheller, G., Guger, C., Müller, G., Krausz, G. & Neuper, C., "Brain oscillations control hand orthosis in a tetraplegic", Neuroscience Letters, vol. 292, 211-214, 2000
11. Wessberg, J. et al., "Real-time prediction of hand trajectory by ensembles of cortical neurons in primates", Nature, vol. 408, 361 ff., 2000

MYOELECTRIC PROSTHESES WITH SENSORIAL FEEDBACK

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INTRODUCTION

One of the main problems of commercial myoelectric prostheses is that they do not have parts that can give back to the patient the sensorial feedback. None of them are able to produce sensations like pressure, which can give to the patient the information of the measure of the force in the final effector or temperature, which causes the prosthesis not to be totally controllable.

Although the scientific techniques and developments are more advanced in other countries, at this moment prosthetics systems like the one propose in this work are not commercially available anywhere. Furthermore, if at some point in a near future these systems become available most of the amputated patients in underdeveloped countries will economically struggle to acquire them due to low incomes.

This is the reason that motivated the development of this work, in which the objective is to solve the problem of sensation of the lost limb. The problem is solved through development of a microcontroller system for myoelectric prosthesis with sensorial feedback. This system allows any person who has been amputated, to replace the perception of the pressure of the amputated member. In addition, this development will allow to create a prosthesis of low cost and easy acquisition. This can be used like start point to other projects in the prosthetic area, such as the extension of the control system for complex prosthesis.

The work is divided in three parts, in the first chapter the design of the system as well as the minimum configuration of the design for the implementation of the hardware and the detailed analysis of the variables to manipulate is described. The second section presents the implementation of the system, hardware, software and tests conducted. In the last section the conclusions are drawn with comments and future developments included.

SYSTEM DESIGN

A. Control Type

The first step in the design of this system was selecting the type of control to use for the myoelectric prosthesis based on different methods [1,2,3,4,5,6,7,8,9]. In concordance with the objectives of this work a proportional control system was designed based on the amplitude of the surface myoelectric signal (SMES) implemented in Fuzzy logic. Through this proportional system the motor speed of the prosthesis, as well as the force, can be controlled according to the SMES amplitude generated by the patient. Due to the direct control on the speed and the force of grasp of the hand, the patient feels it more natural and easy to handle. The comfort increases as the effort diminishes.

B. General Description of the System.

In this design there are two input channels (figure 1), one for each muscle of the antagonistic pair. The number of channels that the circuit has depends on the number of antagonistic pairs and the amount of degrees of freedom that will be allowed in the prosthesis.

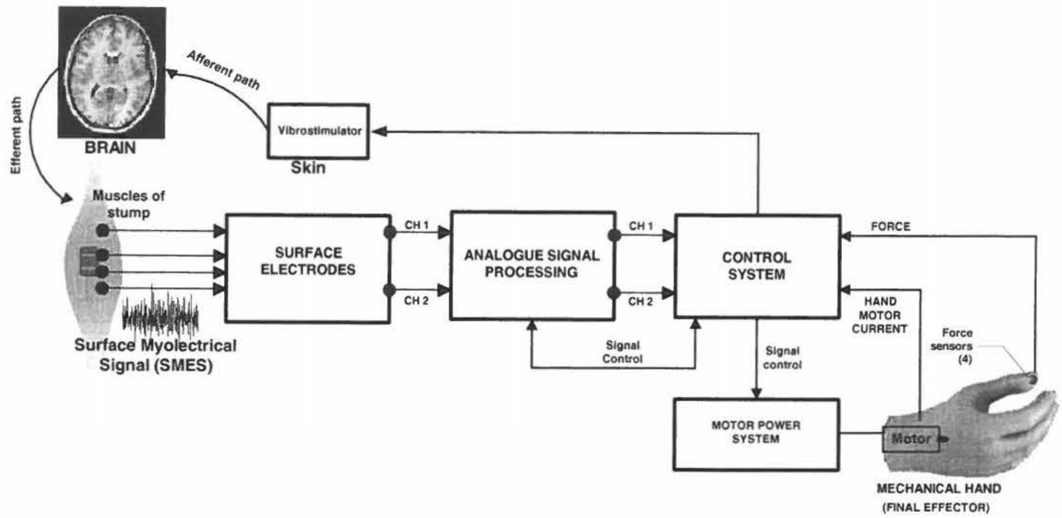

Figure 1: *General System Diagram.*

The surface electrodes are placed in the socket of the prosthesis where the amplitude of the measurement of the SMES is the highest for each antagonistic pair, in order to recollect SMES information correctly for every patient tested. The acquisition stage of the SMES is designed considering safety standards of electric risk for the patients. The SMES signal is processed in the control system, which generates the necessary commands to control of the prosthesis actuator [10,11,12,13,14]. The SMES is filtered to remove noise and avoid non-desired responses in the control.

In order to control the prosthesis response speed in movements going up or down, opening or closing the gripper, the same input signal of each one of the channels is used and it is compared with some reference levels, In this way for each input channel (by type of muscle and direction of movement) there are two speed actions, one high for a high amplitude, and a low one for a lower level of amplitude.

It has with 4 additional pressure sensors located on the surface of the artificial gripper to return feedback of pressure on the surface by the tips, on the held object. This information is sent to the control system and processed to give back to the patient the perception of pressure sensation of the amputated limb, by means of a vibrostimulator constructed.

A prosthesis prototype of upper limb to replace mid third level extremity was designed with a mechanical hand. At this level we can leave a great number of extremities functions and therefore easily extending the control system to other prosthesis with smaller index of complexity. Superior levels of amputation require the replacement of the elbow and/or the shoulder and its respective control system.

In order not to be concerned with importing foreign expensive component, special molds were developed to make at home the necessary prosthesis component, using polypropylene, a material easy to handle and to store. Its use does not require additional chemical agents, and another advantage is that it can be recycled. This drastically reduces the cost of the orthopedic apparatuses and allows that more amputated people may benefit from the technology.

The control system will be defined in terms of the variables of the force pressure and opening of the gripper, but they are portable to other artificial systems.

C. Position control in the amputated subject.

The feedback to set the position point to point to the amputated patient is almost impossible by means of propioceptive routes [15,16,17,18]. The only thing that can be done is a state of virtual propioception in which the patient notice the grip that is applied to the object and through to the experience he learns to move the prosthesis for proper subjection. Other way would be to maintain visual contact with the object after the sensation and to observe the change of the opening (figure 2).

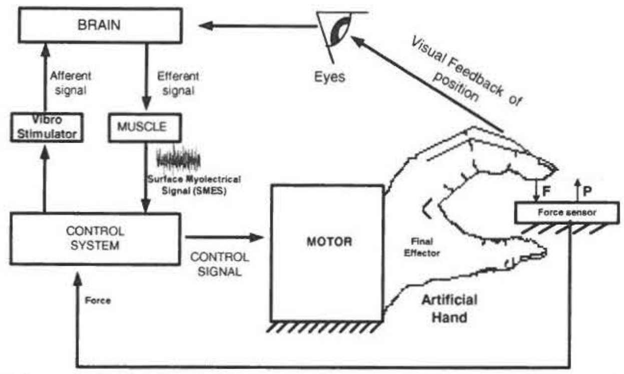

***Figure* 2:** Control system developed. Note: even though there is visual feedback, it is not necessary in most cases.

SYSTEM IMPLEMENTATION

A. Processing stage:

The control system was implemented in a microcontroller using fuzzy logic. The filters were programmed internally.

B. Pressure feedback Stage

The factors that determined the selection of the sensor were the following:
Similarity with the sensors of pressure of the body, easy implementation, size and price. This last factor is very important because if the used sensor is very expensive the prosthesis becomes too expensive to a greater number of patients and the sensation function would become "unattainable".

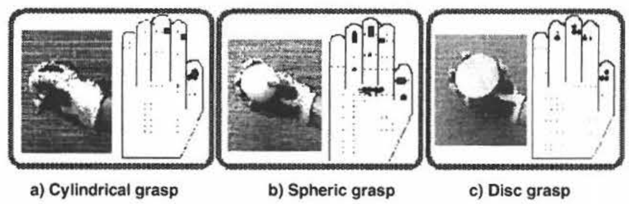

***Figure* 3:** Pressure map obtain by GUAN

Pressure sensors: Force sensor resistors (FSR) are used. These devices are made with thick layers of polymers, decreasing its resistance according to the applied force to the active surface.

Pressure sensor location in the prosthesis: Using the pressure map [19] of figure 3, it is possible to locate the pressure sensors, only disregarding the 4th finger, warranting the grip sensation for the mayor activities. Therefore the sensors are located this way: one in the tip of index finger, one in the thumb finger and two in the palm surface, in order to have a complete coverage of the holding area of the most important functions of the hand: Fingertip grasp, palmar grasp, spherical grasp and cylindrical grasp.

C. Feedback stage

A tactile vibrostimulator was developed, with the following characteristics:

- Minimal power consumption
- Maximum comfortableness during stimulation
- Minimum irritability post – skin stimulation
- Ample dynamic range: maximum level of comfort during sensorial trigger.

The tactile vibrostimulator oscillates to a frequency and amplitude proportional to the pressure made in the hand during the action of holding an object. This device evokes the tactile sensation by means of the mechanical stimulation to the skin with vibrations of frequencies between 10hz to 500hz [20]. The device was located in the socket of the prosthesis so that it made permanent contact with the skin of the stump of the amputated patient, due to the controllable ranges of frequency and amplitude it can be placed anywhere in the arm causing the same sensation.

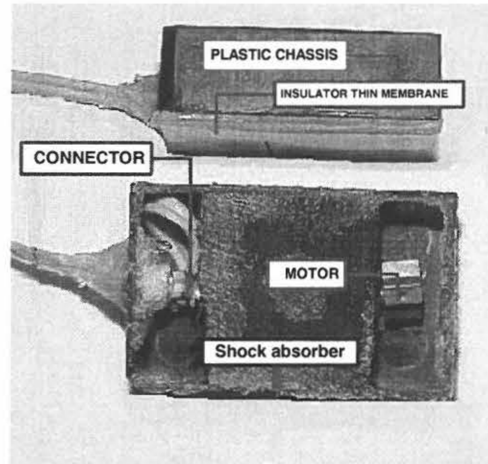

Figure 4: *Vibrostimulator design*

RESULTS

During the calibration stage, different tests with 30 amputated subjects of upper limb of forearm level were made (all the patients had only one arm amputated). To determine the best speed for the grip, measurements in the end of the fingers were made. That included the manipulation of several objects of common use: a pencil, a glass and tools of work as screwdrivers, hammers and keys. In addition tests were made during working periods, in which it was required of the combined manipulation of objects to carry out a specific task and a more complex one, like the elaboration of a breakfast. It was noted that for the totality of patients tested, 8,25 cm/sec is the minimum accepted for most of activities of " Holding" and "Releasing". During precision grips, in which required more attention, the speeds needed were between 1 cm/sec to 2 of cm/sec.

Once the prosthesis prototype was finished and calibrated, it was observe through practical use, with the amputated patients who used the system with sensorial feedback, that only visual control was necessary, in the cases where the weight or the characteristics of hardness of the object was not known, since it creates distrust to damage the object. There were tests made with the same patient were they had their eyes closed and they could hold objects without knowing its weight or forms only by feeling the pressed force of the sensorial feedback. The tests were made for cylindrical grasp and fingertip grasp. The used objects were utensils and tools of common use in the daily life, the weights vary from 10gm (pencil) to 2.2Kg (wine Bottle). Margin of maximum error is of 30% in

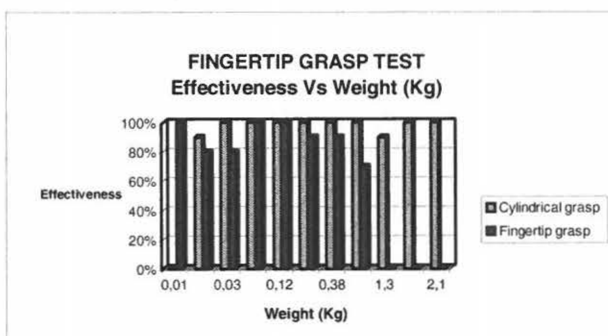

Figure 5: *Relation weight/effectiveness in the test of cylindrical and fingertip grasp (This function is recommended in the manipulation of light objects).*

the case of too heavy objects for the fingertip grasp (weight approximately > 1.5Kg) (figure 5).

The results show that the knowledge about the gripped object characteristics with each new attempt increases the effectiveness until reaching 100% in most of the cases in the fourth trial (figure 6).

The causes of failure in the attempts were: a) The object slips; if the weight is greater than the minimum grip force, b) The object is damaged; If the force is greater than the force of fracture of the element. In order to try to solve this problem it is recommended to use sensors of displacement in the surface of the hand that give to the amputated patient the capacity to know the minimum force to hold stable the object and that it is handled directly by the control system.

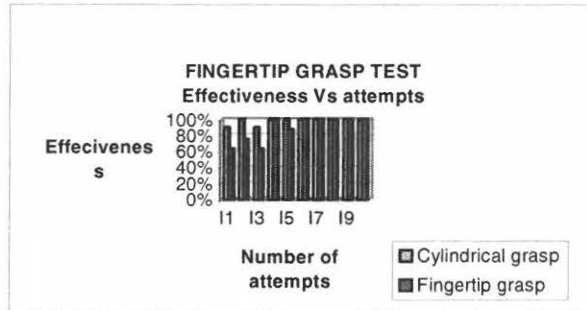

Figure 6: *Test results of fingertip grasp. The effectiveness increases as they increase the attempts. In the cylindrical grip effectiveness reaches a 100% between the third and fourth attempt.*

CONCLUSIONS

It was designed and implemented a control system for a prototype of myoelectric prosthesis, for forearm at third distal level. It allows the use of sensorial feedback to the patient for means of a vibrostimulator system of pressure signals that guarantee the manipulation of different objects with no need to sacrifice other functional abilities (vision). The implementation of the control system is very simple and economical because most of components are acquired locally.

The control system with sensorial feedback developed allows the amputated patient the following advantages from the conventional prosthesis:

- Low mental load. The patients during the tests could be making several activities and maintain grip an object without any problem
- Learning facility and friendly use. 100% the patients who used the system (10 patients) in the tests learned how to use it as soon as it was put on them for the first time.
- Independence in the multifunctional control. The control developed allows that a person with prosthesis in both arms was able to use each extremity independently.
- Direct access and instantaneous answer. All the functions are controlled in real time.
- Not sacrifice functional abilities. The control system designed eliminates in an 80% the used of visual control in the conventional prosthesis.

The control system can be used to handle any type of prosthetic mechanism as long as it fulfills the parameters within the controllable range.The implemented system gives a start for future investigations in the field of the myoelectric prosthesis in Latin America, and it is a first step for a complete arm at very low cost.

ACKNOWLEDGMENTS

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REFERENCES

1. Kelly, Michael; Parker, Philip y Scott, Robert. "Myoelectric signal analysis using Neural Networks". IEEE Engineering in Medicine and Biology Magazine. Vol. 9. No 1. Marzo 1990.

2. Neville Hogan and Robert W. Mann, "Myoelectric Signal Processing optimal estimation applied to electromyography part II derivation of the optimal myoprocessor". IEEE Trans. Biomed. Eng, Vol BME 27 No. 7 pp.382, 395,July 1980.
3. Daniel Graup, Jan Magnussen and Aloysius A. Beex, "A microprocessor system for multifunctional control of upper limb prostheses via myoelectric signal identification", IEEE trans, Automat. Ctr. vol AC- 203, No 4, pp 538 - 544, Aug 1978.
4. O' RIAN M.D., D.T. Gibbons, K.A. Reuben, "Extended Physiological propioception in a Microprocessor controlled prosthesis" Proceedings of international conference on rehabilitation engineering.
5. DUDLEY S. Childress, " Control of limb prostheses", Atlas of Limb Prosthetics, Surgical prosthetic and Rehabilitation principles. Part 6: The upper limp, Cap 6D, edited by American Academic of orthopaedic Surgeons, 1992.
6. BARRY D.T., J.A. Leonard, "acoustic Myography as a control signal for an externally powered prosthesis", arch. Phys. Rehab. Vol 67 pp 267- 269 April 1986.
7. BERGMAN K., L. Ornholmer, K Zackrisson, "Functional benefit of an adaptative Myoelectric prosthetic Hand compared to a conventional Myoelectric Hand", Vol. 16, pp.32-37. 1992.
8. JACOBSEN S.C.; Knutti D.G Johnson R.Ty Sears H.H. "Development of the UTAH artificial arm". IEEE Trans. on B.E. Vol 29. No 4. Agosto 1982.
9. SARIDIS, George y Lee, Sukhan. "The control of a prosthetic arm by EMG pattern recognition". ," IEEE Tran. On Automatic Control. Vol.29, No. 4, Abril. 1984.
10. Graupe D. y Cline W. K. "Functional separation of EMG signals via ARMA identification methods for prosthesis control purpose". IEEE Transaction on Systems, Man and Cybernetics. Vol 5 .N 3. March
11. Graupe D., Magnusse J. "A Microprocessor system for multifunctional control of upper Limb Prostheses and orthoses via Myoelectric Signal Identification". IEEE Trans. on Automatic Control. Vol 23 August 1978.
12. Hogan, Neville y Habul-Haj, Cary J. "Functional Assessment of Control System for Cybernetic Elbow Prostheses -Part I Description of the technique". IEEE Trans. on B.E. Vol 37. No 11. Nov 1990.
13. Hogan, Neville y Habul-Haj, Cary J. "Functional Assessment of Control System for Cybernetic Elbow Prostheses -Part II" Application of the technique". IEEE Trans. on B.E. Vol 37. No 11.
14. Hogan, Neville y Mann, Robert W. "Myoelectric signal processing: Optimal estimation applied to electromyography. - Part- I: Derivation of the optimal myoprocessor". IEEE Trans. on B.E. Vol 27. No 7.Julio 1980.
15. S.B. KANG and K. Ikeuchi, "A grasp abstraction hierarchy for recognition of grasping tasks from observation," Proc. IEEE/RSJ Int'l Conf. on Intelligent Robots and Systems, Yokohama, Japan, July 1993.
16. S.B. KANG and K. Ikeuchi, "Grasp recognition using the contact web," Proc. IEEE/RSJ Int'l Conf. on Intelligent Robots and Systems, Raleigh, NC, July 1992.
17. S.B. KANG and K. Ikeuchi, "Toward automatic robot instruction from perception: Recognizing a grasp from observation," IEEE Int'l Journal of Robotics and Automation, vol. 9, no. 4, Aug. 1993.
18. S.B. KANG and K. Ikeuchi, A framework for recognizing grasps, Tech. Rep. CMU-RI-TR-91-24, Carnegie Mellon University, Nov. 1991
19. Y. Guan and H. Zhang "Kinematic Graspability of a 2D multifingered hand", Department of computing science, University of Alberta, Canada 2000
20. KURT A. Kaczmarek, " Electrotactile and Vibrotactile Displays for sensory substitution system" IEEE Trans. on B.E. Vol 38. No 1. Jan 1991.

NEUROFUZZY LOGIC AS A CONTROL ALGORITHM FOR AN EXTERNALLY POWERED MULTIFUNCTIONAL HAND PROSTHESIS

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INTRODUCTION

We are developing a controller for a multifunctional hand prosthesis based upon multiple surface electromyograms (sEMG) using neurofuzzy logic technology. The sEMG signal is successfully used as a means of control in current commercially available myoelectric prostheses. However, these are either single degree-of-freedom (DOF) devices or sequential controlled devices with locking mechanisms to switch between DOFs. There have been several proposed algorithms of control extended to multiple DOF prostheses. Early myoelectric prostheses, such as the Sven Hand [1] and the Philadelphia Arm [2], involved the use of electrode arrays and used adaptive weighted filters to process the signals. More recently, Hudgins et. al [3] proposed extracting several parameters out of the first 200 ms of EMG activity in an effort to obtain a higher ratio of controllable functions to inputs. Other algorithms have also proposed control based upon pattern classification / feature extraction [4, 5] with the use of neural networks for system training [6], while others, specifically for hand control, have been based upon spectral analysis of the EMG signal [7]. These have met with varying degrees of success, but have for the most part been limited to laboratory success, and to our knowledge, have not been demonstrated as clinically practical solutions to the multifunctional control problem.

We propose an algorithm based upon neurofuzzy technology. We believe that because of the inherent "fuzziness" of human activity, a control algorithm based on fuzzy logic may have advantages for multifunctional prosthesis control. We seek an acceptable compromise between the number of electrode sites used and processing complexity, and thereby desire not more than three to four control sites to control three to four DOF. This approach delivers more information to the system and, by using fuzzy logic, reduces the complexity of the processing.

BACKGROUND

Fuzzy set theory maintains that all elements have a degree of membership (DOM) in all sets ranging from 0 to 1 inclusively. Figure 1 shows the difference in approach between fuzzy and classical set theories applied to the seasons of the year. A fuzzy logic system is composed of membership functions (MF) that fuzzify inputs, a rule set that governs how the system operates (see Table 1 for an example), and an output membership function that defuzzifies the system output. The advantage of such a system is that it is robust enough to handle small inaccuracies in data without significantly affecting the output. The system is clearly defined and modifiable, but not trainable.

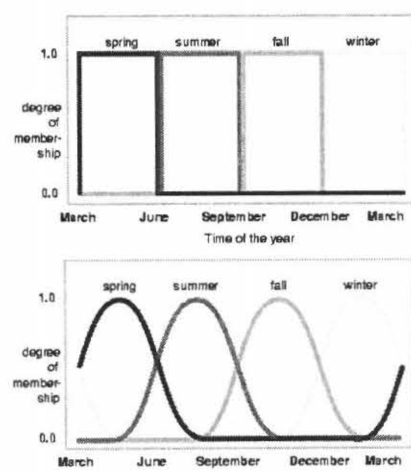

Figure 1: *Seasons of the year according to classical set theory (top) and fuzzy set theory (bottom)*

Artificial neural networks (ANN) allow for system training based upon sample input data. As sample data is fed into the system, outputs are calculated by associating a degree-of-strength (DOS) to each input. The error difference between the desired and actual output is then back propagated through the system, and the DOS are adjusted accordingly. This learning process is characterized by the choice of learning method and the number of winner neurons, where the number of winner neurons is the number of DOS altered during one iteration. Such a system is advantageous because of this ability to be trained on sample data. However, this system is essentially a black box because it is not modifiable, nor are the parameters of the system intuitive.

Neurofuzzy logic attempts to combine the advantages of a fuzzy and a neural net system to present a system with clearly defined variables and parameters but with the ability to be trained. The fuzzy inputs are mapped to ANN input nodes, the fuzzy outputs to ANN output nodes, and the fuzzy inference system to hidden nodes, with connecting lines serving as degrees- of-strength (DOS) for the rule sets. As the ANN is trained, the DOS of the fuzzy system are altered, along with the parameters of membership functions, using the standard back propagation algorithm. For more information about fuzzy logic, ANN, and neurofuzzy technology, see [9, 10].

EXPERIMENTAL PROTOCOL

Muscles controlling wrist extension, wrist flexion, ulnar deviation, and finger flexion are located using standard clinical myotesting techniques. In ideal cases, we search for the extensor digitorum, extensor carpi muscle group, flexor digitorum superficialis, and flexor carpi muscle group. The main focus was to find three to four independent control sites. Other motions of interest were pronation, supination, and radial deviation. Upon location, a standard sEMG Ag/AgCl differential electrode is placed over each muscle in question.

All EMG data is collected using Noraxon's telemeter Telemyo 8 System. The system records data at a rate of 1400 Hz, and applies a band pass filter (16 Hz low cutoff, 500 Hz high cutoff, gain of 2000) to the analog signals. For our purposes, individual gains are adjusted to obtain greater isolation of individual sites. The raw EMG signal, along with the associated signal RMS value (64 length sample collected at 1400 Hz), are collected, viewed, and saved in real-time on a PC computer running Labview 5.1.1. Ten seconds of quiescent data is simultaneously collected from all channels for reference. The subject is then asked to perform eight trials of each motion associated with each detected control site. Data is collected from the time prior to motion initiation until several seconds of steady state data are taken. Each file containing RMS values of the collected EMG signals is analyzed to determine the time instant of motion initiation by determining the first point in a series of five consecutive RMS points that are at least three times the quiescent RMS standard deviations (σ_0) above the quiescent RMS mean (μ_0). The initiation instants are verified through offline visual inspection and minor adjustments are made as needed. The data used to check the final fuzzy system is composed of all trials of all motions. The data used to train the initial fuzzy system is composed of the odd numbered trials of all motions. This procedure is followed to capture the effects of time and possible fatigue during the duration of the protocol. The results presented are for one subject without limb loss and one with a short trans-radial limb loss from birth.

FUZZY SYSTEM DESIGN

The fuzzy logic system is designed using Inform Technologies' fuzzyTECH ver 5.52 MCU-320 edition development environment. This environment was chosen due to its ability to automatically convert designed systems into MATLAB, C, and assembly code for easy future implementation into a microcontroller of choice. Each input to the system consists of three membership functions ("low", "medium", "high"), where the points of intersection are determined by visual inspection of the "checking" data. The typical operating point of the corresponding input when its associated muscle function is isolated is chosen to be the value corresponding to unity membership in the "medium" membership. Following conventional fuzzy membership design, the "low" and "high" membership functions are then constructed such that there is fifty percent overlap between adjacent membership functions.

The fuzzy rules implemented are designed to cover all scenarios of EMG amplitude from inputs demonstrated by the "checking" data set. Thus, rules are first added for the case of complete input isolation (one and only one input is active). The next set of rules is added to cover cases of incomplete input isolation (more than one input is active). Because of the stochastic nature of the EMG signal, there exist cases where, upon initial observation, the same set of linguistic inputs can produce different outputs. Thus, conflicting rules were also added into the initial fuzzy classification system, leaving resolution of these conflicts to the training process. The DOS of all rules are set to zero initially, effectively producing an unintelligent system. Using the back error propagation technique for system training, we looked at inclusion vs. exclusion of automatic membership function learning in the training process, and the effect of altering the number of winner neurons. The system with the minimum average error for each set of training parameters was saved. The same parameters that produce the highest accuracy for the normal subject data are then used to train data for the subject with limb loss.

RESULTS

Figure 2a shows the percent accuracy for all training parameter sets for data of the subject without limb loss. The system that produced the most accurate classification was the one that was allowed to modify the membership functions and the number of winner neurons was limited to one, resulting in a 99.15% classification rate. Figure 2b shows this system to be optimized after 85 iterations. Using the same parameters to train the system designed for data of the subject with limb loss, we achieved an accuracy rate of 96.57%.

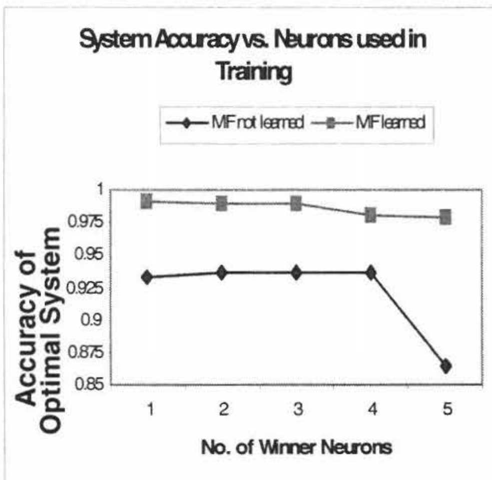

Figure 2a: Accuracy for each set of training parameters (MF = membership function).

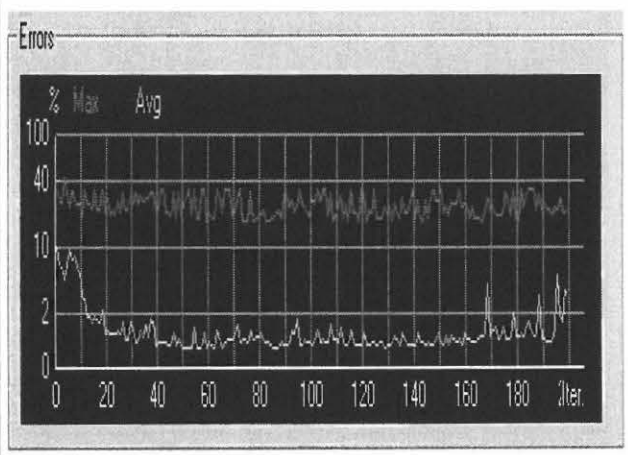

***Figure 2b:** Error during training for the most accurate system. The system converges to the minimum error after 85 iterations. The results of* over training can be seen after 160 iterations of training

Table 1: *Final rule set and associated DOS for limb deficient subject data. The DOS determine the relative contribution of each rule to the system response.*

IF			THEN	
emg1	**emg2**	**emg3**	**DoS**	**classification**
Low	low	medium	0.92	motion3
Low	low	high	0.95	motion3
Low	medium	low	1.00	motion1
Low	high	low	0.77	motion1
medium	low	low	0.36	motion0
high	low	low	0.46	motion0
high	medium		0.13	motion0
medium	medium		0.09	motion0
medium	high		0.06	motion1
medium		high	0.09	motion3
Low	low	low	0.01	off

DISCUSSION

The high accuracy of classification for the subject without limb loss is encouraging in terms of possibly using this algorithm in a clinical prosthesis. There were several trials of motions where the neurofuzzy system achieved excellent classification for the entire duration of the motion. This implies that it is possible for the classifier to be effectively trained and thus extremely accurate – the only requirement would be that the user would need to learn to repeatedly perform the desired motion in the same manner.

The classifiers presented here are limited in that they only allow for one DOF to be active at a time. However the concepts demonstrated can be extended into simultaneous classification (recognizing combinations of motions). This should be readily achieved through the addition of membership functions to the output and the implementation of another rule set that would cover the scenarios of simultaneous activity of multiple DOFs. The system would then again need to be trained with sample input scenarios.

Since the proposed classifier system makes a decision at time *t* based solely upon RMS EMG information at time *t*, the conversion to a real-time system is straightforward. In addition, the short computation time required to process a fuzzy inference system is an advantage of such systems. This advantage is magnified by the fact that there now exists hardware dedicated to processing of fuzzy inference systems, such as the Motorola 68HC12 microcontroller family [11]. With further research, it is felt that the implementation of neurofuzzy technology as presented here may be applicable to practical clinical devices.

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REFERENCES

1. Lawrence, P. D., and Kadefors, R., "Classification of Myoelectric Patterns for the Control of a Prosthesis", Proceedings of the Conference on the Control of Upper-Extremity Prostheses and Orthoses, Göteborg, Sweden, pp. 190-200, October 6th-8th, 1971, In: The Control Of Upper-Extremity Prostheses and Orthoses, Herberts, P., Kadefors, R.,

Magnusson, R. I., and Petersén, I., (Editors), Charles C. Thomas, Springfield, Illinois, 1974.
2. Taylor, D. R., and Wirta, R. W., "Development of a Myoelectrically Controlled Prosthetic Arm." Proceedings of the third International Symposium on External Control of Human Extremities, Dubrovnik, Yugoslavia, August 25-30, 1969, In: Advances in External Control on Human Extremities, Yugoslav Committee for Electronics and Automation (ETAN), Belgrade, Yugoslavia, 1970.
3. Hudgins B, Parker P, & Scott RN, "A new strategy for multifunctional myoelectric control", IEEE Trans. Biomed. Eng., vol. 40, no. 1, 82-94, 1993.
4. Saridis GN & Gootee TP, "EMG pattern analysis and classification for a prosthetic arm", IEEE Trans. Biomed. Eng., vol. 29, no. 6, 403-412, 1982.
5. Dening DC, Gray FG, & Haralick RM, "Prosthesis control using a nearest neighbor electromyographic pattern classifier", IEEE Trans. Biomed. Eng., vol. 30, no. 6, 356-360, 1983.
6. Gallant PJ, Morin EL, & Peppard LE, "Feature-based classification of myoelectric signals using artificial neural networks", Med. Biol. Eng. Comput., vol. 36, 485-489, 1998.
7. Farry K, Walker I, & Baraniuk R, "Functional separation of myoelectric signals using Thompson's multiple window method", MEC '95, Fredericton, New Brunswick, Canada, 57-66, Aug 1995.
8. MATLAB ver 6.0 Fuzzy Logic Toolbox Help ver. 2.1
9. von Altrock C, Fuzzy Logic and NeuroFuzzy Applications Explained, Prentice Hall PTR, New Jersey, 1995.
10. Ross, Timothy J., Fuzzy Logic with Engineering Applications, McGraw-Hill, New York, 1995.
11. Drake P, Sibigtroth J, von Altrock C, & Konigbauer R., "Demonstration Model of fuzzyTECH® Implementation on M68HC12", Motorola AMCU Division (Austin, TX), Inform Software Corp. (Chicago, IL), AN1295/D, August 1996

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