CONFERENCE PROCEEDINGS

MEC '99

Narrowing the Gap

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

August 25 -27, 1999
Fredericton, NB, Canada

Hosted by:
INSTITUTE OF
BIOMEDICAL ENGINEERING

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MyoElectric Controls/Powered Prosthetics Symposium
Fredericton, NB CANADA
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Welcome from the MEC '99 Organizing Committee

Welcome to **MEC '99**. 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 three speakers in addition to scientific papers and commercial exhibits and presentations. The three speakers include: ***Mary Point Novotny***, President of the Amputee Coalition of America, will focus on the impact of consumerism on prosthetic technology; ***David Gow***, Director of Rehabilitation Engineering Services at Princess Margaret Rose Orthopaedic Hospital in Edinburgh Scotland, will speak on the development of the Edinburgh Modular Arm System; and ***Adele Fifield***, Director of Prosthetics and Counselling and Director of the National Amputee Centre for the War Amps of Canada, will speak on the needs of amputees in terms of prosthetics.

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 Sheraton for the wine and cheese reception on Tuesday August 24 from 7 to 10 p.m. and also for the Banquet on Thursday August 26th at 6:30 p.m.

We hope you enjoy **MEC '99** and your visit to Fredericton. Please don't hesitate to ask if we can help in any way during your stay.

MEC '99 ORGANIZING COMMITTEE

Dennis Lovely, *Director*
Greg Bush
Robert Caldwell
Kristel Copp
Angela Hamilton
Bernie Hudgins
Dinah Stocker

VENDORS PRESENT WILL DISPLAY PRODUCTS FROM:

VASI (Variety Ability Systems Inc.)
Scarborough, ON Canada

Liberty Technology
Hopkinton, MA USA

Motion Control
Salt Lake City, UT USA

International Prosthetic & Orthotics
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Campbell, CA USA

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of Canada
Oakville, ON Canada

Maryland Electrolimb
Silver Spring, MD, USA

FINANCIAL SUPPORT

The Institute of Biomedical Engineering and the *MEC '99* Organizing Committee would like to recognize the following organizations for their contributions to the symposium:

The Fredericton Foundation

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University of New Brunswick

War Amputations of Canada

The Whitaker Foundation

Thank you for making this symposium a success.

The Conference Proceedings were sponsored by:

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WEDNESDAY, AUGUST 25, 1999

8:45 am — *Welcome to **MEC '99***

9:00 am — **Keynote Address:**
Mary Point Novotny,
Amputee Coalition of America, Knoxville, Tennessee

10:00 to 10:30 am — *Refreshment Break*

10:30 am to 12:10 pm — **Studies:**

A Comparison of Upper Limb Prostheses users in Europe
P.J. Kyberd

Prosthetic Management of a Child with Phocomelic Deficiency
H. Bauer-Hume, S. Wierzba, J. Bishop, and A. Woo

A Pilot Study Examining Repetitive Strain Injury in People with Limb Loss
G.L. Neufeld and D. Stocker

Fitting the Humeral Level Brachial Plexus Amputee with Externally Powered Myoelectric Control
G. Stevens and T. Farnsworth

12:10 to 1:30 pm — *Lunch*

1:30 to 2:45 pm — **Outcome Measures:**

Using Computers to Make Outcome Measurement Easier (PUFI)
F.V. Wright, S. Hubbard, J. Jutai, S. Naumann, and R. Schuller

Assessing the Satisfaction of Young Users of Myoelectric Prostheses
F. Routhier, C. Vincent, L. Desaulniers, and M.-J. Morissette

Quantifying Impaired Hand Function in the Clinical Environment
C.M. Light, P.H. Chappell, and P.J. Kyberd

2:45 to 3:15 pm — *Refreshment Break*

3:15 to 4:55 pm — **Clinical Reports:**

Experience with Hierarchically Controlled Hand Prosthesis
P.J. Kyberd

Experience with the Rimjet Locking Humeral Rotator
J.E. Uellendahl and C.W. Heckathorne

Experience with Custom Silicone Suspension Sleeves for Self Suspending Trans-Humeral and Trans-Radial Prostheses
M. Broomfield and W.G. Dykes

Clinical Application of Roll-on Sleeves for Myoelectrically Controlled TR and TH Prostheses
W. Daly and T.W. Williams, III

5:15 pm — *Shuttle Bus from UNB to Sheraton*

THURSDAY, AUGUST 26, 1999

8:45 am — *Daily Notices*

9:00 am — **Keynote Address:**
David Gow,
Princess Margaret Rose Hospital, Edinburgh, Scotland

10:00 to 10:30 am — *Refreshment Break*

10:30 am to 12:10 pm — **Intelligent Control Systems:**

A Totally Modular Arm Prosthesis
P.J. Kyberd

The Development of an Advanced Multi-Axis Myo-Prosthesis and Controller
C.M. Light and P.H. Chappell

Teleassistance for Electronic Components in Rehabilitation Technology
A. Davalli, R. Sacchetti, and P. Ferrera

Forequarter Prosthesis with Interchangeable Elbow, Forearm, and Hand
R.D. Lipschultz

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1:30 to 2:45 pm — **Current Research:**

Optimal Fixed Wrist Alignment for Below-Elbow, Powered Prosthetic Hands
J. Landry and E.N. Biden

Magnetic Resonance Imaging of Congenitally Deficient Upper Limbs
K.A. Farry, L.K. Kramer, R.K. Gupta, D. Atkins, and W.H. Donovan

The Use of the Hilbert Transform in EMG Analysis
S. Taffler and P.J. Kyberd

2:45 to 3:15 p.m. — *Refreshment Break*

3:15 to 4:55 p.m. — **Myoelectric Control and Prostheses:**

Fuzzy Logic in the Interpretation of EMG Signals for Prosthesis Control
S. Tafler and P.J. Kyberd

Natural Control of Key Grip and Precision Grip Movements for a Myoelectric Prostheses
M.C. Santa-Cruz, R.R. Riso, and B. Lange

A PC Based System for Selecting and Optimizing Myo Controls to the Patient's Needs
G. Haslinger

Low Level Response to Bock and Steeper Electrodes
C. Wallace, T.W. Williams, III, and N. Taneja

5:15 pm — *Shuttle Bus from UNB to Sheraton*

FRIDAY, AUGUST 27, 1999

8:45 am — *Daily Notices*

9:00 am — **Special Guest Speaker:**
Adele Fifield,
War Amputations of Canada

10:00 to 10:30 am — *Refreshment Break*

10:30 am to 12:10 pm — **Programmable Controllers:**

Programmable Control: Technical Aspects
I. Kurtz, W. Heim, H. Bauer-Hume, S. Hubbard, and S. Ramdial

Programmable Control: Clinical Experience at Bloorview-MacMillan Centre
I. Kurtz, W. Heim, H. Bauer-Hume, S. Hubbard, and S. Ramdial

New Clinically-Useful Multi-Device Control Strategies Made Possible by the VariGrip II Controller
C. Wallace and T.W. Williams, III

Clinical Experience with Programmable Controllers
G. Stevens and T. Farnsworth

12:10 to 1:30 pm — *Lunch*

1:30 to 2:45 pm — **High Level Fittings:**

Fourquarter Amputation: A Self-Suspending Shoulder Cap
D. Allen and W.G. Dykes

Adaptation of Locking Shoulder Joints to Increase Functional Range of Motion for Bilateral Upper Limb Deficiencies
G. Stevens and T. Farnsworth

MicroFrame Interface Design for High Level Myoelectric Prostheses
J. M. Miguelez

2:45 to 3:15 pm — *Refreshment Break*

3:40 to 4:55 pm — **Vendors Forum:**
This last session of MEC '99 will be devoted to any vendor that wishes to talk, show or comment on any aspect of prosthetic components

Participation by representatives from Hugh Steeper, Otto Bock, Liberty Technology, VASI, and Motion Control are welcomed.

5:15 pm — *Shuttle Bus from UNB to Sheraton*

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THE IMPACT OF CONSUMERS ON PROSTHETICS & ORTHOTICS

Mary P. Novotny RN, MS
President Amputee Coalition of America
Director National Limb Loss Information Center

Attempts to find artificial substitutes for missing extremities can be traced back through the history of man. A variety of devices including carved crutches and bent knee devices have been identified from earliest records. One of the oldest artifacts in existence is on display at the Royal College of Surgeons in London, a device dating back to the Samite Wars in 300 BC.

Looking into the history of prosthetics, there is no doubt consumers are the "raison d'être" for its development and ongoing progress to the 20th century. Evidence of devices constructed as a result consumers' determination to find suitable replacements begins with the prosthesis worn by Goetz Von Berlichingen in the early 1500's. This man, who lost his hand in the siege of Land Shut in Bavaria, requisitioned a replacement limb be constructed by an armorer. He is succeeded by a long line of inventors who were either discouraged with their inability to locate appropriate devices or dissatisfied with the lack of existing prostheses to meet personal expectations for performance and comfort.

There is no doubt that the ingenuity and creativity of consumers who needed replacements for missing limbs were key to adaptation and the quality of survival. Because the amputee was the master of the situation he controlled the people providing care. In the Classical and medieval periods limbs were secured from armorers, artisans, craftsmen and carpenters. Even during the Industrial period where amputation rates increased due to destructive wars and machinery there was little skilled prosthetic care available and amputees often sought out, designed and constructed their own devices.

In the United States, it was not until the post Civil War era, fueled by the desperation of amputee veterans, that the Union army began investing in programs to provide artificial limbs. This investment in prosthetics was so great that it exceeded the cost of the entire war effort of the Union Army. Later, not only was the survival of amputees greatly increased, so also was the involvement of the federal government and business in developing formalized organizations and educational programs to provide a structure administration and delivery of prosthetic care.

More recently, we have experienced a shift in the development and delivery of care. The focus of prosthetics in many areas has become an aberration of what has occurred for centuries. Instead of the amputee being central to decision making and master of his future, he finds himself objectified by a systems of "managed healthcare" where his options are dictated by "experts". Rather than being the focus of the TEAM, in many instances, the consumer finds himself outside the "inner circle" of influence. Instead of being the controlling factor, the amputee often lacks advocates for a voice in decision making.

As we move into the 21st century, it is incumbent upon professionals to learn from history. As health care providers we must recognize and appreciate the complex interaction between art, science and humanistic aspects of caring. It means making the amputee the captain of the TEAM as the central decision-maker regarding what is needed, based on the degree to which any device will be incorporated into his/her own body image and lifestyle. As in medicine, where the focus should be on treating the patient rather than the disease, let us remember that in prosthetics the goal is not merely replacing a limb but rebuilding a life! Who better than the consumer should be central to identifying the goals and determining what will be measured as a satisfactory outcome of that plan for his replacement limb and his life?

A COMPARISON OF UPPER LIMB PROSTHESES USERS IN EUROPE

Peter J. Kyberd, Oxford Orthopaedic Engineering Centre, Nuffield Orthopaedic Centre, Oxford.
OOEC, NOC, Headington, Oxford, OX3 7LD.
David Gow Director of Rehabilitation Engineering Services, Edinburgh, UK.
Helen Scott, Senior OT, Princess Margaret Rose Hospital, Edinburgh, UK.
Malcombe Griffiths Branch Manager RSL Princess Margaret Rose Hospital, Edinburgh, UK.
Lena Sperling, Leif Sandsjö, Christian Almström, Constanze Wartenberg - LindholmenUtveckling, Göteborg, Sweden.
Stewe Jönsson,- Sahlgrenska University Hospital, Göteborg, Sweden.

INTRODUCTION

An important part of the design process is to be able to identify the needs of a target population. Once incorporated into the designs the information can then be incorporated in to the measurement of the success of the design. This study forms part of the design phase for an advanced artificial arm system (TOMPAW).

The survey sought to assess the attitudes and requirements of the user who attend the centres were the project takes place. It was a postal questionnaire sent to adult users at three centres. The questionnaire consisted of both open and closed questions asking the subjects about their use and attitudes to their prostheses. The study builds on an earlier survey conducted in the UK and Italy [1,2] and it enabled a wider comparison to be made.

METHOD

The original survey was designed after interviews with ten users who routinely attended the Oxford Limb Fitting Clinic. Subsequent modifications were carried out after discussions. A pilot phase was initiated. The questions were presented orally to a number of clinic attendees and their responses used to hone the questions. Ten questionnaires were then posted to individuals. The envelope contained a covering letter, the questionnaire, and a prepaid return envelope.

The questionnaire was designed to be answered easily and quickly by the least technically confident person, it was kept to a minimal length and was divided into two sections. The first section used closed questions about the circumstances of the device and it was this section that was used in the current survey and is detailed in this paper. The intention of the questions was to provide both a comparable data set, yet to allow the user to express any feeling about the fit, function or use of their various prostheses.

The pilot had a 50% response rate and the results showed no substantive changes were required. Another 70 subjects in Oxford were chosen, and the questionnaire was dispatched to them. Additionally 35 from the INAIL Centro Protesi in Budrio Italy, were given a version of the survey.

In 1998 this survey was then presented to users attending the clinics at Sahlgrenska University Hospital in Göteborg and the Red Cross Hospital in Stockholm as well as to users who attend the centre in the Princess Margaret Rose hospital in Edinburgh, Scotland. While a larger, more detailed, second was presented to this population, the first section remained common to all surveys.

This comparison will be predominantly concerning the UK and Sweden populations, but will use data from the earlier survey where it is available.

RESULTS

The response to the survey was very positive (75% in 1998 and 65% in 1992) with the same sort of response rate for all centres. This compares favourably with published figures that range from 10 to 50% [3]. By the nature of the selection procedure the respondents are heavy users of their prostheses (Table 1), but many use cosmetic prostheses most often. Both groups use their devices for work, leisure and driving (Table 2).

However the spread of tasks they report using their devices is significant. The differences *between* different countries is far less than between centres, suggesting the differences are more due to local variations, such as the provision of different types of prostheses etcetera.

POPULATION

Cause and Gender

There is a greater number of males in both survey populations. An explanation would be that this reflects a larger number of males that loose limbs through accidents. However, if the numbers for the individual centres are observed, Edinburgh has a propotionatly higher number of persons with congenital loss, making the spread more even at about 50-50 compare to those from other centres (at about 70-30).

Side of loss

These are very evenly split. The survey is biased towards males and acquired losses of the arms. This type of loss might tend to favour the dominant side

When the countries are broken down, Oxford and Göteborg, and Stockholm and Edinburgh are more alike with biases towards the right in the former and the left in the latter.

Level of loss

In agreement with other statistics there are more individuals with losses below the elbow (B/E) compared with that above (A/E). However the Italian experience is more strongly biased towards the longer remaining arm. The differences between the centres is more marked with Stockholm and Oxford being most evenly distributed and Göteborg being most biased.

Age range

The survey was directed at adults (above 16), the result is that the population seems to reflect a broad range with a fall off towards the older age range. The median for the UK is in the 40s (and is similar to the 1992 survey), while the Swedish majority is in the 50s. Edinburgh is extremely evenly spread. Oxford has the peak in the 40's, Göteborg in the 50's and Stockholm has the youngest population.

Time since loss

For the second survey the persons with a recent loss were excluded in the earlier Oxford based survey this was not the case sothe vast majority of the population (80-90%) has many years of experience (greater than 5 years) except Oxford whose users are more evenly spread.

Type of prosthesis used

This shows some quite strong national choices. The Italian and British populations using mechanical hands of different sorts while they are rarely used in Sweden. A similar number use electric instead. All three areas use many cosmetic devices. There has been, for many years, an emphasis in Sweden to fit electric hands to children as young as possible. It is probable therefore, that many more Swedes have been exposed to the technology and so use it. The most important question (an one this survey cannot address) is the possibility that the earlier exposure encourages a greater long term usage.

When the individual centres are studied, Edinburgh is more like Stockholm than Göteborg with very large numbers of cosmetic hand users (73% and 62%). However when a comparison is made with the reported problems significantly fewer in both surveys marked cosmesis as a problem (8% and 3%). This implies they are happy with their provision.

PROSTHESIS USAGE

Time

It has been recognised that the level of usage of a prosthesis is very difficult to measure [4], this was the purpose of using two different questions concerning the length of time worn. When asked how long they wear their prostheses the respondents suggest that the majority wear them for well over 8 hours a day. But as they are also in majority those that say they use their limbs *regularly*, then this finding may be less than profound as it shows that once users commit to a device then they will tend to wear it.

This confirms that if individuals use their prosthesis they tend to use them all day. This is wore and not used. In a similar survey conducted at Cambridge in the early 90's 58% of the respondents indicated greater than 8 hours.

Directed use

The majority of the users in both countries use their devices for work (79% and 68%, little difference in the individual centres). The comments they added include that this is often for the benefit of co-workers or customers as much as for themselves. By contrast, the use for sport is far more likely to be because the device gives them real function of some sort within that activity.

Finally, the use of a prosthetic device for driving is again purely a functional concern (although ironically a survey the UK found that the respondents often *used* a cosmetic devices for this task). Fewer Swedish drivers reported that they didn't driveGenerally in both countries there are same percentages of people who do drive , and don't use their prosthesis for the task. According to the Swedish Central Bureau of Statistics, about 62% of the Swedish population hold drivers licenses. In 1996, 67% of the UK population held licenses but fewer prosthesis users say they don't drive.

Problems

This is the most mixed selection. Substantially more Italians complain of problems of maintenance. A similar percentage in Göteborg as Bologna had problems of cosmesis. All the concerns are universal and evenly spread and the only matter that has a of lower concerns is the cosmesis in certain centres.

Discussion

Although the survey has results for a self-selecting group of users, who in general identify themselves as regular users of prostheses, the results of such a survey are no less valid as these individuals are those most likely to use a design of an artificial arm system. It has been observed before [2], that the question of why others *don't* use prostheses is a similarly important question, but this method of enquiry is unable to address it directly.

Longevity of the device is clearly important, this has a direct impact on the design of any prosthesis that is externally powered.

It can be seen that the four major groups have similar characteristics. Only the use of cosmetic hands strongly separates the populations and their priorities, thus findings from the more detailed survey are likely to be applicable across a wider population.

One key finding, is that the users asked for improvements in the prosthesis. That is to say that users of cosmetic hands asked for better cosmesis and users of functional hands asked for improved function. This can be interpreted as there is a need for a prosthesis with improved both function and cosmesis)

SUMMARY OF THE SURVEY POPULATION

The population has more males, with an evenly spread age range. Over 60% have an amputation, which occurred more than 5 years ago, probably below elbow. Their prosthesis is cosmetic, but they are more concerned about the fit, function and maintenance of the limb, which they use for work and a little less for sport and driving, and for well over eight hours a day.

REFERENCES

[1] Kyberd P.J., Beard D.J. & Morrison D. J. "The population of users of upper limb prostheses in the Oxford region", Prosthetics and Orthotics. International, 21, 85-91 1997.

[2] Kyberd P.J., Beard D.J., Davey J.J. & Morrison D. J. "A Survey of upper limb prostheses users in Oxfordshire", Journal of Prosthetics and Orthotics, 10(4), 85-91, 1998.

[3] Burroughs SF. & Brook JA., "Patterns of Acceptance and Rejection of Upper Limb Prostheses", Orthotics and Prosthetics, 39(2), 40-47, 1985.

[4] Fraser C. "A Survey of Users of Upper Limb Prostheses", British Journal of Occupational Therapy, 56(5), 66-168, May 1993.

[5] Swedish Central Bureau of Statistics, "Statistics for 1997" Report.

APPENDIX - THE QUESTIONNAIRE

1) How frequently do you use an artificial limb? *Daily, Occasionally, Never*
2) What joints does your artificial limb replace? *Shoulder, Elbow, Wrist, Fingers*
3) Please indicate which type of limb do you use? *Cosmetic, Split hook, Cable Hand, Electric*
4) Which is the one area that you find most problems with*? Fit, Function, Cosmesis, Maintenance*
5) If you work, do you use your limb at work? *Yes/No - Occupation*
6) Do you use your limb for sport or leisure activities? *Yes/No - Activity*
7) On average, please estimate how many hours a day do you wear the arm?
8) Do you drive using the arm? *Yes/No/Don't drive*

Category	Group	UK	Sweden	Italy
Gender	Female	42	28	
	Male	57	67	
Side	L	47	49	
	R	51	49	
Level of loss	A/E	44	43	26
	B/E	56	58	74
Type of loss	C	35	21	14
	T	61	76	86
Type	none	1	0	
	Cosmetic	63	56	57
	Mechanical	23	3	29
	Electric	11	28	14
	Unknown	1	14	

Table 1 Basic details of the respondents

Category	Group	UK	Sweden	Italy
Use	Daily	88	93	
	Never	4	0	
	Occasionally	8	7	
Time	< 8hrs/day	24	17	26
	> 8hrs/day	75	83	74
Work	No	27	14	
	Yes	67	80	
	?	6	6	
Sport/leisure	na	3	7	
	No	44	28	
	Yes	53	66	
Drive with arm	Drive with	55	67	
	without	18	20	
	don't drive	25	12	

Table 2 Types of use for the survey population

Category	Group	UK	Sweden	Italy
Problems	Cosmesis	12	14	22
	Function	26	23	17
	Fit	24	22	25
	Maintenance	22	18	37
	none	7	0	

Table 3 Reported Problems

PROSTHETIC MANAGEMENT OF A CHILD WITH A PHOCOMELIC DEFICIENCY

Heidi Bauer-Hume, B.Sc., C.P.(c), Sol Wierzba, RTP(c), Anne Woo, B.Sc., O.T., John Bishop, E.Eng.T.

INTRODUCTION

The subject of this case study is an 8 year old girl who has a phocomelic deficiency of her left upper extremity. Her phocomelic hand consists of 3 digits including a thumb and has limited strength and range of motion. She also has an unusually prominent and mobile "spiked" clavicle and no scapula. All other limbs and systems are unaffected and she is a bright, energetic little girl.

HISTORY

Initial Prosthesis

Prosthetic fitting began for Katy at 6 months of age. She received a passive shoulder disarticulation prosthesis. This was done to aid in sitting balance and establish a wearing pattern for future, more complex prosthetic fitting. The socket was a hemi-thoracic frame-type laminated hard socket in which the "spiked " clavicle and phocomelic hand protruded through openings in the socket. The prosthetic components were: an Otto Bock 12S5 flexion/ abduction friction shoulder joint, a Steeper children's 1½" passive elbow and a Centri crawling hand and glove.

Katy wore the device for most of her waking hours.

First Powered Prosthesis

At 1¾ years of age, Katy was fitted with her first powered prosthesis. Only the hand was activated. A similar socket design to the passive device was used, with some modifications. Now, there were control issues to consider.

The new laminated frame-type socket required modifications in order to accommodate the additional weight of a powered fitting. Due to Katy's small size and "spiked" clavicle, the amount of suitable area available for vertical loading was very limited. To increase comfort, the upper portion of the hard socket was lined with a softer laminated silicone material. This served to better distribute weight over the narrow trapezius (weight-bearing) region just proximal to her clavicle.

Control of the electric hand had to be simple and direct, considering her young age. After much experimenting, it was decided to use the thumb on Katy's phocomelic hand to push a single-acting micro-switch. This would control the VASI 0-3 electric hand via a Steeper voluntary opening circuit. The micro-switch was positioned on the outside of the socket where Katy's thumb rested. A precisely crafted cradle of acrylic laminated material was fitted to the microswitch's lever. This cradle ensured support and feel for Katy's thumb to successfully depress the switch. The prosthesis was powered by a 6 Volt 180mAh Nicad battery pack (5 AAA cells) installed within the humeral section.

This system worked quite well and Katy became increasingly bimanual.

Second Powered, more Complex Prosthesis

Seven months later, at 2½ years of age, Katy's mother decided to add weights to the prosthesis in preparation for the addition of an electric elbow in the next fitting. Again, Katy accepted the change well. Katy's mom had been a strong advocate for powered prosthetic fitting from the beginning. She felt that the powered elbow would enhance Katy's function and thereby help her to incorporate the prosthesis into her self-image. Katy's mother has provided the continuous support, training and inspiration necessary for a successful outcome.

We were faced with a new challenge. The primary problem with the addition of an electric elbow was one of control. Katy's phocomelic hand was not strong or mobile enough to access another pair of inputs. Myoelectric control was out of the question as mounting the electrodes over the pectoralis and trapezius muscle groups would have resulted in too much inadvertant elbow activation due to Katy's constant movement within the socket. Force Sensing Resistors (FSR's) required more strength than Katy had. The only option possible was Capactive Touch Control, requiring only slight touch for activation. After much coaxing, it was determined that Katy could consistently and deliberating articulate her clavicular "spike".

Once the control strategy was decided upon, the hardware needed to be designed and fabricated. There was nothing on the market to link the CTC's to the VASI 3-8 electric elbow (the only elbow appropriately sized). A special CTC board needed to be designed. Initially, a miniaturized version of Bloorview MacMillan Centre's wheelchair touch-plate control was developed by our interfaces program. On the board, each channel had its own potentiometer and LED. The LED's were crucial in being able to monitor each channel individually to determine if the channel was being activated.

Two CTC buttons were mounted on a narrow, curved laminated strip that bridged over, but did not touch, Katy's "spiked" clavicle. By moving her "spike" posteriorly, she accessed one of the CTC buttons and was able to extend the elbow. Conversely, by moving her "spike" forward she could make contact with the other CTC button and flex the elbow. Bridging the laminated strip in one piece over her clavicle prevented clothing from disrupting contact with the buttons. The buttons were Otto Bock's round gold-plated electrode contacts.

Hand control continued to be the single-acting microswitch operating a voluntary opening VASI 0-3 electric hand. This prosthesis was powered by a 6 Volt 450mAh internal battery pack installed within the humeral section.

The incorporation of silicone into the socket of the previous prosthesis had been successful and was repeated for this one. The Otto Bock 12S5-shoulder joint was also used again. A cosmetically shaped laminated shoulder joint cover was installed, even though it limited the joint's range of motion slightly.

Functional Status

Katy wore the device for 8 hours a day except on very hot days. She used it for all age-appropriate activities. Her control of the elbow was spontaneous and subtle. There were no extraneous movements as would often be seen with any other mechanical switch option, i.e. pull or push switch etc. Her control of the electric hand was also quite smooth.

Frequency of Repairs

There were problems with breakdown of the Otto Bock shoulder joint and the switch-mounting lever where Katy's thumb rested. The breakdown of the switch-mounting lever was simply caused by wear and tear on the components and required several replacements over the lifetime of the prosthesis. In the case of the Otto Bock shoulder joint, it would lose the friction after several months and fall apart necessitating reassembly of the prosthesis and repair of any broken wires caused by the failure of the shoulder joint.

Third Powered Prosthesis featuring Additional Complexity and Improved Component Design

By the time Katy was 5 years old she had become an excellent user of her prosthesis. It was decided that she should advance to 2-site hand control and have a VASI 2-6 hand. Again, control was an issue. The logical choice was to replace the problematic push-switch with two CTC buttons, one for hand closing and the other for hand opening. Very precise positioning of the buttons on either side of her phocomelic hand allowed Katy to control the VASI 2-6 electric hand. She did this by shifting her hand back and forth approximately ¼ inch touching one button with her thumb and the other with her third digit. As before, a special board was designed and fabricated at the Bloorview MacMillan Centre, this time, it contained 4 channels. It was fabricated by our Electronics Department.

This CTC board also featured a series of LED's and potentiometers, each one, corresponding to a CTC button. The board was housed in the humeral section of the prosthesis and could be easily accessed by removing a specially laminated cover. The LED's were even more important now for troubleshooting since we had 4 channels to deal with. They were very helpful in determining the correct placement of the CTC buttons. For example, if a particular LED was always lit up, the CTC was obviously positioned too close to Katy and needed to be repositioned.

In order to address the shoulder joint problem, a different design was adopted. The Steeper children's 1¾" passive elbow was used as the shoulder joint. To mount this shoulder joint to the prosthesis, a reinforced laminated bulkhead was attached to the socket frame. The bulkhead and joint combination were carefully shaped to preserve symmetry. A custom shoulder flexion/extension stop pin was installed in order to prevent excessive flexing of the wiring bundle that passed through the shoulder joint to the humeral section. A corresponding curved groove was carved into the bulkhead allowing 90 degrees forward flexion and 10 degrees posteriorly. Shoulder abduction was limited to 90 degrees laterally. The 0-degree adduction stop had to be very strong in order to protect Katy's fingers from being pinched between the humeral section and the frame. This is a constant concern as Katy often "roughhouses it" with her 3 siblings and playmates. The abduction/adduction stops were achieved with precise shaping of the upper edges of the humeral lamination which stop medially against the bulkhead and laterally against a step on the Steeper joint surface.

Both of these changes in component design alleviated the previous repair problems. This prosthesis was virtually maintenance free.

The socket design and battery capacity were the same as the second powered prosthesis. The only other difference was the addition of the VASI OMNI wrist. It boosted the functional capability of the prosthesis by allowing 30 degrees of passive wrist motion in all directions. Katy very readily and automatically used the flexibility of the wrist to position the hand as needed.

Current Prosthesis

Katy's current prosthesis is identical in design and components to the previous one. The only change is the size of the hemi-thoracic frame and the humeral and forearm sections, she simply outgrew them. The hand and elbow were overhauled and reused. The same CTC board was also reused. The only new components were the shoulder joint and the OMNI wrist. They are more subject to wear and tear and don't lend themselves to refurbishing.

SUMMARY

Katy is an excellent user of her prosthesis, despite the high level of her deficiency. The latest control strategy takes advantage of four very distinct movements which translates into four separate functions. This simple, direct

system has enabled her to achieve a high degree of spontaneity and function in the use of her prosthesis.

Katy is proof that a high-level limb deficient child can effectively wear and use a complex externally powered prosthetic device. When the clinic team, parents, child and supporting technical expertise unite, a successful outcome can be realized.

ACKNOWLEDGMENTS

We would like to acknowledge the following people for their efforts and expertise in the preliminary work with Katy. Greg Bush, B.A., C.P.(c), Jerzy Antczak, B.Sc.,(HON), P.Eng., Sheila Hubbard, Dip., P.& O.T., B.Sc.

A PILOT STUDY EXAMINING REPETITIVE STRAIN INJURIES IN PEOPLE WITH LIMB LOSS

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INTRODUCTION

Over the years several clients of the Institute of Biomedical Engineering with upper extremity limb loss have complained of symptoms consistent with repetitive strain injury. At least two of these were diagnosed with carpal tunnel syndrome of their non-prosthetic limb. These observations lead to the development of a pilot study to look at a sample of the clients from the Institute. The goal was to determine if people with unilateral below-elbow amputations or congenital limb deficiencies tend to develop symptoms consistent with repetitive strain injuries on their non-prosthetic side.

This study was a special joint project between the Institute of Biomedical Engineering and the School of Occupational Therapy at Dalhousie University in Halifax. As final requirement for a Bachelor of Science in Occupational Therapy with Honours at Dalhousie University, students are required to successfully complete an independent study incorporating all the elements of a full research study. The research question with regard to repetitive strain injuries was submitted by the therapist at the Institute. Once the question was chosen the therapist became the project tutor, supervising and directing the pilot study resulting in an outcome relevant to current clinical practice. The student was the principal investigator.

THE RATIONALE

Based on the observations and complaints presented by clients at the Institute and in discussions with therapists from Workers' Compensation Board, a pilot study was proposed to see if further study was warranted. From experience it is known that clients rely heavily on the use of their non-prosthetic limb for performance of daily activities. Even with the advancements in prostheses no prosthesis can fully replace an actual limb. The person has lost dexterity and sensory feedback which will never be replaced [4,9,15]. Given this, it can be assumed that the person will overuse their intact limb. If the overuse persists over time, the client could be put at risk of developing a repetitive strain injury on their non-prosthetic side. Those clients who do develop repetitive strain injury pose a challenge for treating professionals. The traditional intervention for RSI such as resting the affected limb [1] is not applicable since the client cannot compensate with the prosthesis alone. Do people with unilateral below-elbow amputations have a tendency to develop repetitive strain injuries?

Observations of clients over time have revealed several types of choices are being made with regard to prosthetic wearing. One, the clients opts not to use or wear a prosthesis. They rely exclusively on their intact limb to perform their occupations, having learned compensatory one-handed techniques [4]. The residual limb is occasionally used by some as a stabilizer depending upon the length of the limb. The second choice is to use an active prosthesis, such as a myoelectric or conventional, usually for daily activities particularly where an active stabilizing or gripping is needed. The final choice is to wear a passive/cosmetic prosthesis and occasionally use it during light duty activities for passive stabilization. What influence does the type of prosthesis use or lack of it have on the evidence of repetitive strain injury?

LITERATURE REVIEW

Defining RSI

Repetitive strain injuries (RSIs) can also be commonly referred to as cumulative trauma disorders, overuse syndromes or occupational disorders [12]. Repetitive strain injuries are characterized by pain, weakness, paresthesia, limited movement and numbness [10]. Local tenderness, cramping/aching muscles can also occur. For the purpose of the study, repetitive strain injuries referred to a number of these symptoms recognized as resulting from engaging in activities involving sustained, repetitive motion, static postures, vibration, poor body mechanics and cold temperatures [1,10].

The underlying physiology of repetitive strain injuries is not clear. Fast [5] notes that RSIs are not true strain injuries, that is, injury to soft tissue resulting from excessive stretching. RSIs are thought to develop when repeated wear on body tissues exceeds the body's ability to repair itself [7]. In earlier stages of the disorder, neural compression, inflammation of muscle and tendons are involved [10, 14]. If the body tissues continue to be exposed to repetitive movements, local tissue ischemia results, which leads to metabolic and enzymatic changes [10, 14]. The affected muscles and tendons can become more susceptible to micro-tears and inflammation [14]. Initially, symptoms will be present only during performance of the culpable activities; however, as the condition worsens, the symptoms, particularly pain, persist during performance of activities and at rest [3, 14]. Cases have been reported in which a person subjectively feels pain, yet no evidence of pathology has been found [5]. There is some consensus that neural dysfunction of pain transmitters in the dorsal horn of the spinal cord is responsible for the pain experienced in RSIs [5]. Non-activity factors, such as age, gender, pregnancy, and chronic disease, have also been found to be correlated with the development of these conditions [13]. Regardless of the mechanism at work, RSIs seriously jeopardize an individual's ability to perform activities of daily living: self-care, productivity, and leisure [11,14].

Literature Relevant to People with Amputations

In the literature review performed, there were no studies of overuse syndrome in unilateral upper extremity amputees. Ironically, just after the completion of this study a paper appeared in Prosthetics and Orthotics International [8] on this same subject. The authors of that article stated: "It is surprising that no one has formally described overuse injuries in the remaining arm of upper limb amputees". Their study noted that 50% of their population studied complained of overuse problems or repetitive strain injuries.

METHODS AND PROCEDURES

A questionnaire was designed for the purpose of the study and mailed to 18 clients of the Institute of Biomedical Engineering.

Subjects

Criteria for inclusion in this study were: unilateral below elbow amputation or congenital limb deficiency, ages 12 or above, and English speaking. Exclusions included: clients with more than one limb affected by injury or deficiency; clients with partial hand losses; and those with learning problems. Eligible for inclusion were 9 adults with acquired below elbow amputations, 6 adults with congenital below elbow deficiencies and 11 adolescents with congenital below elbow deficiencies. 6 of the 9 adults with acquired losses and 6 of the 11 adolescents with congenital deficiencies were randomly selected to participate in the study. There were a total of 18 subjects.

Instrument

A letter of introduction and informed consent introduced the study to the clients prior to mailing of the questionnaire. After a deadline to return the questionnaire, clients who had not replied were contacted by the principal investigator and provided with the option of completing the questionnaire over the telephone.

The questionnaire (Repetitive Tasks Questionnaire) was specifically designed for the purpose of the study. The questionnaire was derived from various sources including clinical tools and instruments cited in the literature. It was divided into three subsections: List of Repetitive Activities, Symptom Survey, and Demographics. The questionnaire took about 30 minutes to complete. It is important to note that one cannot make a diagnosis of a repetitive strain injury with a mailed questionnaire [6]. Thus, the participants were asked to report on the symptoms they experienced during the performance of repetitive activities.

Clients were provided with a 72 items List of Repetitive Activities and asked if they performed the activity, wear their prosthesis for performing the task and if they have pain in their non-prosthetic arm when performing the task.

In the Symptom Survey section, they were asked to code diagrams of their hands and arms to indicate pain, tingling, numbness and decreased sensation in their non-prosthetic arm. They then were asked to answer some open ended questions with regard to the symptoms they portrayed. The final part of the symptom survey asked a series of twelve questions relating to pain, numbness, weakness, tenderness and tingling. The person was asked to respond to one of five options which were a form of rating scale. An example is: Do you have weakness in your hand or wrist: 1. No weakness, 2. Mild weakness, etc. to 5. Very Severe weakness.

The Demographic section asked specific questions related to the client's situation. The person was asked if they were diagnosed with a repetitive strain injury, if they have seen a doctor about arm symptoms, the dominance prior to surgical or traumatic amputation, type of prosthesis used and what they do for a living as well as an request for a final report of the study.

RESULTS

Descriptive statistics were used to analyze the data collected in this study due to its non-experimental nature [2].

Demographics

Response Rate

14 of 18 subjects completed the Repetitive Tasks Questionnaire which represents a 74% response rate. Nine subjects responded by mail, four completed the survey over the telephone and one completed the survey in a face to face interview.

Age and Gender

Four of the 14 were adults with traumatic amputations and a mean age of 38.0 years, 5 were adults with congenital deficiencies and a mean age of 24.8 years, and 5 were adolescents with congenital deficiencies whose mean age was 14.6 years.

Females accounted for 50% of the survey respondents.

Vocation

At the time of the survey, 57% were full time students, 29% worked full time and the remaining 14% did not work or go to school.

Prostheses

43% wore myoelectric prostheses; 36% wore passive prostheses only and 21% had a prosthesis but were considered "non-wearers.

List of Repetitive Activities and Symptom Survey

The subjects reported experiencing pain while performing an average of 7.14 activities. Both the adult groups, traumatic and congenital reported a mean of 11.0 activities where they experienced pain while the adolescent group reported only 5.6 activities on average.

Eight of the 14 subject indicated numbness, decreased sensation, pain and /or tingling on the hand diagram of the Symptom Survey. The specific grouping is as follows:

Table 1
Proportion of Subjects Indicating Symptoms on the Hand Diagram

Group	Proportion of Subjects
Adult-Traumatic	4/4
Adult-Congenital	1/5
Adolescent-Congenital	3/5

4 subjects indicated symptoms suggestive of carpal tunnel syndrome. One subject indicated patterns of symptoms that would be found in individuals with ulnar nerve entrapment and tennis elbow. Some subjects also indicated experiencing wrist and shoulder pain.

With respect to the multiple choice questions on these symptoms, the subjects reported an average of 3.29 symptoms, with the Adult Traumatic group indicating 7.75 symptoms on average. This was much higher than the mean responses of the congenital groups, with adolescents reporting 2.4 symptoms on average and the adults, 0.60 symptoms.

Dominance

Half of the subjects in the Adult-Traumatic group lost their dominant limb while the other half lost their non-dominant limb. The group who lost their dominant limb reported more activities (12.5) on average during which they experience pain as opposed to 9.5 for those who lost the non-dominant limb. There was little difference between the two groups with regard to the average number of symptoms indicated. The individuals switching dominance due to the limb loss reported and average of 8.0 symptoms and those maintaining dominance a mean of 7.5 symptoms.

Table 2
Mean Responses of Adult-Traumatic Subjects with Dominant vs. Non-dominant Losses

Group (n=2)	List of Repetitive Activities	Symptom Survey
Dominant Loss	12.5	8.0
Non-dominant Loss	9.5	7.5

Age and Gender

Of the four subjects in the Adult-Traumatic group, two subjects were over age 50 while the remaining two subjects were under age 50. Two-tailed, unpaired t-tests were performed to determine whether there were differences between responses of subjects over age 50 and under age 50; however, the test results were not significant. The older group did report more symptoms (10.0) than the younger group (5.5). They also reported fewer activities produced pain.

Similarly the responses for the all the subjects with relations to gender were tested since the literature indicated there is a correlation [13]. Of the seven males and seven females the t-tests results were not significant. Since the number of subjects is small it is worthwhile examining the trends in the data.

Table 3
Mean Responses of Male and Female Subjects

Group (n=7)	List of Repetitive Activities	Symptom Survey
Males	5.86	4.14
Females	8.43	2.43

Prosthesis Use

Some interesting findings appeared with regard to the type of prostheses used or worn and the types of activities and symptoms which were problematic.

Table 4
Mean Responses of Subjects According to Type of Prosthesis Used or Worn

Type of Prosthesis Worn	Problematic Repetitive Activities	Symptoms Indicated
None	6.0	3.67
Passive prosthesis only	5.6	1.4
Myoelectric prosthesis	9.0	4.67

This might lead to several questions about the prosthesis users:
Are they more active?
Do people without prostheses or with only passive prostheses avoid performing repetitive activities?
Does having a prosthesis encourage more activity?
Does prepositioning of the wrist add to strain?
Does the prosthesis force them to use the dominant hand more?
Does this group use a prosthesis to decrease symptoms of preexisting RSIs?

Type of Activities and Vocation

As to be expected, activities that could be considered heavy work, requiring exertion of high force or static in nature provided the most responses to the Repetitive Activity List. There were some tasks such as raking, shovelling snow, and pushing a lawnmower that were not included which subjects suggested should be included on the list.

The responses of the subjects according to their primary vocational roles were also examined as follows:

Table 5
Mean Responses of Subjects Grouped According to Primary Vocational Role

Group	List of Repetitive Activities	Symptom Survey
Full-time Workers (n=4)	4.75	3.0
Full-time Students (n=8)	7.13	2.88
Non-Worker/Non-Student (n=2)	12.0	5.5

It was not determined why the individuals were non-worker/non-students. We might also question if the person was a non-worker/student because of having repetitive strain injury.

DISCUSSION/CONCLUSION

This study found that approximately 57% of the people studied with unilateral below-elbow amputations or congenital limb deficiencies do develop symptoms consistent with repetitive strain injuries on their non-prosthetic side. This is consistent with the findings of Jones and Davidson [8]. It was also expected that the group studied would follow the same patterns of evidence of RSIs as shown in the literature according to age, gender, and vocation. Our results were not conclusive.

Wearing a myoelectric prosthesis was thought to provide extra function for the individual thus assisting in preventing evidence of RSIs. In this study, however, it does appear that this group of myoelectric users indicate more symptoms when performing repetitive activities. It is not evident if these individuals lead a particularly active lifestyle or perhaps wear a prosthesis to decrease an already evident problem or if indeed the prosthesis use

increases the amount of repetitive strain on the non-prosthetic side. It is hoped that future studies will shed more light on this problem.

It was evident that individuals who loose their dominant limb seemed to be more prone to symptoms consistent with repetitive strain injuries than those losing a non-dominant limb. Care should be taken when treating these individuals with dominant limb loss. Treatment approaches such as using repetitive activities to improve dexterity for dominant use should be undertaken cautiously. Clients should be made aware of the potential for developing a repetitive strain injury and strategies should be discussed with regard to joint protection, and principles of work simplification. Provision of a prosthesis may provide some assistance in static/holding activities.

This study was a pilot study to investigate if this subject warrants further investigation. It is recognized that the numbers were small, the questionnaire requires revision and further testing, and that a matched sample may be of value. Nevertheless, further study is required before conclusions can be specifically drawn.

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FINAL NOTE

The contents of this paper were take from the Final Report on this study written by Genevieve L. Neufeld while at Dalhousie University. Joan Versnel, M.Sc. O.T. (RegNS), the Independent Study Course Coordinator at Dalhousie University provided valuable advise for this project. After completing this study, Genny received the Fred Sammons Award for Undergraduate Research as a result of receiving the highest grade for the project.

FITTING THE HUMERAL LEVEL BRACHIAL PLEXUS AMPUTEE WITH EXTERNALLY POWERED MYOELECTRIC CONTROL

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INTRODUCTION

Brachial plexus injuries present unique challenges to both the patient and prosthetist. The brachial plexus injury can be classified in many categories: complete or incomplete lesion, with either an intact or amputated limb as a result of the injury. The intact yet flail limb presents with lack of sensation, shoulder subluxation, severe pain, and is often subjected to unintentional burns and cuts. Many brachial plexus patients with an intact limb elect to undergo limb amputation to reduce complications or improve function with prosthetic fitting options. Unfortunately, some patients continue to experience similar post-amputation complications. Fusion of the shoulder joint is a surgical option in an attempt to decrease pain and reduce shoulder subluxation. This procedure may be performed before, after, or in association with amputation of the extremity.

BACKGROUND

Traditional prosthetic management for this type of patient includes both body powered and externally powered designs. Patients experience many limitations when fit with body powered designs due to inadequate range of motion in the shoulder region. Typical myoelectric fittings often failed due to the extreme difficulty in fitting this level. Patients find the weight intolerable and a poor socket fit leads to inconsistent readings at the myoelectric sites.

DISCUSSION

Myoelectric control offers many advantages for the high level amputee. Brachial plexus injury patients benefit from an aggressive prosthetic approach. In the last year, six patients have been fit as shoulder-level amputees using myoelectric control. Critical features of this approach include socket design and stabilization, as well as the location selection of myoelectric sites (i.e. trapezius, pectorals and rhomboids).

Inherent with this patient population, myoelectric site selection is often challenging. Site selection should be pursued with the knowledge that future adjustments or repositioning of the sites may be necessary and myoelectric training is required. One success factor is the design of a test socket with myoelectrodes installed and connected to an EMG analyzer. This allows the patient to receive ongoing and intensive therapy from an experienced occupational therapist prior to fabrication of the definitive prosthesis. Consistency of electrode placement with the donning and doffing of the prosthetic socket is also critical to successful myoelectric fittings. Once the patient has successfully isolated effective muscle groups, re-evaluation of site fittings is suggested and continued fitting toward a definitive prosthesis.

Rationale for fitting this patient group as a shoulder-level fitting as opposed to a humeral-level fitting includes: improved socket stabilization, myoelectric sites placed within prosthetic socket, increased range of motion with

prosthetic shoulder joint, reduction of force on the anatomical shoulder joint, easier donning/doffing of the prosthesis and clothes. As a result, improved functional independence can be achieved.

Obtaining an accurate cast begins by wrapping the residual limb and torso in SaranÔ wrap. Anatomical landmarks and initial trimlines are identified. The torso is wrapped with elastic plaster to create a base, over which rigid splints are then applied. Elastic ace wraps are applied over entire cast for compression before the plaster splints have set. Positioning of the residual limb and hand molding techniques establish the anterior-posterior dimension and proximal aspect of socket mold. After the mold is poured and rectified, 3/8" DurrPlex is pulled over the model to create the initial test socket for electrode placement and training.

The socket is a critical component of any prosthetic design. A modified shoulder-style frame socket is used to maximize stabilization of the prosthesis on the residual limb and torso. Loading of the anatomical shoulder girder is obtained through the pyramid-shaped socket, with anterior and posterior compression. (Fig. 1) That feature, in combination with encompassing the trapezius area, eliminates any distal migration of the socket. Rotation is controlled through the posterior-proximal and anterior-distal aspect of the socket. The definitive prosthesis consists of a flexible inner-socket and a rigid laminated frame. The flexible inner socket improves patient comfort by allowing increased flexible, improved socket adjustability, and specialized design features such as "floating" electrodes. These combined features allow the patient to maintain electrode contact while retaining a greater range of motion.

There are various options in addressing the residual limb. These include enclosing the limb within the socket with or without a prosthetic shoulder joint or leaving the limb open and free. Harnessing considerations for this design must allow for ease of donning and provide adequate stabilization. Elastic harnesses attached to the flexible inner socket provide a dynamic assist to suspension and positioning of the socket.

Components

Shoulder joints commonly available for this design are friction flexion-abduction or locking flexion-extension with friction abduction. Alignment of the joint is slightly internally rotated and placed just distal to the anatomical shoulder joint. The joint is also placed as close to the residual limb as tolerated. It should also be noted that in most brachial plexus injuries, the shoulder is atrophied to a degree to allow for such placement of the joint while still allowing the prosthetist to provide a cosmetically acceptable definitive prosthesis.

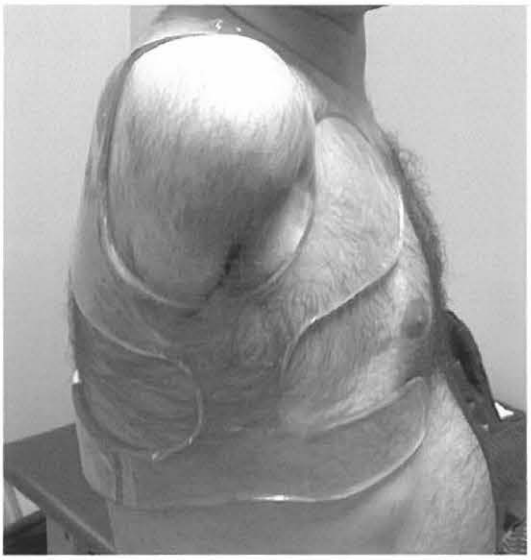
There are a variety of prosthetic options and combinations of myoelectrically controlled externally powered elbows, wrists, terminal devices and controllers that will effectively address the needs of the patient. Specific component selection must be determined on an individual basis.

Diagnostic Phase

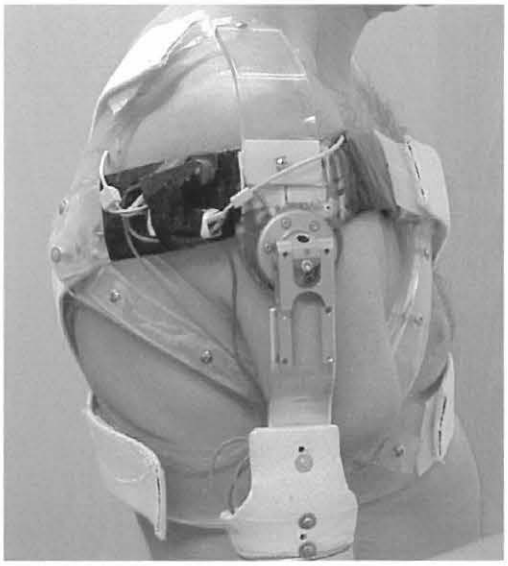
An initial rigid DurrPlex test socket is fitted and trimmed to simulate final socket design. A functional diagnostic test socket is then fabricated consisting of a flexible inner socket and rigid outer DurrPlex frame. (Fig. 2) At this point, shoulder joint placement is established and all components and electrodes are mounted. This diagnostic phase allows for continual adjustments to the socket, change of electrode site placement, movement or realignment of the shoulder joint, and determines the efficacy of the design and component selection to meet the needs of the patient. The patient is now ready for occupational therapy and prosthetic training. After undergoing intensive training procedures and all adjustments have been finalized, the prosthesis is ready to finish.

CONCLUSION

Six patients were fitted with similar designs outlined in this presentation. Four of the six patients had previously been unsuccessfully fitted with traditional designs. Two patients were new users. Functional skills and abilities were improved in all cases with 100% acceptance. Continued research and development will lead to functional improvements in components and design for unique patient groups such as brachial plexus injuries.



(Fig. 1) Forces applied from socket



(Fig. 2) Diagnostic prosthesis

USING COMPUTERS TO MAKE OUTCOME MEASUREMENT EASIER: INTRODUCTION TO THE SOFTWARE VERSION OF THE PROSTHETIC UPPER EXTREMITY FUNCTIONAL STATUS INDEX (THE PUFI)

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The work presented in this paper is part of a larger project designed to evaluate the long-term benefits, both functional and psychosocial, of upper limb prosthetics for children. Given the lack of a comprehensive functional status measure designed specifically for children who wear upper extremity prostheses, there is a corresponding absence of knowledge of the functional benefits that a child derives from using one of these devices.

At the time we started this project the only available evaluation tool for this population was the University of New Brunswick's Test of Upper Extremity Function. This observational test was designed for use by therapists to assess a child's prosthetic skill and progress during functional use training. However, other relevant questions arise relating to the extent to which the child actually uses the prosthetic limb for daily activities, the extent of independence achieved with its use, pattern of use from early childhood years through adolescence, and patterns of use associated with different types of prostheses. To address these issues, a new parent/child, self-report questionnaire, the Prosthetic Upper Extremity Functional Index or PUFI was developed by our clinical research group.

Intended to be used as an outcome measure (evaluate client's abilities over extended follow-up period), development of the PUFI started about 5 years ago and has occurred in a series of stages. This multi-stage developmental process has been typical for other functional measures such as the PEDI, [1] and the GMFM. [2] Initial steps included a literature review, generation of a list of two-handed activities common to childhood, a content validity check to ensure "2-handedness" of the tasks, an item reduction process to select the most relevant tasks, development of response options and clinician consensus on content and format.

The instrument begins with Part One: a short questionnaire asking parents to rate overall usefulness of the prosthesis for their child (very, somewhat or not at all) for cosmetic benefit and then 8 broad functional categories. These include personal care, dressing, relaxation at home, school, work, social events, sports and play. This section gives us a global view of parent's/child's impressions of the prosthesis and serves as an introductory section to familiarize the respondents with the concept of function.

Part Two of the PUFI was designed to: evaluate the ways that a child performs 2-handed activities; evaluate the success and value of the prosthetic device use as compared to functional ability without a prosthesis; identify performance difficulties associated with device use; and evaluate change in a child's abilities over time. The PUFI contains 5 response columns for each functional activity. Each column provides different information about the use of the prosthesis (ability to do the task, usual method of performance, ease of prosthetic use, usefulness of the prosthesis, and ease of performance without the prosthesis). The total scores for each column are calculated separately to give a picture of the child's overall status in each column for all tasks.

We designed 2 versions of Part Two to enhance its developmental appropriateness. The younger child version (ages 3-5) has 26 items while the school-aged version (ages 6 - 18) has 38 items. Both versions were designed to be parent report questionnaires. However, the older child version was modified so those children above the age of 8 could respond to the questions themselves, in effect creating a 3rd questionnaire format. The response options structure and scoring of the two versions are the same, allowing continued use of the PUFI with a child through his/her developing years.

The second stage consisted of a reliability study. The PUFI's test-retest, inter-rater reliability and aspects of construct validity were evaluated with a sample of children at Bloorview MacMillan. The 25 subjects included 10 young children (3 1/2-5) and 15 older children (6-16) fitted with myoelectric prostheses.

Test-retest results for columns B (usual method of performance), D (usefulness of the prosthesis for the specific task) and E (ease of performance without the prosthesis) were above 0.7. The children appeared to be more consistent in their responses than their parents. Column C, (ease of prosthetic use), proved to be the least reliable. This may have been due to a problem in interpretation of the distinction between the response options.

Inter-rater reliability of parents and children was also examined. Intra-class correlation coefficients were high for Columns B (method of performance) and E (ease of performance without the prosthesis) but low for columns C (ease of performance with the prosthesis) and D (usefulness responses). Again, we felt it was necessary to examine the problem of differentiation of response options for Column C. In the case of column D (usefulness), parent/child agreement was low because the children generally reported their prostheses as being more useful than their parents did. This may be the result of an environmental effect as the children were school-aged and therefore, away from parental observation for a large portion of the day.

Subtle refinements were then done to enhance clarification of wording in the instrument and measures taken to ensure that directions were clear to clients/families. Obviously the pilot study numbers were too small for definitive conclusions but we felt confident that the PUFI would prove to be a reliable measure of a child's ability to perform upper extremity activities with and without a prosthesis.

Stage three, a multi-centre trial designed to investigate the construct validity of the PUFI, (comparison of PUFI scores with UNB and actual observation of PUFI tasks) commenced on May 1, 1998 and will conclude on August 31, 1999. In addition to the PUFI, the families are also being asked to complete two other scales which measure family dynamics: the Family Assessment measure, FAM, [3] to measure areas of family strengths and weaknesses and the Impact on Family Scale [4] to study burden/stress factors.

Four paediatric amputee teams (not including Bloorview MacMillan) in Canada and the United States are participating. Data for 26 children has been entered to date and it is expected that 40 children will be enrolled in total. Involvement of these four centres will provide opportunity to test PUFI with a variety of different types of prostheses. The results will be combined with Bloorview MacMillan validity data.

At the end, we will take a final look at the items to make sure that each one is telling us something of value. Items consistently considered non-applicable or gender-biased will be eliminated. Ultimately we will also want to evaluate PUFI's ability to measure change over time.

In its present form, the PUFI (like other parent and client self-report measures) requires that the child and parent complete multiple, word-only paper forms which may act as deterrents to full completion by respondents. Scoring of paper forms is also time-consuming and cumbersome for clinicians because of the use of reverse patterns in scoring for some questions, separate scoring domains within the questionnaire, multiple choice options and many items, i.e. 38 items on the older - child PUFI but up to 200 items on some other questionnaires. Scoring difficulties encountered by clinicians may inhibit timely sharing of results with families and limit use of the measures. There are also potential problems with: the accuracy of scoring measures that have several domains and multiple response options, the conversion of raw scores to scaled scores and normative scores, and the ability to generate meaningful numeric, descriptive or graphic summaries of the results for the client/family.

It was therefore decided to proceed with a project to design and evaluate point-of-service software to provide a user-friendly, efficient and accurate method for administering and scoring the PUFI. This software will enable the children/parents to use a computer to complete the PUFI questionnaires independently. Using appealing graphics and icons, the software will have a clearly guided completion format with periodic feedback to client, sequential presentation of items, and a mechanism to ensure that questions are not skipped (reduction of missing data). Built-in scoring will allow immediate summarization of scores once the questionnaire is completed so that

results can be shared promptly with the child/parents. Items achieving a criterion score will be displayed to identify strengths, problems, and facilitate goal setting. Results will be saved and called up after subsequent completions to allow ready comparisons of follow-up results. A demonstration version of the PUFI software will be included in this presentation.

We are currently seeking collaboration for field and reliability testing of the software version of the PUFI. Development of the protocol for this field test is being done in accordance with standard practices for pre-testing questionnaires [5]. Field trials will involve having some children/parents at each facility complete the PUFI using the software, and then rate their impressions on the experience using a clinical utility questionnaire. Clinicians will also be asked to rate their experience with the PUFI software focusing on training, administration, feasibility, and scoring issues.

The investigative team and computer programmer will make revisions to the software based on the results of the field-testing. Once completed, test-retest reliability evaluation, in which the child does two separate completions of the computerized PUFI, 2-3 weeks apart, will be undertaken at Bloorview MacMillan and the external facilities. We are hoping to enroll up to 6 children per facility in order to obtain a sample size of 30 children. The reliability phase of the project is expected to commence in November of 1999. The reliability results will be compared with the test-retest results from our previous reliability study with the original PUFI paper forms to ensure that the computerized PUFI approach is at least as reliable as the paper forms. Assuming the reliability results are acceptable, final revisions will be made to the software and manual, and they will be made available to the participating Centres for clinical and research use.

The combined use of the UNB Test (to assess the child's ability to use the prosthesis in a controlled situation) and the PUFI (to evaluate actual performance within a natural environment) is recommended as the way to gain a comprehensive picture of the child's functional capabilities and actual use of a prosthesis. The use of one measure i.e. the PUFI, should increase the interpretability of the other, i.e. the U.N.B. Test.

The cost-benefit of prosthetic fitting for children has been a subject of debate for many years now. The issue is a complex one involving both functional and psychosocial factors and cannot be resolved through the use of a single evaluation tool. Functional benefits of prosthetic fitting do need to be evaluated within the context of a total outcome measure core set. Additional information such as developmental level, family support, peer acceptance, athletic and extra-curricular needs as well as levels of psychosocial adaptation to disability and emotional functioning need to be considered as well. The psychosocial impact of wearing a prosthesis will be explored further in research work planned for this coming year.

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ASSESSING THE SATISFACTION OF YOUNG USERS OF MYOELECTRIC PROSTHESES

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INTRODUCTION

The *Institut de réadaptation en déficience physique de Québec* (*IRDPQ*) began a delivery program in 1990 to provide upper limb myoelectric prostheses for young children. This centre is located in Quebec City (Canada) and provides rehabilitation services for the population of the eastern part of the province of Quebec. This territory is home to about 2.25 million people and represents an area of 525,000 km^2 (in comparison, Sweden has more than 8.5 million people living within an area of 450,000 km^2). Since the beginning of the program, 18 children have been fitted with a myoelectric prosthesis and trained in how to use it.

The occupational therapists and prosthesists involved in this delivery program were first guided by the literature that presents experiences in other rehabilitation centers (Toronto, Fredericton, Sweden, etc.). At the time the first fitting was carried out, in 1990, many studies supported the hypothesis that patients who had been fitted at a very young age, as early as 2 years old, were more likely to accept their prosthesis and to become full-time users. The occupational therapist used, whenever necessary, the *UNB Test for Prosthetics Function* [1] in order to evaluate the progress of the children and to adapt their training to the needs and personality of each patient (e.g. age, experience with upper limb prosthesis, distance between their residence and the rehabilitation centre).

Nine years later, the dropout rate and the level of satisfaction or non-satisfaction of the young users of myoelectric prostheses who received them at the *IRDPQ* remain unknown. The purpose of this study is to determine the dropout rate of the group of young users who have been fitted and trained with an upper limb myoelectric prosthesis at the *IRDPQ* since 1990. Other objectives of the study are to assess their importance and their satisfaction levels with respect to their prosthesis and to gather information about factors that influence the use or the non-use of the myoelectric prosthesis.

LITERATURE REVIEW

In the 1980s and earlier, many studies indicated that early fittings of myoelectric prostheses were possible and could be beneficial for children. This was supposed to lead to a better acceptance of the prosthesis and a higher likelihood of the child's becoming a full-time user. It is generally accepted that a myoelectric prothesis provides an acceptable combination of aesthetic appeal and functionality for most children. Acording to various studies, dropout rates were estimated to be between 13% and 37% [2, 3, 4, 5, 6, 7, 8]. Similar results were obtained more recently by Day [9] who found a dropout rate of 29% within the first two years of fitting, whereas Datta and Ibbotson [10] estimate this rate at 27% for their entire group. Trost [11] has added that the myoelectric prosthesis is a valuable, but not a complete and functional substitute for the conventional prosthesis. It is a well-accepted fact that fitting/training is more successful in cases of below-elbow amputation [3, 5, 11]. In this study, we consider the fitting/training process as all activities carried out by occupational therapists, prosthesists and other professionals

which enable a child to use a myoelectric prosthesis, including pre-fitting information and follow-up. In the case of below-elbow amputation, Scotland and Galway [3] reported a dropout rate of 13% for children who have a congenital upper limb deficiency. In the Glynn *et al.* study [5], this rate drops as low as 10% (congenital or traumatic deficiency). All the studies reported here involved between 11 and 124 children.

The use of the myoelectric prosthesis was also studied with respect to time of use per day and use patterns. Sörbye [2] stated that 65% of his entire group were full-time users, that is to say they use their prostheses all day and for all activities. Mendez [4] pointed out that 60% of the children who received myoelectric prostheses in the United Kingdom wore them continuously and derived benefit from them. In this study, 20% of the entire group were identified as intermittent users. In a 1989 study conducted at the Hugh MacMillan Rehabilitation Center (Toronto, Canada) [7], almost 69% of children were identified as daily users (18% full-time, 8% all day except for sports, 22.5% for school plus social use, and 20% partial-days). More recently, Heard *et al.* [12] demonstrated that a full-time wearer is not necessarily a full-time user. It is possible to wear a prosthesis without using it as a prehensile tool. The authors state that: "*A child will wear a prosthesis when they find it useful for them, and simply remove it when they can accomplish an activity more quickly and efficiently.*" Mechanical criteria such as grip strength, type of grasp, durability, reliability, control and others elements are identified as needing to be improved [7, 8], but other factors for fitting/training success are also presented. Training, follow-up, appearance, comfort, and parental attitudes are considered to be predictors of success [2, 7, 8].

Finally, Hubbard and colleagues [8] presented a parent-satisfaction evaluation of the myoelectric prosthesis for user and non-user children. They reported that 41% of those surveyed found the prosthesis very useful for day-to-day activities, 54% found it somewhat useful and 5% found it not useful at all.

It is difficult to compare these various studies. Their main limitations reside in the number of subjects, the issue of training and the definitions of full-time user, occasional user and other definitions, all of which vary from study to study. The evaluation techniques are not standardized and the periods of follow-up are not the same. Nevertheless, by referring to the literature, one is able to characterize the efficiency of the delivery program at the *IRDPQ* as concerns achieving the objectives set for our study.

PARTICIPANTS AND METHODS

The *Quebec User Evaluation of Satisfaction with Assistive Technology (QUEST)* was used [13] in order to assess the degree to which the children were satisfied with their prostheses. The first part of the *QUEST* provides information about the participant, as well as his or her prosthesis and environment. The second section of the *QUEST* reveals the importance that the participant attributes to each of the variables evaluated and rates their degree of satisfaction with each of the variables considered. It is also possible to gather sources of information concerning non-satisfaction. For this study, 34 variables were considered. Assessments were conducted at the child's residence or at the *IRDPQ* during an interview carried out by an occupational therapist who is familiar with upper limb myoelectric prosthesis fitting/training. Between 60 and 90 minutes were necessary to complete the interview. The children's files were consulted when information was missing. When children were judged too young to answer, parents were asked to respond for them. Parents were advised to priorize the importance their children give to the respective variables as well as to express the degree of satisfaction the children feel regarding each one. In this paper, the term *child/parent* is used to identify the participant.

All children who have been fitted and trained since 1990 at the *IRDPQ* with an upper limb myoelectric prosthesis were asked to participate in the study. Eighteen children in all were targeted: 5 boys and 13 girls. First, a letter was sent by the occupational therapist or the archivist to the parents, in order to explain the study and to obtain their consent regarding participation. When no answer was forthcoming, contact was made by telephone. Ten children/parents accepted to participate : 3 boys and 7 girls. Five children/parents refused to participate in the

study (lack of interest, not enough time, etc.) and 3 could not be reached.

RESULTS

Details concerning the ten children who acepted to participate in the study are presented in Table 1. All of them are still visiting the *IRDPQ* on a regular basis. Seven of the participants live with motor activity disabilities. Life habits identified as problematic without the prosthesis were: dressing/undressing (8), recreation (7), transportation (6), nutrition (5) and education (4). For example : *to tie shoelaces, to do up buttons, to use a skipping rope, to ride a bicycle, to cut with a knife, to hold geometry instruments*. Table 2 provides the number of participants who found the prosthesis useful when carrying out various problematic life habits.

Details of participants (n=10)
• 7 girls and 3 boys
• Age : 2½ to 16 years old (mean : 6½ years)
• Time since the first fitting : average of 4 years
• All congenital deficiencies (5 right and 5 left)
• 3 at/below-wrist amputations, 6 below-elbow amputations and 1 above-elbow amputation
• 9 VASI and 1 Steeper (adult size)
• 10 having previously received a cosmetic prosthesis
• 2 having previously received a hook

Table 1: Profile of participants at the interview.

Life habits carried on with a myoelectric prosthesis	*Number of participants*
Recreation/games	7
Education	7
Transportation	5
Dressing/undressing	5
Interpersonal relations	3
Nutrition	2

Table 2: Life habits carried on by participants with a myoelectric prosthesis.

Based on data obtained from the child/parent, Figure 1 illustrates the time of use per day of the myoelectric prosthesis. The answers reveal that children are mostly part-time users. We have determined that 20% of the children are full-time users, that is to say they use their prosthesis more than 6 hours per day and for a minimum of 5 days per week. In this way, we set at 80% the ratio of part-time users. The data also indicate that 70% of the children still use their prostheses, but that 80% of the entire group stopped using them for some period of time. The point at which the children stopped using their prostheses ranges from 3 months to 3 years after the first fitting.

In the second column of Table 3, for each variable listed in the first column, we find the number of participants who indicated that this variable is *important* or *very important*. We find the level of satisfaction for the same variables in the third column. Finally, the children/parents were asked to indicate their overall satisfaction with the myoelectric prosthesis. The results indicated that 6 children/parents are *very satisfied* or *satisfied*, 3 children/ parents are *somewhat satisfied* or *rather unsatisfied* and 1 participant is *not satisfied at all*.

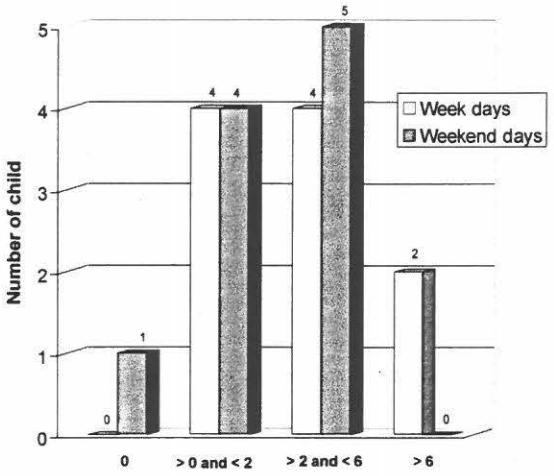

Figure 1: Number of hours of use per day.

Variables	Answered as *very important* or *important*	Answered as *somewhat satisfied, rather unsatisfied* or *not satisfied at all*
Weight	10	7
Comfort	10	5
Freedom (from harnessing)	10	5
Installation	10	4
Effectiveness	10	4
Technical follow-up	10	3
Appearance	10	3
Flexibility	10	1
Simplicity of use	10	0
Training	10	0
Heat	9	8
Repairs/servicing	9	4
Tip grip	9	4
Service delivery	9	4
Safety	9	4
Technical assistance	9	3
Cylindrical grasp	9	3

Variables	Answered as *very important* or *important*	Answered as *somewhat satisfied, rather unsatisfied* or *not satisfied at all*
Maintenance (by yourself)	9	2
Durability (battery)	9	2
Prehension force	9	1
Dimensions	9	0
Personal acceptance	9	0
Effort	9	0
Adjustments	8	5
Durability (glove)	8	3
Rehabilitation assistance	8	2
Durability (mechanism)	8	2
Rehabilitation follow-up	8	2
Rotation of the hand	7	3
Closing/Opening time	7	1
Support from others	7	0
Reaction of others	7	0
Tactile sensation	6	4
Noise	5	1

Table 3: Results concerning importance and satisfaction.

Finally, the non-participant group was described with the help of the occupational therapists' knowledge and the children's files. The non-participant group is composed of 8 children (6 girls and 2 boys) between 6 and 18 years old. All of them have congenital deficiencies. Four participants present a below-elbow amputation (1 of them has a multiple malformation), 2 have a wrist disarticulation and 2 have an elbow disarticulation. Seven of the group have been diagnosed with motor activity disabilities. The average time having elapsed since their first fitting was 6 years and 2 months at the time of the study. Only 7 files can be consulted (out of a possibiliy of 8), so we do not know whether one of them still wears his prosthesis or not. We do know that about 6 months to 2 years before the study, only 1 child was still wearing his prosthesis and using his prostesis whereas 6 had stopped using it. We suppose that the same situation pertained at the time of the study.

DISCUSSION AND CONCLUSION

Results indicated that participants are generally satisfied with their prostheses, even though 80% of them stopped wearing and using them at one particular point in time. Sixty percent (60%) of participants are *very satisfied* or *satisfied* with their prostheses. The dropout rate for the participant group is 30%. This rate increases to 53% if we include the non-participant group (9 children stopped using their prostheses out of a total of 17 children). Others studies do not indicate the profiles of non-participants. We can suppose that one reason non-participants refuse to participate is that they do not in fact use their prostheses. Most users belonging to the participant group are identified as part-time users (80%). We observe that all below-wrist amputees as well as those with wrist disarticulation and elbow disarticulation eventually stopped using their prostheses and were not using them at the time of the study (participant and non-participant groups). The only above-elbow amputee participant still uses his prosthesis, but had previously stopped using it for a long period of time. This confirms results obtained by others studies which indicate that children having undergone below-elbow amputation are good prospective subjects for being fit with the myoelectric prosthesis and trained in its use. As concerns our results, we think that it is important to inform children and parents that the success rate for fitting/training is lower for other types of deficiencies. Our results, in terms of the dropout rate, time of use per day and satisfaction, correspond to those obtained in other aforementioned studies.

The variables identified as being most important for children/parents when it comes to wearing and using their prostheses are: weight, comfort, freedom, installation, effectiveness, technical support, appearance, flexibility, simplicity of use and training. The sources of non-satisfaction come from heat, weight, discomfort, lack of freedom and adjustments. Many comments indicated that even after receiving information about the advantages and disanvantages of the myoelectric prosthesis, parents still want their children to have the most recent type of prosthesis. Only after using their protheses do children and parents admit that there are disadvantages and that there is a risk of the apparatus being discarded. The main features associated with the prosthesis which lead to negative comments include its weight, heat, delivery delays, fragility of the mechanism, difficulties with rotating the wrist, problems with the battery and cord, dimensions and loss of tactile sensation. We think that environmental factors such as training, technical support, maintenance service and others should be improved by the staff involved in providing fitting and training services for young users of myoelectric prostheses at the *IRDPQ*. We have noted that these factors are important in the success of the fitting/training process and that they are more difficult to improve for children/ parents who live in outlying areas. Moreover, like environmental factors, technical factors could also be improved, as it is supported by other studies. The authors think that it is clear that some technical aspects could be improved by research and development teams.

The main weakness of this study is a result of the number of participants involved. It would be interesting to assess the degree of satisfaction of all users of the myoelectric prosthesis in the province of Quebec and to try to evaluate the impact of enviromental factors such as training, distance from the rehabilitation center, technical support,

services, etc. Also, interviews were conducted by the two occupational therapists who participate in the fitting/ training of myoelectric prostheses at the *IRDPQ*. It probably would have been preferable if interviews had been carried out by another therapist. However, we adopted our strategy because the two aforementioned therapists are the only ones involved in providing myoelectric prosthesis fitting/training for young users at the *IRDPQ*. Being involved in the interviewing process was very useful for them in their clinical work. One of the positive contributions of this study concerns the knowledge of occupational therapists who provided information about non-participants. Through their contribution, we were able to establish a more accurate estimate of the real dropout rate. Another positive aspect is that the interviews were conducted by individuals not belonging to any technical research and development team. We think that this condition better reflects the reality experienced by many clinicians.

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QUANTIFYING IMPAIRED HAND FUNCTION IN THE CLINICAL ENVIRONMENT

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ABSTRACT

The Southampton Hand Assessment Procedure (SHAP) has been developed specifically to enable the broad-based evaluation of hand function irrespective of the disability, thereby allowing assessment of both natural and prosthetic hands [1,2]. This technique enables a contextual result to be formed (relative to normal hand function), hence providing a quantifiable index of functionality rather than the more conventional subjective measures. The establishment of normative data trials and subsequent statistical analysis demonstrates the procedure to be both reliable and repeatable.

The procedure has been undergoing evaluation at hand rehabilitation and prosthetic fitting centres. The subject group consists of those with impaired natural hand function (ranging from traumatic injury to diseased joints), as well as unilateral amputees and those with congenital deficiencies of the upper limb. These initial cases have assisted in the refinement of the index of functionality that results from the procedure.

The perceived hand function of these case studies is presented in comparison with the SHAP results. Quantification of functionality is of clinical importance to allow surgeons and therapists to monitor rehabilitation, and preliminary results suggest that the Southampton Hand Assessment Procedure provides a critical contribution to this process.

INTRODUCTION

Conventional methods of assessing hand function do not address the need for a standardised and objective procedure [1]. Yet, there is a key requirement to quantify the hand's functional capability to allow surgeons and therapists to monitor progress during rehabilitation.

Although there are a significant number of natural hand assessment procedures, the evaluation of hand prosthesis function is limited in scope. Moreover, results obtained from quantifying the effectiveness of these devices are largely obscured by a lack of reference to a benchmark. Consequently any objective evaluation of hand function should be capable of application to both natural and prosthetic hands.

The quantification of hand function during such procedures traditionally has occurred through various methods ranging from time measurement to subjective scoring. Although time is obviously an easy parameter to measure and manipulate statistically, it is not necessarily the most valid appraisal of hand function [3-5]. However without suitable alternatives there are few other objective methods by which to evaluate functionality.

Any hand assessment procedure must ensure that all ranges of grip are included, with direct relevance between prehensile patterns and the selection of activities of daily living. In order to develop a series of tasks based on prehensile patterns, the classifications of grip must be defined (see Figure 1). Although there appears to be little conformity to specific categories of grip description, the general characteristics remain similar. Based on the classifications of Kamakura [6], the following categories were used in the development of a new test, the Southampton Hand Assessment Procedure (SHAP): lateral, power, tripod, tip, extension, and spherical (also termed flexion).

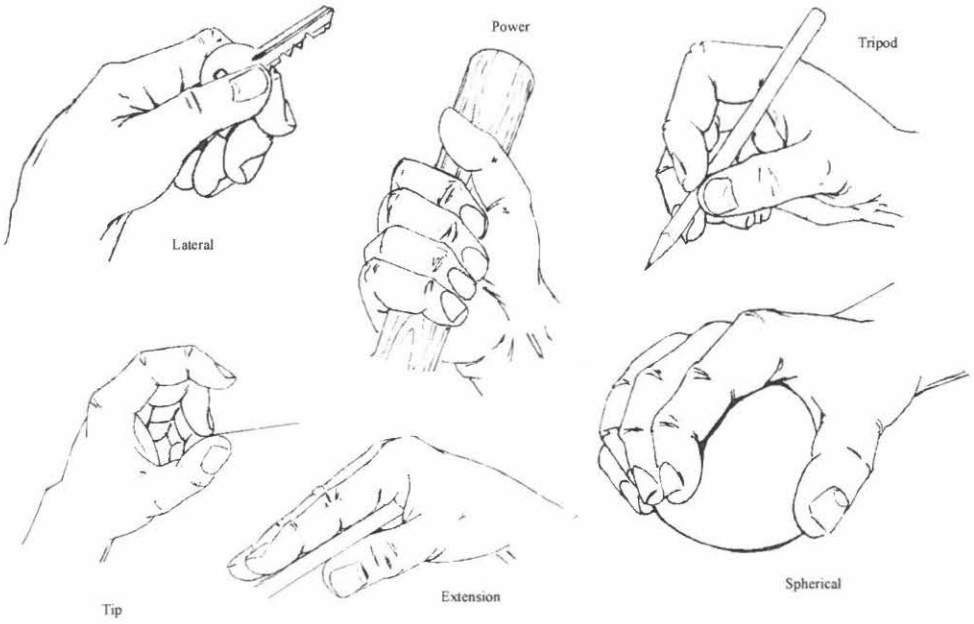

Figure 1 - Classification of Prehensile Patterns

The purpose of prehension classification for hand assessment tasks is to ensure the evaluation of a complete range of grip patterns, thereby providing a balanced assessment of functionality. Despite the differences in determining individual prehensile patterns, the vast majority encompasses the same range of movements regardless of classifications or terminology. Therefore their primary use is the assurance of adequate functional range assessment, rather than the identification of grip patterns during grasping.

SHAP Methodology

The stability of grip or deviation of prehension from a norm has frequently been evaluated by assessor opinion. The Southampton Hand Assessment Procedure [2] has been designed around the timed measurement of standardised tasks. The test requires the subject to perform a series of 12 'abstract' tasks, as well as 14 activities of daily living (ADL). The form board (or abstract) tasks are used to assess prehension without the complication of tools or equipment used during ADL that often cause intermediate grip patterns.

The form board objects are produced in two sets (see Figure 2a); a set of lightweight objects (predominantly balsa wood) for each of the six prehensile patterns, and a similar set constructed from aluminium. This enables assessment of not only the grip form but also compensates for subjects with weak grip pressure (thereby highlighting a difference in performance whilst executing the heavier object tasks).

Sollerman and Sperling [7] estimated the percentage use of their eight types of grip pattern during activities of daily living. The number of SHAP activities of daily living (see Figure 2b) utilising each grip was compiled in approximate proportion to these values to ensure that the assessment of functionality was directly related to the likely everyday use of each grip. This enables the measure of overall functionality to be obtained from a summation of results without the need for any weightings or adjustments of the data.

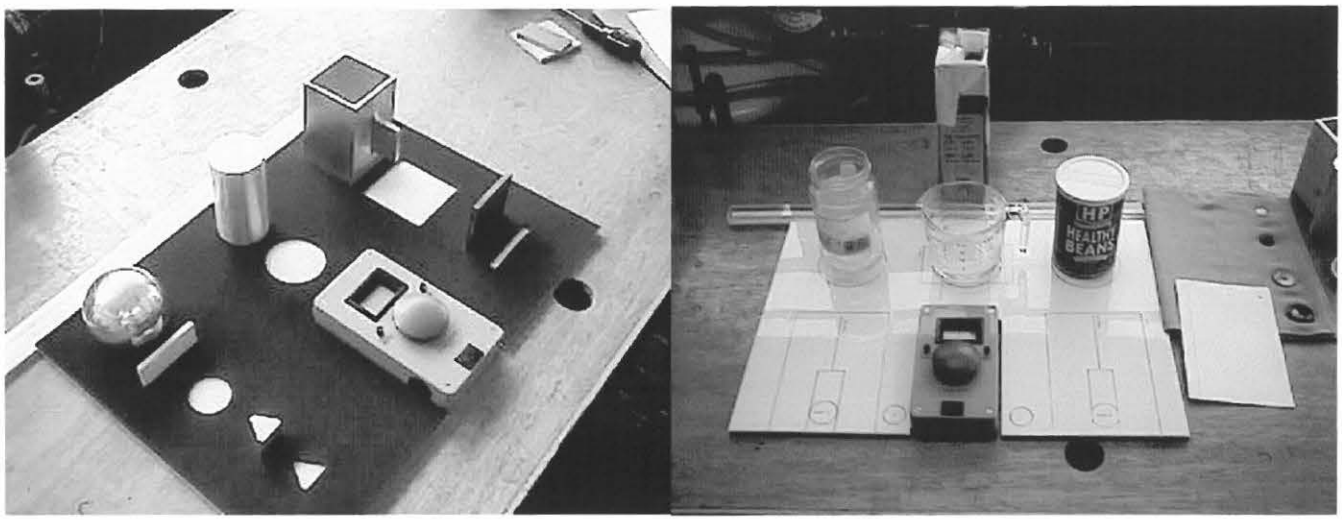

Figure 2 - (a) Abstract Object Form Board (b) Activities of Daily Living

Any activities requiring subjective assessment, or likely to cause a large variability in timing were omitted. For example, the task of writing, although an important everyday activity, was excluded on the basis of large variability in the performance of writing skills with no relevance to hand functionality (i.e. the speed of writing is not necessarily linked to hand dexterity).

Fourteen activities of daily living were selected based on these criteria (see Table 1).

Task Number	Task	Grip Classification	Task Number	Task	Grip Classification
1	Pick-up coins	Tip/Tripod	8	Lift large 'heavy' object	Power
2	Buttons	Tripod	9	Lift large light object	Power
3	Food cutting	Tripod/Power	10	Lift weighted tray	Lateral/Extension
4	Simulated page turning	Tripod/Extension	11	Rotate key 90	Lateral
5	Remove jar lid	Spherical	12	Zip	Lateral/Tip
6	Pour water from jug	Tripod/Lateral	13	Rotate screw 90	Power (with precision)
7	Pour water from carton	Spherical (flexion)	14	Rotate door handle	Power

Table 1 - Selected Activities of Daily Living

Normative data trials assessed nine males and nine females, with each evaluation replicated three times by the same assessor. Interrater trials were also carried out to ensure repeatability (and inherent standardisation) of SHAP. A combined normative database of 24 subjects, each with three replicate assessments, was subsequently used to provide the mean task times and standard deviations from which to evaluate the clinical subjects.

The variability of these times was found to be in increasing proportion to the mean time to complete the task. Hence any attempt to utilise the total task time or a mean overall time will cause an unbalanced result; for example if two tasks take 1 sec and 10 sec to complete respectively, then an arithmetic mean of these scores would be weighted much more heavily by the test taking 10secs. In order to reduce this weighting effect (which is likely to be more noticeable in pathological hand function), then the logarithm of each task time is taken. Thus the normative database consists of 26 individual task times

$$\mu_{total,task} = \log_{10}\left(\frac{1}{n}\sum_{n=1}^{24} t_{total,task}\right) \qquad \textit{Equation 1}$$

(μ_{task}), total time (μ_{total}), and standard deviations obtained from the times (t) of 24 normal subjects (n) according to Equation 1.

CLINICAL DATA ANALYSIS

At the time of writing, the hand assessment procedure has been used to assess 18 patients with varying hand trauma or disability, and one prosthesis user fitted with an Otto Bock myoelectric hand and wrist. The functionality of these patients has been quantified relative to 'normal' times by the use of z-scores and a relative scale. The z-score quantifies a patient's result in terms of standard deviations from the norm according to Equation 2.

$$z_{t,i} = \frac{x_{t,i} - \mu_{t,i}}{\sigma_{t,i}} \qquad \textit{Equation 2}$$

$$FS_{t,i} = 100 - (z_{t,i} \times FI) \qquad \textit{Equation 3}$$

where

z = z-score	μ = *normative mean time (from Equation 1)*
t = total	σ = *normative standard deviation*
i = 1,2,...6, representing the six prehensile patterns	*FS = Functional Score*
x = subject time	*FI = Functional Index*

This enables a measure of 'deviation' from the norm for each patient. A z-score is obtained for the overall assessment time (z_t), for each of the prehensile patterns (z_i), and if necessary, for each task. The total time z-score reduces a multidimensional problem from 26 variables to one, and therefore must be considered as an approximate quantification of function rather than an accurate representation. Hence, the assessor is provided with an overall approximation of function (from z_t), and then may subsequently choose to study the prehensile patterns (z_i) to highlight specific areas of functional difficulty (whilst the analysis of individual tasks is warranted only for the study of specific anomalies in test results).

In the case of a subject taking an excessive period of time to complete, or unable to accomplish the task, then a boundary must be introduced. Other assessment procedures [3, 8] have imposed boundary times and conditions without consideration of the individual nature of the task. For example Jebsen [3] limited subjects to 80 secs for all tasks that ranged from 'writing' (with a normal mean time of 12 secs) to the 'moving of large light objects' (with a normal mean time of 3 secs). However a more accurate estimate of the limit is obviously the time beyond which one can assume minimal function. Given consideration of previous functional assessments, as well as an analysis of existing clinical data, a boundary factor of 20 times that of the norm was imposed. Subjects were allowed to perform the task, but their times were subsequently limited if necessary.

Having imposed such a boundary condition, the z-score measure can then be converted to a sliding scale in a similar method to that of IQ ratings. The normative mean time is centred about a score of 100 on a new functional rating scale, according to the formula in Equation 3.

The functional index is used to equate a unit of 'functional score' to the number of standard deviations from the norm. This figure is determined based on the assumption that time in excess of 20 times the norm results in zero function. By multiplying the normative times by this limit, and using Equation 2 as before, the z-score obtained is approximately 23 (i.e. a subject performing each task at the boundary condition will be twenty three standard deviations from the norm). Given the functional rating range of zero (for non-functional) to one hundred (for normal function), then each standard deviation from the norm equates to $^{100}/_{23}$ rating points, which is represented in Equation 3 as the Functional Index. The current index value has been rounded to 5, however further clinical trials are expected to contribute to the refinement of this value. Although of no impact to the relative functional rating of each subject, the index ensures that all subjects remain above the realistic function boundary of a zero score.

Example results are given below for a section of the clinical trial group (Table 2). As can be seen from the data, although an overall functional score may indicate a level of dysfunction, the specific prehensile patterns may hold more information as to the true difficulties that the subject encounters. For example, subject E (suffering from a fractured wrist) encounters little relative difficulty in performing fine precision tasks such as tip and tripod grips, yet has notable adversity in carrying out lateral grip tasks. This is potentially due to the pronation/supination of the forearm usually associated with these tasks that also effect wrist movement.

The data also highlights specific difficulty for the prosthesis user, who as expected has minimal functionality compared to the rest of the group. This subject has severe impairment in the performance of power grip tasks, which is likely to be attributable to the limitations of the single degree of prosthesis.

Subject	Dysfunction	Overall Functional Score	Prehensile Group Functional Score					
			Spherical	Tripod	Power	Lateral	Tip	Extension
A	Flexor tendon injury (middle and little finger)	82.34	82.35	81.89	84.33	84.47	80.22	83.66
B	Dypytrens Disease Release	83.14	80.32	84.68	79.67	81.73	87.72	87.50
C	Proximal IP joint arthrodesis (little finger)	82.56	84.54	82.77	79.48	76.71	82.80	85.89
D	Flexor tendon injury (middle and little finger)	77.89	77.06	80.08	81.24	76.08	80.81	80.55
E	Fractured wrist	78.5	83.59	82.42	72.73	66.61	90.33	84.74
F	Amputation (index, middle and ring)	64.78	68.33	64.56	65.78	53.03	70.98	72.00
G	Below Elbow Prosthesis User	32.61	33.41	22.97	21.50	25.21	32.98	42.16

Table 2 - Functional Ratings of Clinical Data Sample

CONCLUSION

There are a significant number of natural hand assessment procedures in existence, however few address

the requirements of objectivity and standardisation. These factors are necessary to provide a reliable quantification of hand function for use as a clinical outcome measure. Conversely the measurement of hand prosthesis function has tended to centre on engineering assessments that fail to provide any reference or benchmark for the results obtained. Consequently the Southampton Hand Assessment Procedure has been developed to address the broad-based requirements of evaluating hand function, whether natural or prosthetic. The procedure aims to cover a full range of prehensile patterns by the use of both abstract objects (designed to encourage specific grip forms), and activities of daily living, totally 26 individual tasks.

Normative data trials were carried out to enable the subsequent evaluation of impaired hand function with respect to a benchmark standard. A mean total time to perform all tasks, and mean times for each of the 26 tasks, was calculated along with associated standard deviations. Each task has an associated prehensile pattern classification (lateral, power, tripod, tip, extension or spherical), and these times were summed to form a result for each prehensile group.

The results of the impaired function subject group was then quantified with respect to this normative database. A z-score is obtained for the overall assessment time, and for each of the 6 prehensile patterns, which rates the subject's performance by standard deviations from the norm. This is then converted to an arbitrary functional rating scale, centred about a normal result of 100.

The ability to provide an overall measure of hand function is important, but perhaps secondary to the ability to identify specific areas of functional difficulty, as produced by the prehensile pattern group results. The procedure is not aimed at quantifying the quality of life of the subject, nor indeed the subject's overall functional ability. Instead the emphasis lies in evaluating the performance of the hand, whether natural or artificial, so that rehabilitation, treatment, or the focus of additional research (in the case of prostheses) may be better evaluated and directed. The continued clinical trials of this procedure are hoped to aid in the refinement of the statistical evaluation whilst simultaneously providing valuable information to therapists and researchers alike.

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EXPERIENCE WITH HIERARCHICALLY CONTROLLED HAND PROSTHESIS

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INTRODUCTION

The principles of hierarchical control of hand prostheses have been described in previous MEC meetings [1]. Simply put: It is possible to achieve easy use of a multi-axial artificial hand if the detailed control of the hand's grip posture and force is devolved to a microprocessor controller. Thus the controller is given simple instructions to open and close the hand and the controller decides on the grip posture based on the contact between hand and object and then adjusts the grip force depending on any motion of the object relative to the hand. Thus if the object slips the hand automatically grips tighter.

This idea has been considered for many years [2,3]. Now it is sufficiently well taken that similar methods are being applied by prosthesis producers [4]. The detection of the slip is achieved using an acoustic method [5], where the sound of the start of the slide is detected. Although this may not mean that the entire object is moving relative to the hand, but only parts of the object, so called *partial* slip [6]. A second method is that of measuring the change in the contact forces and inferring slip from those changes, [4,7]. This method can be confused by a change in orientation of the object or hand that occurs *without* slip.

The concept has been realised by a number of designs of prostheses in the past. They can be used with different numbers of degrees of freedom, from one through to four (and most recently six) [3,7,8,9,10]. The basic control remains the same, the only difference is the number of different types of basic grip form can be achieved. For example: In the two degree of freedom hand, the controller can select from precision or power type prehension. Additional degrees of motion allow the controller to add, two or three jaw precision, lateral prehension and a more compact form of the power grip [7,11]. The trigger between the different grips is the point of first contact between the hand and the target object during the reaching phase of a manipulation.

The current generation of device that has been produced at the Oxford Orthopaedic Engineering Centre has been designed in order to be produced in small numbers and to be used extensively in the field. To that end a lightweight, robust, device was designed an tested and is now undergoing field assessment at Oxford [12].

PROSTHESIS CONTROL

To make it possible to construct and use the device the design choices centred on the ability to produce a hand light enough to be worn conventionally. Thus lightweight materials were used and a three finger, two degree of freedom format was chosen. More details of the basic device are in [1,12].

Some of the significant mechanical design features are: The digits curl continuously from straight to fully flexed. The thumb abducts with flexion. The mechanism is reversible. The thumb can be braked or released easily when the hand is deactivated.

The electronics are in a separate package which includes a microprocessor. The input electronics (such as amplification or filtering) are separated so that it can take a verity of different input forms: EMG, switch, FSR. The essential difference with the control of the hand to the more conventional schemes is that the degree of flexion is proportional to the level of input (such as EMG tension). It is thus more akin to a body powered device, voluntary opening/involuntary closing. It uses a second control input (such as the opposing muscle) to be used for the control of the hand in the HOLD mode (where the automatic slip/grip reflex is used). Application of the tension switches into HOLD. If the operator then wishes to override this reflex they reapply the second EMG channel. Any further tightening of the hand will only be proportional to the tension in the second muscle.

This makes the control of the hand progressive and logical. The hand is always opened with one muscle and hold switching and squeezing is with the second. As the hand places the lightest possible force on the object the operator only has to let the hand close on its own, without close supervision.

Microprocessor control makes the device more readily adaptable and allows the information about the hand, controller and user to be made available, this telemetry line can be used in training and diagnostics. Monitoring of the input channel and its use in training is a well established technique, both in the experimental [9] and the commercial [13]. It is the logical extension of the use of analogue meters used as a training aid. These, however, do not exploit the full potential of the system. It is possible to provide the telemetry link that informs of the inner state of the hand and this can drive a computer display [12]. This display can drive a simulation of the hand or drive a game to make training more interesting [14]. The link can also output data on the controller states and the sensory inputs. When this link is run over a wireless connection the user can undertake standard tasks while the therapist can observe their progress.

While the control is different to that of a conventional device it is designed to be easily learned and progressive in nature. Extensive use of computer interfaces are used in the training and assessment of the device. In addition the Southampton Hand Assessment Protocol (SHAP) [15], is used to monitor the subjects' progress.

The control of the hand is quite simple but it does have differences to the conventional myo-electric hands. As observed the control is single sided, more akin to body power than myoelectric. The prehension control is proportional, the opening hand proportional to tension. A relaxed extensor means that the hand closes. The slip-grip control is turned on by tension in the opposing muscle. The reflex is also turned off and over-ridden by re-application of the same muscle. Thus a delicate object can be held or extra tension voluntarily placed on the target. The releasing of an object can be made more deliberate as the threshold tension is greater to open the hand when it is in the hold state.

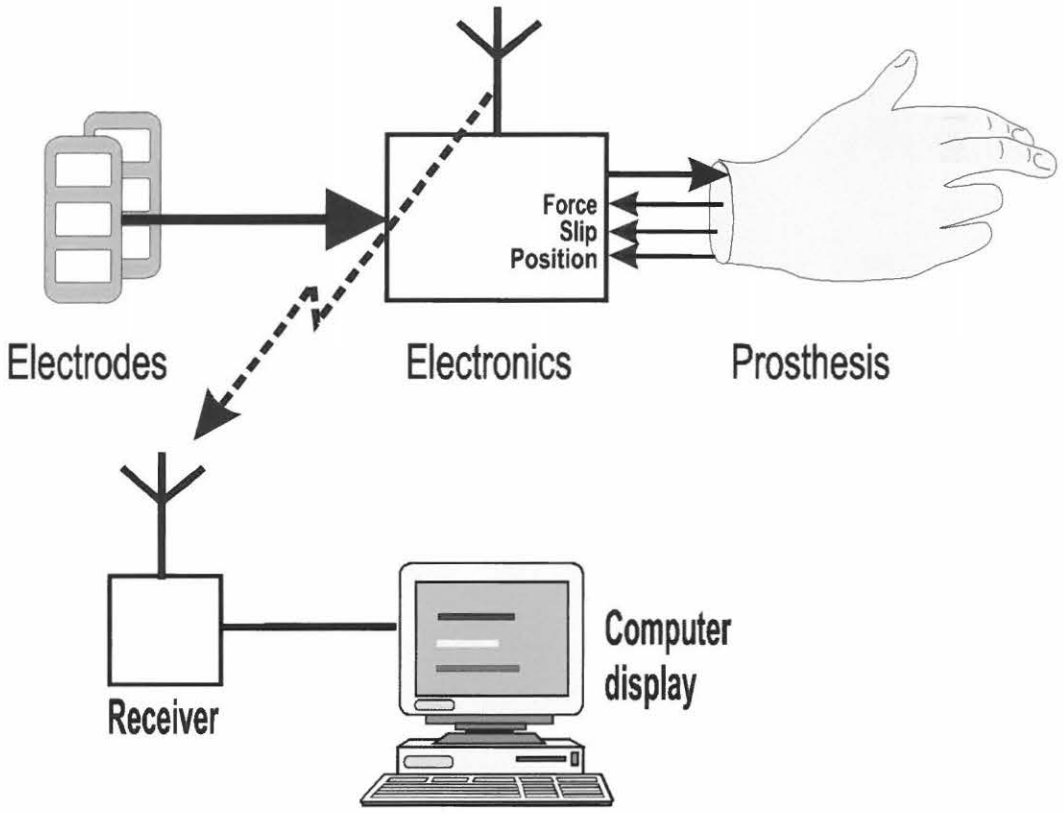

Figure 1 - Telemetry system for the LO/SH Hand

This difference means that some experienced myoelectric hand users can take time to adjust to the difference. Assuming that the user employs the conventional extensor/flexor opposing muscle pairs, then using the telemetry display, experienced myo-users can be observed to driving the hand closed like a standard voluntary opening voluntary closing hand in vain. This has no effect on the hand controller. As the commanding extensor muscle is relaxed the hand will close irrespective of the state of the flexor muscle. However, before the operator is allowed to switch the hand into the hold with their flexor mode they must relax it first. The apply the tension so that it cannot slide into a holding state accidentally.

So seen from the outside it does not make any difference to the motion of the hand. However it has been observed that with the slow voluntary opening and voluntary closing method employed by conventional hands the operator opens the hand wide and early in a reach phase of the manipulation [16]. Natural hands are opened much later in the reach, and then only about 10% greater than the minimum required. Once it is appreciated that the hand can be opened quickly and precisely and much later in the operation and without the detailed conscious control of the conventional system.

Even so those who have no preconceptions about myo-electric control because they have no experience prior to their exposure to the hand they have little difficulty in grasping the concepts.

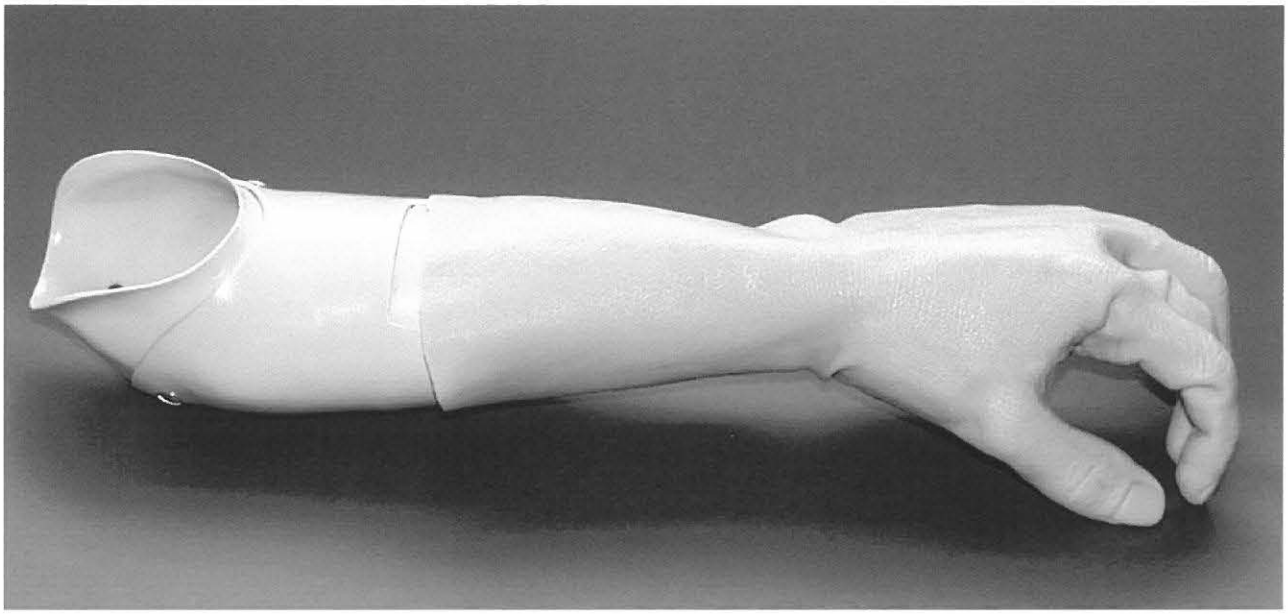

Figure 2 - LO/SH Hand

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EXPERIENCE WITH THE RIMJET LOCKING HUMERAL ROTATOR

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In our experience, persons with bilateral arm amputations greatly benefit from devices that allow flexibility in positioning of the terminal device in space. We have reported our experience with the Four-function Forearm Set-up (FFS) where cable control of wrist rotation and wrist flexion are added to the conventional body-powered trans-humeral configuration [1]. We have recently added humeral rotation to the prosthetic design used by three of our experienced clients (prosthesis users for more than five years) who have high-level bilateral amputations.

At present, these individuals consider this new component as an improvement. Two of these clients present with bilateral trans-humeral amputations and one has bilateral shoulder disarticulations. For the two with bilateral trans-humeral amputations, it was necessary to use a conventional laterally-routed cable in order to retain the functional advantages of the Four-function Forearm Set-up. This required trade-offs in the positioning of the humeral rotation device.

The manufacturer of the locking rotator recommends a medially-routed cable [2]. With this routing, the forearm generally moves toward internal rotation when the rotator is unlocked because of tension applied to the control cable at the time of unlocking. Since a medially-routed cable is not compatible with the Four-function Forearm Set-up, the ease of internal humeral rotation was compromised somewhat for the trans-humeral fittings. This problem was resolved in one case by the addition of an external spring which biased the unit into internal rotation. This problem is not an issue for the shoulder disarticulation client because he uses a nudge control for unlocking the humeral rotator and therefore the control cable can remain slack allowing easy internal rotation by gravity.

We are currently investigating two modifications to the Rimjet Locking Rotator. The first is the addition of an internal spring mechanism to facilitate internal rotation. The second modification is to increase the number of locking positions. Our clients felt that there were too few locking positions on the standard Rimjet unit. We will discuss these issue and others found in fitting this device.

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EXPERIENCE WITH CUSTOM SILICONE SUSPENSION SLEEVES FOR SELF SUSPENDING TRANS-HUMERAL AND TRANS-RADIAL PROSTHESES

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Upper limb amputees attending our prosthetic clinic with trans-radial amputations have been routinely fitted with hard shell self-suspending sockets for many years. It has not been until more recent times and the development of the use of silicone materials that it has been possible to design an effective self-suspending prosthesis for some trans-humeral amputations.

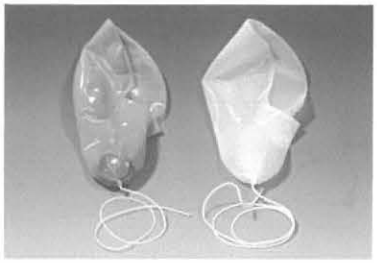
While it is possible to make and fit a hard shell self-suspending socket for a suitable trans-humeral amputation practical difficulties arise in fitting and retaining suction. The socket has to be an extremely accurate fit with absolutely no possibility of leakage. Unlike the lower limb suction socket gravity is acting against the socket at all times, a one way valve as used in lower limb sockets to expel air is of no use, any loss of suction will result in the loss of prosthesis retention and the need to reapply the socket. The range of mobility of the arm and the subsequent change of cross sectional shape throughout this range of movement compounds this problem.

Donning of this type of socket can be difficult with one hand, as a pull in will always be necessary. Applying a suitable bandage wrap and applying the prosthesis correctly is not an easy task for a one-arm person. Hence, although first fitted more than two decades ago the hard shell self-suspending socket has never really been successfully used.

A number of years ago one of the authors experimented with the use of polyethylene trans-humeral sockets for self-suspension. When bubble draped over a trans-humeral cast and suitable thinned this material is flexible and to a certain extent will accommodate changes in volume.

Suspension with this type of socket was excellent but the problem of donning still arose and unexpectedly it proved extremely difficult to remove the sockets after a period of time. If the amputee removed the socket within a few minutes, it could be removed without difficulty. If left long enough for normal body perspiration to build, it proved very difficult to remove the socket and sometimes patients participating in the trial had to pour a soapy water mixture inside the socket to allow removal. Because of these difficulties, the polyethylene socket was rarely used.

Six years ago a female trans-humeral attending our clinic said she would desperately like to be rid of the uncomfortable and non-cosmetic suspension straps on her prosthesis. Thinking back to problems previously encountered, it was obvious that a socket material that was easy to don/doff would have to be extremely flexible and perhaps have the ability to stretch. Silicone rubber seemed to be a material that would fit the requirements.



Custom trans-humeral liners

Self-suspending trans-humeral

A negative plaster of Paris wrap cast was taken and the positive cast produced from this reduced in circumference. Through trial and error, this reduction was worked out at 25%. This may seem a rather large amount but it must be remembered that the silicone sleeve produced is only 1.5-umm in thickness and stretches very easily.

Materials used in the lamination of the sleeve are Otto Bock or any other brand of similar chemical cure silicone, ladies stockings for the lay up material and braided nylon cord obtained from the local boat chandlers.

A PVA barrier is applied over the modified positive cast in the normal way and one layer of stocking applied. The braid of the cord is unravelled and laid longitudinally along the laminate (the cord is non elastic and hence prevents longitudinal stretch of the liner. The second layer of stocking is applied and covered with a PVA sleeve and the silicone mix poured. When cured the liner is cut from the cast (the cast is left intact and can be used for duplicate liners).

The next stage is a patient fitting where the liner is applied to the stump and tested for retention. If everything is satisfactory a layer of "cling film" is applied over the liner, tension applied to the cord (to simulate the lanyard tension and shape the stump correctly) and a plaster wrap taken to produce the outer humeral shell.

Various types of humeral shell have been used, frame or full lamination depending on individual requirements. A number of methods of securing the pull in cord have been tried but we now routinely use the excellent small lanyard securing cleat produced by Össur, Iceland.

Results using this type of suspension have been excellent. No skin problems have been experienced. Donning and doffing is easy. Retention has not been a problem. The original patient fitted has now been using her prosthesis continually, 14-16 hours every day, for almost 6 years. She wears her prosthesis during exercise sessions, including step aerobics, and has had no failure of retention.

Since that time many more trans-humeral through to elbow disarticulation levels have been successfully fitted. Most have been lightweight prosthesis intended mainly for cosmetic purposes but a small number of myo-electrically controlled externally powered devices have also been successfully fitted. The silicone sleeves last on average 6-12 months before requiring replacement.

Silicone sleeve for longer stumps

A total of round 40 silicone sleeves have been fitted with no rejections. Universally the statement has been "why couldn't I have had this sooner" or "please never ask me to return to straps again".

"Off the shelf" silicone liners are now commercially available but these have originally been developed for lower limb use and are much thicker and more difficult to don. It must be remembered that unlike most lower limb amputees the unilateral arm amputee has only one hand to assist in donning the liner. The remaining muscle and soft tissue at trans-humeral amputation level is usually mobile and floppy. When trying to don these thicker liners it can be very difficult to stabilise the tissue. Several of our amputees have tried these sleeves and all but one prefers the thinner custom-made version.

Problems experienced with commercial liners have been difficulty in donning, and blistering of stump tissue.

Most trans-humeral stumps have fairly "floppy" tissue and it is difficult for the amputee to don the thicker commercial sleeve straight. When tension is applied to the pull in lanyard or shuttle pin the sleeve is straightened and a constant tension applied to one side of the stump tissue. I believe that this is what causes the blistering problem.

One disadvantage of the custom made sleeve is the time involved in manufacture and the fact that at least two casting sessions are required. It is the opinion of the authors that it would be possible to develop a standard range of "off the shelf" liners which, like those developed for lower limb amputation levels, would fit the majority of stumps. If this were to happen it would be possible to produce a complete prosthesis in a shorter time with only one plaster cast required.

Trans-humeral congenital

Silicone self-suspension

More recently further development of the technique has taken place on trans-radial applications, both adult and children. An example of this is a young woman who requested a more cosmetic translation line between her supra-olecranon self-suspending socket and her arm that would not show through her wedding dress. Using the thin silicone liner for suspension, we were able to produce a suitable prosthesis. The technique has been particularly useful when used on infants both trans-humeral and trans-radial and examples will be shown during the presentation.

While this type of suspension has much to recommend it, like all upper limb prescriptions, careful consideration must be given to the actual needs of the amputee both psychological and functional.

If function is a high priority and body power is the preferred method it may be, for some, that little is to be gained in using silicone sleeve over harness suspension. If externally powered components are used then consideration should be given to the additional mass and the subsequent rotational forces applied to the socket. It is very difficult to effectively stabilise the rotational forces within a trans-humeral socket relative to the humerus without the assistance of suspension straps.

CLINICAL APPLICATION OF ROLL-ON SLEEVES FOR MYOELECTRICALLY CONTROLLED TRANSRADIAL AND TRANSHUMERAL PROSTHESES

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ABSTRACT

Three transhumeral and five transradial prostheses have been successfully fit to patients using electrodes installed in roll-on sleeves. New metal electrodes and wiring have been developed and tested along with techniques for installing electrodes so that the sleeves are not damaged. It is now possible to fabricate a system where the sleeve lasts for one year. The roll-on sleeve is a superior way to achieve suspension, especially with patients who change weight.

BACKGROUND

It has been the goal of myoelectric upper limb prosthetics to restore as much function as possible to the amputee. With the use of roll on suction liners the suspension and function of a myoelectric prosthesis can be much improved. This article based on four transhumeral and five transradial fittings which have lead to improved suspension and function for these below elbow and above elbow amputees.

The difficulty with myoelectric fittings has been to maintain suspension and yet provide access to the electrode sites for a prosthesis. The electrodes need to maintain good contact with the skin and not shift as the person moves the arm and lifts objects. With conventional fittings this has required great care in fitting and a very stable limb volume. Roll-on suspension systems provide a secure attachment for the prosthesis and have proven successful in lower limb fittings. Roll-on suction liners are made from polymer materials and will not conduct the myoelectric signal. They provide excellent suspension, but do not allow myoelectric control of the prosthesis. Several methods have been proposed to gain control including cutting holes in the liner and applying conductive posts through the liner. All of these methods have drawbacks and have not proven successful in our facility. This presentation will present a method that provides both good suspension and myoelectric control for the prosthesis.

DESCRIPTION OF A SUCCESSFUL SUSPENSION SYSTEM

A Liner that works.

Ohio Willow Wood Alpha™ liners were chosen for this system after experimentation with many other liners. The Ohio Willow Wood Alpha liner is different from other roll-on liners in that the material is a thermoplastic and has a fabric covering. The fabric cover eliminates the need for a lubricant to don the liner and is easier for a one

handed individual to control. The thermoplastic material allows the shape to be customized and if an error in electrode placement is made the holes can be "healed" with a heat gun or adhesive to eliminate air leakage and loss of suspension.

Passing electrode signals through the liner.

The difficulty with all liners is to get the myoelectric signal from the skin to the electronics. With this system electrodes pass through the liner and are connected to the preamps with snap connectors and shielded cables. At the present time the electrode snap connectors are hand fabricated from Liberty Technology reference electrodes or male clothing snap heads. A 2-56 screw pierces the liner and provides the conductor to the shielded cable. Standard shielded EKG cables or cables from Motion Control are used to conduct the signal to the preamps which are modified as shown. For transhumeral fittings the cable is connected to directly to the preamps.

Similarity to a standard myoelectric fitting.

Electrode placement is determined as with any myoelectric system and the electrodes are placed in the liner over the selected sites. The wires are contained inside of the socket which is bulged out slightly over the electrode area to allow for their bulk. The preamps can be located at any convenient location in the prosthesis. The wiring is the same as usual from the preamps to the hand and/or elbow. If there is insufficient length for a shuttle lock, the Velcro-on-a-lanyard system will provide a secure suspension. The proximal socket shape is modified to prevent rotation and allow good range of motion. With the improved suspension from the suction liner a greater range of motion can generally be achieved by opening up the proximal trim lines without the loss of suspension. The remaining wiring and fitting proceedure is the same as for a typical myoelectric prosthesis.

Table 1. Transhumeral Patients

Patient	Cables	Electronics	Metal Electrodes
EN	Shielded	Liberty Elbow	Liberty Technology
RP	Utah	Utah Elbow	Liberty Technology
PC	Utah	Utah Elbow	Liberty Technology
JS	Utah	Utah Elbow	Utah and Liberty Tech.

Table 2. Transradial Patients

Patient	Cables	Electronics	Metal Electrodes	ROM	Pull off
GC	Custom	Bock preamps	Liberty Technology	5-120°	45 lb.
EM	Custom	Bock preamps	Liberty Technology	0-135°	45 lb.
RF(1)	Utah	Bock preamps	Gold Plated	5-125°	50 lb.
GJ	Utah	Bock preamps	Liberty Technology	5-110°	37 lb.
DC	Utah	Bock preamps	Stainless Steel Snap	115-110°	50 lb.
RF(2)	Utah	Bock preamps	Stainless Steel Snap	5-130°	50+ lb.

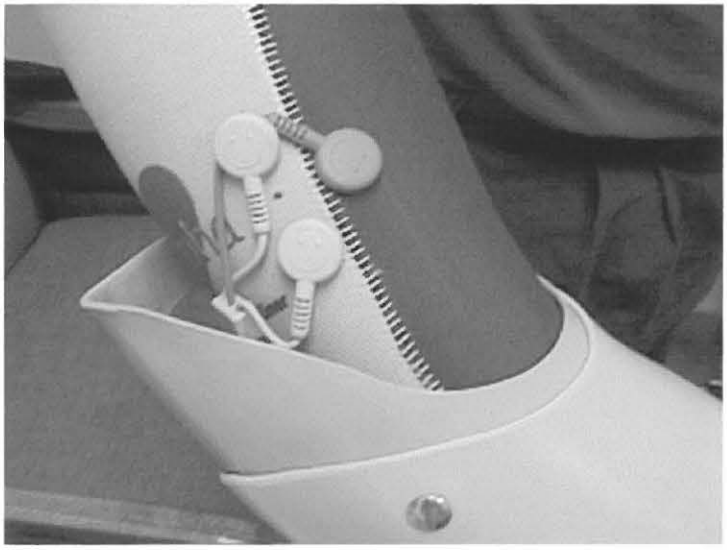

Figure 1. *The Motion Control wire harness attached to an Alpha sleeve. There is a need to reduce the size of the snaps and the splice.*

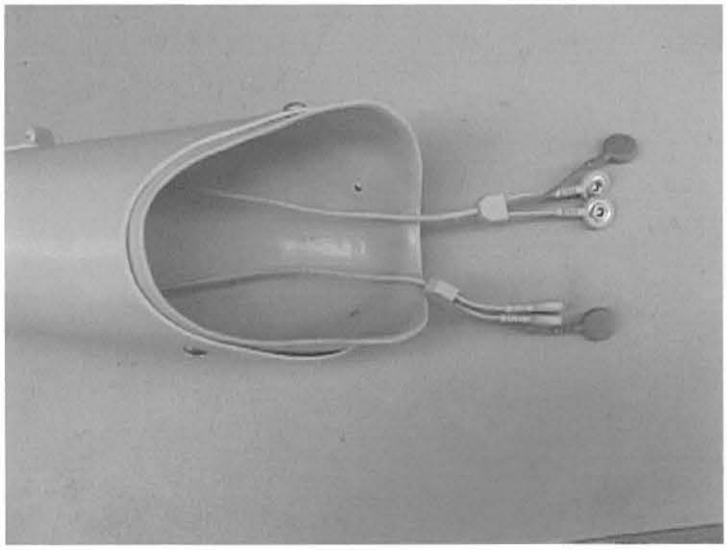

Figure 2. *Two wire harnesses. The user will snap these to the stainless buttons onto the sleeve while inserting the sleeve covered arm into the prosthesis.*

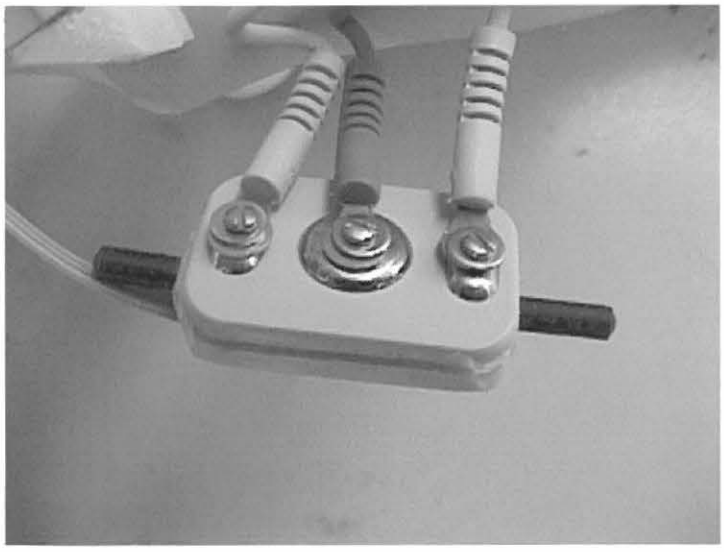

Figure 3. *The Motion Control wire harness is attached to an Otto Bock 13E125 Electrode with miniature screws.*

NOTES

THURSDAY, AUGUST 26, 1999

8:45 am — *Daily Notices*

9:00 am — **Keynote Address:**
David Gow,
Princess Margaret Rose Hospital, Edinburgh, Scotland

10:00 to 10:30 am — *Refreshment Break*

10:30 am to 12:10 pm — **Intelligent Control Systems:**

A Totally Modular Arm Prosthesis
P.J. Kyberd

The Development of an Advanced Multi-Axis Myo-Prosthesis and Controller
C.M. Light and P.H. Chappell

Teleassistance for Electronic Components in Rehabilitation Technology
A. Davalli, R. Sacchetti, and P. Ferrera

Forequarter Prosthesis with Interchangeable Elbow, Forearm, and Hand
R.D. Lipschultz

12:10 to 1:30 pm — *Lunch*

1:30 to 2:45 pm — **Current Research:**

Optimal Fixed Wrist Alignment for Below-Elbow, Powered Prosthetic Hands
J. Landry and E.N. Biden

Magnetic Resonance Imaging of Congenitally Deficient Upper Limbs
K.A. Farry, L.K. Kramer, R.K. Gupta, D. Atkins, and W.H. Donovan

The Use of the Hilbert Transform in EMG Analysis
S. Taffler and P.J. Kyberd

2:45 to 3:15 p.m. — *Refreshment Break*

3:15 to 4:55 p.m. — **Myoelectric Control and Prostheses:**

Fuzzy Logic in the Interpretation of EMG Signals for Prosthesis Control
S. Tafler and P.J. Kyberd

Natural Control of Key Grip and Precision Grip Movements for a Myoelectric Prostheses
M.C. Santa-Cruz, R.R. Riso, and B. Lange

A PC Based System for Selecting and Optimizing Myo Controls to the Patient's Needs
G. Haslinger

Low Level Response to Bock and Steeper Electrodes
C. Wallace, T.W. Williams, III, and N. Taneja

5:15 pm — *Shuttle Bus from UNB to Sheraton*

THE DEVELOPMENT OF THE EDINBURGH MODULAR ARM SYSTEM

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BACKGROUND

Historically, the major research programme undertaken by the Bioengineering Centre in Edinburgh was the development of pneumatic powered limb prostheses for children affected by Thalidomide. From 1963 to the early 1980's the Centre's main prosthetic work was the design, development and fitting of carbon dioxide powered limbs to children from Scotland and Northern Ireland. This work was carried out under the leadership of Professor David Simpson. Work on these prostheses was eventually discontinued and new work began on body powered and electrically powered components for unilateral amputees.

INTRODUCTION

In 1986 the prosthetic programme became the responsibility of the author who rationalised the work to combine the Centre's various strands of prosthetic research into an integrated programme, based on modularity. It was recognised that despite the undoubted merits of the gas powered arms, there were many reasons for altering the direction of the work of the Centre. Not least was the observation that there was no real, continuing demand from the users of the pneumatic prostheses. Undoubtedly a number of factors influenced this including compensation packages and other outside pressures affecting the lives of the users and their families. From a functional viewpoint, however, it was observed that the Simpson arms were complex and required a high degree of maintenance. In addition, the supply of pneumatic gas cylinders was inconvenient and cumbersome but the core reasons centred on limitations in functionality when compared to using the lower limb with the inherent tactile acuity.

Logistically the Edinburgh work had concentrated on designing for bilateral absence and this limited the general applicability of the resultant technology. There had also been significant developments in electrical prostheses which had shifted the focus of prosthetic research and service provision to this power medium. There were still, however, serious gaps in the available hardware. For example, a survey of available componentry in the mid 1980's indicated a shortfall in the availability of components for functional partial hand replacement or for shoulder disarticulation or forequarter absence. It was also clear that the commercial systems which were available were broadly incompatible with each other and offered the Scottish Health care system a variety of components with different cosmetic appearances and with major planned obsolescence, in the form of children's prostheses and passive prostheses with work hardenable fingers.

DESIGN GOALS

In particular, consideration was given to looking at designing versatile componentry to encompass as many of the patient groups as possible. It was felt that it should be possible to develop common components for use with different power regimes without compromising function. This work lead to the concept of designing components with structural neutrality to allow the elimination of left and right handed units and the design of modules for the most

distal level of absence. The end result was hoped to be the transferability of components between age groups with for example adult elbows being used as children's shoulders. Simply stated, the design aims can be summarised as follows:

- designing based on the "shortest and smallest" principle
- neutral symmetry of form to eliminate "handedness"
- transferability of parts between power regimes and age ranges to reduce component count
- development of a "kit of parts" approach to allow flexibility, experimentation and re-use

INITIAL WORK

Initially there was a desire to look at developing electrically powered versions of arm devices based on the Simpson-Edinburgh pneumatic arms. The work was originally promulgated at the First International Workshop of Rehabilitation Robotics in Ottawa, Canada in June 1988 [1]. Prototype units were built and research grants obtained from the Scottish Office's Chief Scientist in 1990 and work continued full time until 1995 when funding eventually ceased. It became clear from early fittings that although the work showed considerable promise there were serious limitations to the degree of modularity and the prototype was only suitable for adult males. In addition the nature of the structural design meant that there was little or no room for on-board batteries.

The power of the concept of modularity was illustrated when a solution meant to alleviate size problems in partial hands was applied to the full arm system with exciting results. One of the aims of the modular approach was to try and produce powered partial hands for adults and children to meet the perceived clinical need. A number of trial fittings in Edinburgh, Örebro and Gothenburg had shown that the design was only suitable for patients aged nine or over [2]. In attempting to find a design which would allow smaller prostheses to be developed a hand was built from powered prosthetic digits. The thumb and finger phalanges were constructed from housings containing motors and gearboxes driving worm and wheel pairs. Unlike previous designs, however, the worm wheel was held fixed and the motor/gearbox phalange rotated instead giving a compact and neat prosthetic actuator system. Using this approach, prostheses suitable for children of four years of age became possible. In addition the hands were structurally neutral and left and right prostheses could be built from a "kit of parts".

When this same design idea was applied in an attempt at solving the perceived problems of arm development, larger power units were constructed to give elbow and shoulder joints. The motors and gearboxes could be fitted inside a carbon fibre tube thus freeing up valuable space around the diameter which could be used to mount batteries and effectively created an endoskeltal system. In addition the compact nature of the design allowed shorter actuators to be built and allowed limb segments to be cut to length to suit children and adults. The modular approach seemed to yield the same advantages for high level absence as it did for more distal absence.

CURRENT STATUS

Although external funding ceased in 1995, work has continued on a part time basis as part of the organisation's service development strategy. This has meant that progress is slower than we would have preferred, and to some extent the work has concentrated on the externally powered aspects of the modular concept. To date the two major manifestations of these design principles have been the development of a powered prosthetic digit system known as PRODIGITS™ and the construction of an endoskeletal arm system known as the Edinburgh Modular Arm System or EMAS. A prototype for high level unilateral absence was fitted in August 1998 to a 47 year old man. The prosthesis contained powered shoulder, elbow, wrist and hand functions and was controlled by microswitches and pressure pads activated by residual movement of the acromion. The prosthesis is still in working

order over 10 months later. Redesign work and laboratory testing are currently underway based on the valuable feedback obtained from the user's comments Further volunteer patients have now been identified to test all levels of the system's configurations.

FUTURE PLANS

A small clinical trial of the partial hand configurations of the Edinburgh Arm System is planned in conjunction with colleagues in Nottingham, England. Other Centres in Chicago, Örebro, Gothenburg and Belgium have expressed an interest in participating.

Further fittings of the complete arm system are planned before the year's end to an adult male in Manchester, England, an adult female in Glasgow, Scotland and to an 11 year old boy in Örebro, Sweden. New degrees of freedom for the wrist and shoulder are at the design stage.

Further work on body powered and passive prostheses is underway to test the possibilities of modular transferability further.

Researchers at the Bioengineering Centre are participating in a European Union project TOMPAW in collaboration with colleagues from Sweden and England.

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A TOTALLY MODULAR ARM PROSTHESIS

Peter J Kyberd, Oxford Orthopaedic Engineering Centre, Nuffield Orthopaedic Centre, Oxford.
OOEC, NOC, Headington, Oxford, OX3 7LD.
David Gow Director of Rehabilitation Engineering Services, Edinburgh, UK.
Lena Sperling, Leif Sandsjö, Christian Almström - Lindholmen Utveckling, Gothenburg, Sweden.
Andreas Boman, Stellan Brimalm - Brimalm Engineering, Vadstena, Sweden

INTRODUCTION

While some groups and centres have attempted to promote greater levels of sophistication in exo-prosthetics [1,2,3], it is only in the past couple of years that the ideas of computer control to be applied in prostheses has received a more widespread acceptance, [4,5]. It is probable that those at the clinical end have only recently become aware of the potentials within the electronics field. Now there are some systems that do incorporate some technology into their products. The latest trainer from Otto Bock is an example of the capabilities of such systems [6]. The VASI systems with their reconfigurable set up has begun to show the extent to which the technology can progress [7]. The TOMPAW project also aims explore this area more fully.

TOMPAW stands for TOtally Modular Prosthetic Arm with high Workability. It is an EU project under the TIDE initiative (DE4210).The projects target is to produce a fully modular and functional prosthetic arm system that will enable the users and the clinical teams to customise the device to the user with the minimum of effort and time.

THE CONSORTIUM

The team that has been constructed combines the three most advanced prosthesis technology projects in Western Europe over the past thirty years with the most exciting surgical program.

The partners are:

Stiftelsen Institute for Management of Innovation and Technology

A research foundation controlled by universities in Gothenburg and Stockholm, Sweden. IMIT is coordinating the project and provides expertise in the fields of technology managaement, work performance and user training.

Lindholmen Utveckling

Lindholmen Utveckling works closely with the Chalmers University of Technology in Göteborg, Sweden. It includes members of the Sven hand project [8]. Specifically, members of the team who worked on the pattern recognition system for the hand. This work used signals from arrays of muscles on the forearm of individuals with a traumatic amputation and utilised the existing phantom arm to control the prosthesis. For the small numbers that used the hand it was a promising technique. Limitations included the ability to carry the electronics needed to undertake the pattern recognition and the difficulty in repeating the electrode placement accurately. Such problems need not prove a barrier today.

Brimalm Engineering

An engineering firm in Vadstena, Sweden, specialised in fine mechanical design and manufacture.

Princess Margaret Rose Hospital

The upper limb research program in Edinburgh has run continuously since its foundation in 1964, (originally under David Simpson, and more recently under David Gow). The team have developed and evolved the concept of Extended Physiological Proprioception (EPP), where the feedback is appropriate to the task and allows as much of the internal control pathways of the user to be utilised efficiently [9,10]. Additionally they have undertaken programs of mechanism design and testing resulting in compact hand designs and the worlds first clinically fitted powered shoulder unit [11] and the development of improved silicon glove techniques.

Oxford Orthopaedic Engineering Centre

Since 1968, the concept of hierarchical control of a hand prosthesis has been promoted by members of Jim Nightingale's group in Southampton under the title of the Southampton hand [11,13]. More recently the concepts have been developed further towards a clinical system in Oxford at the Nuffield Orthopaedic Centre [14,15], while ideas exploring the control continue in Southampton [16].

The Institute for Applied Biotechnology

Professor Per-Inge Brånemark has led a team that has successfully created Osseo-integrated attachments for both artificial arms and legs [17,18]. This remarkable technique is still in its earliest phase and requires caution to ensure that it is applied safely and appropriately to the users. The result of the attachment is an arm without the restrictions and limitations of a conventional socket, including range of motion limitation and the reproducibility of control sites. In addition the feedback through the bones (Osseo-perception) means that the limbs are more part of the user and give them additional information unavailable before.

Sahlgrenska University Hospital

Finally, the team includes highly experienced members of the clinical service in Gothenburg, Edinburgh and Oxford so that this knowledge and experience can be drawn on during every phase of the project.

THE PROJECT

The project is divided into three phases. The first is the design phase where knowledge and experience of users in the three centres along with the providers is collected and refined into a design specification [19].

The second phase is the production of the first prototypes for simple evaluation, before the final field prototypes are produced and users will to try them for extended periods of time. At all points the user opinions will be sought and evaluations be made using the most appropriate tools, such as the Southampton Hand Assessment Protocol (SHAP), [20].

Users and professionals outside the project team will be invited to follow the progress of the project on the project web site: http://www.lindholmen.se/tompaw

A Modular arm

The arm will consist of standard modular units and builds on the experience of the PMR team, [11]. There will be separate modules for Hand, wrists, elbow, shoulder. joint, spacer and cosmetic components. With these a prosthetist will be able to build any prosthesis ranging from a hand to upper arm, in a very short time. The whole system is then programmed according to the needs and abilities of the specific user.

The modularity extends to the software as well so it will be possible to control the arm from a wide range of different control philosophies and input methods. These will include intelligent arm control strategies, pattern recognition of EMGs and appropriate feedback systems. The ease of final configuration of the arm will allow the

potential user to choose the methodology at the time of fitting and easily modified later.

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THE DEVELOPMENT OF AN ADVANCED MULTI-AXIS MYO-PROSTHESIS AND CONTROLLER

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ABSTRACT

Commercial myo-electrically controlled prostheses are currently single degree of freedom devices with highly restrictive function. These artificial hands warrant high grip forces due to their planar pincer movement to ensure stable prehension, thereby inherently requiring precise and conscious effort on the part of the wearer to ensure optimum grip.

The Southampton Hand has demonstrated the ability to devolve low-level user control to the hand prosthesis itself by the use of the Southampton Adaptive Manipulation Scheme [1]. Until recently these multiple-axis prostheses have lacked clinical significance due to poor reliability and user-oriented design [2].

The development of the latest device is centred on the hypothesis of enhancing stable prehension by increasing the adaptability of the prosthesis, whilst simultaneously minimising the necessary grip force. This is to be achieved by increasing the number of independent degrees of freedom of the device without compromising user-effort by utilising the Southampton hierarchical control system. Constraints such as modularity, low weight and power consumption are factors that have been adhered to throughout the design process.

The six independent axes of the hand are controlled by a single microprocessor. The limiting factor in the advancement of artificial hands has frequently been the integration of technology to the device. Consequently several accurate sensing systems were implemented in this design to enable a more comprehensive control of the adaptable hand prosthesis.

INTRODUCTION

Conventional hand prostheses utilise proportional control methods for the user to directly control the position (or velocity) of a single degree of freedom hand via their EMG signals. The inherent disadvantage of these systems is the conscious effort a user must make to ensure the stable and optimum grasping of an object. This form of control frequently results in the user overgripping to ensure that the object does not slip from the hand. Utilising such a method for a multiple degree of freedom prostheses would result in an unacceptable control burden for the user.

The hierarchical control of the Southampton Adaptive Manipulation Scheme (SAMS) forms the foundation of an adaptive intelligent prosthesis philosophy [1], whereby the low-level control of the hand is transposed from the user to the device. The control scheme has been applied to several prototype prostheses (usually four degree of freedom devices), but has only recently seen clinical fitment in the form of a two axis device [3], with separate movement of the thumb and fingers. Use of a microprocessor ensures stable prehension by measuring force, slip, and digit position from the prosthesis.

This paper outlines the development of the latest Southampton hand (funded by the Rehabilitation and Medical Research Trust, Remedi).

A new mechanical prosthesis has been developed along with the SAMS controller implemented on a digital signal processor (provided by Texas Instruments under the Elite Universities Program) in combination with the

UNB myoelectric controller. When put into effect with multiple sensing systems, this design is to result in a highly adaptable and functional prosthesis without additional burden to the user.

MECHANICAL DESIGN

Conventional hand prostheses have limited object stability during prehension. Hence it is hypothesised that an increase in the number of degrees of freedom will improve the adaptability and grip stability of the artificial hand whilst minimising the necessary grip force.

The index and middle fingers of the natural hand are used to oppose the thumb during precision grips, whilst the ring and little finger are vital in strengthening prehension [4]. The lack of these functions is reflected in commercial hand prostheses by a notable absence of stability during power grips, usually only compensated for by the skill and experience of the user. Consequently the number of degrees of freedom in the Southampton-Remedi hand have been increased from the single axis found in a conventional device, to six independent movements, exhibited in the form of four fingers, and a dual axis thumb (to ensure maximal dextrous range during precision grip tasks).

The design of any hand prosthesis must adhere to a number of constraints, and factors such as cosmesis, reliability, modularity, weight and power consumption, must be optimised. These elements are reflected in the design of the Southampton-Remedi hand.

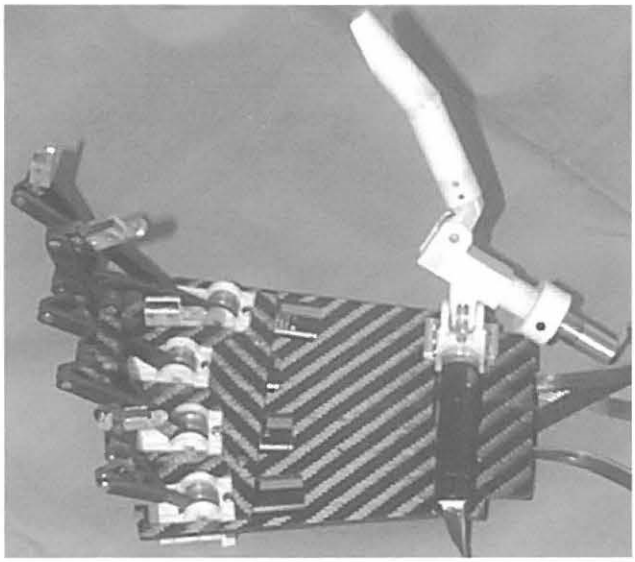

Figure 1 - Prototype Southampton-Remedi Hand

Mechanical linkages have frequently been used for the implementation of artificial hand digits [5,6] in order to produce a 'natural' curling action. A notable exception to the conventional linkage configuration is a 6-bar planar design [7], optimised for mechanical efficiency and trajectory (to reflect that of a naturally curling finger). Mathematical modelling was employed, and prosthesis design constraints imposed, to modify this design for use in the prototype hand. The result is a modular linkage made from carbon fibre (see Figure 1), driven via a motor and worm-wheel system to ensure a passive grip can be maintained without the constant application of power to the motor drive system.

The natural hand's dexterity is centred on the ability of the thumb to oppose the fingers. Indeed it is clinical consensus that a loss of thumb function causes a minimum of 50% of the hand's subsequent disability [8]. Given the role of the thumb as an opposer in the vast majority of grip scenarios, the single axis device currently found in commercial prostheses must severely impinge on functionality. A biomechanical study of the natural thumb revealed an optimum solution of a 2-axis system within the prosthesis. The circumduction and flexion axes in the artificial digit (see Figure 1) illustrated a compromise of the five degrees of freedom exhibited in the natural thumb whilst maintaining a broad range of dextrous opposition. The design exhibits a similar modularity to that of the artificial digits, with the dual axes being powered by motor and worm-wheel drive systems.

These modular components are integrated to the prosthesis through the carbon fibre palm. The biomechanical arches displayed in the palm of the natural hand form a template for adaptation during prehension, whilst the soft tissue and muscle groups enable sufficient compliance for the hand to actually effect an adaptive grip. Unfortunately it is not possible to replicate these palmar arches in the hand prostheses without advanced 3D modelling software

to study the effect of digit interaction. However a limited form of oblique flexion (necessary for the thumb to oppose the little finger), and digit opposition, has been achieved by the spanning of the fingers (along with the mobility afforded by the dual axis thumb).

CONTROLLER DESIGN

The SAMS hierarchical state control scheme requires only simple user inputs from the forearm flexor or extensor muscles (see Figure 2). Extensor tension causes the controller to initiate the proportional opening of the hand, or if the exertion is sufficient, to trigger a RELEASE state. The prosthesis operates on a voluntary-opening, involuntary-closing basis; the hand will automatically close when the subject is relaxed with minimal myo-activity. Each digit of the hand will continue to close until encountering an object, whereupon each individual drive is powered down, and the controller has achieved a TOUCH state. This results in the prosthesis exerting the smallest feasible touch force on the object whilst maintaining maximum surface area contact. To initiate a grasp, flexor tension causes the controller to implement a HOLD state. This mode utilises slip sensor feedback to maintain optimum grip force whilst arresting any instability that may occur when the user manipulates the object. This state can be overridden by a further flexor tension that causes a SQUEEZE state whereby current sensor feedback enables proportional force control of the prosthesis by the user.

Multiple-axis hand prostheses possess the capability of initiating various prehensile patterns (although obviously limited by the number of degrees of freedom within the device). The SAMS state control system is designed to maintain optimum grasp rather than to govern the implementation and discrimination of grip scenarios. Consequently, supplementary sensors on the hand have been used previously [6] to effectuate multiple grip configurations; for example, a force sensitive resistor on the side of the index finger would be activated by the user to initiate a lateral (or 'key' grip). This method of explicitly triggering prehensile patterns is inconvenient. However, a direct implementation of grip type from EMG signals would prove extremely difficult, if not impossible, for the user to achieve due to the limited number of degrees of freedom obtainable from the myoelectric signals.

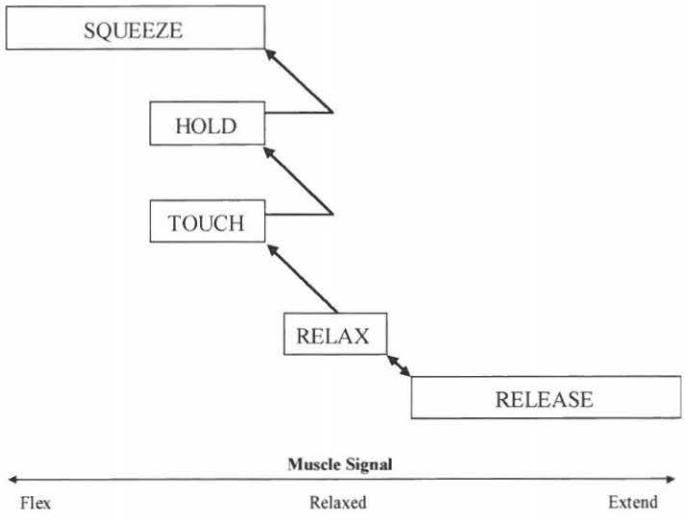

Figure 2 - SAMS Control Structure

Despite this, a unique method of control may be achieved by the combination of the SAMS system and the UNB myoelectric controller [9]. The UNB controller was originally designed for the control of multiple axis upper limb prostheses (such as powered elbow, wrist and hand devices). The controller is capable of determining three independent degrees of freedom from the user's myoelectric signals, and subsequently produces three state outputs and a proportional flexion/extension amplitude signal (see Figure 3).

The proposed control system for the Southampton-Remedi hand is to afford the user maximum versatility. The system essentially consists of two separate controllers implemented on two individual microprocessors. The UNB myoelectric controller is used to determine prehensile pattern (either lateral, power or precision) by the selection of an appropriate degree of freedom, and once initiated, the flex/extend signal is used then to drive the SAMS state control in the conventional manner. The combined control scheme provides the user with ability to directly implement prehensile pattern control with automatic adaptive manipulation (Figure 3).

MICROCONTROLLER, SENSORS AND DRIVE SYSTEMS

In order to keep power consumption to a minimum and maximise motor control, H-bridges are used to drive each motor (see Figure 4). The circuit uses pulse width modulation (PWM) with a frequency of 16kHz, and forward/reverse logic signals to control motor terminal voltage and direction.

Until recently it was not possible to integrate accurate sensing systems within prostheses due to the technological limitations and excessive cost. However the Southampton-Remedi hand has incorporated high-accuracy sensing systems that are capable of providing precise information on position, slip, and motor current (to determine fingertip force and monitor motor performance).

The digital signal processor (DSP) used is optimised for digital motor control and manages all of the input/output (I/O) to the sensors with a minimal requirement for additional hardware. Although designed for governing single brushless dc motors, the prosthesis controller maximises the I/O potential of the DSP by governing the multiple-axis brushed dc motors of the hand.

A small number of hardware interface circuits between raw sensor signals and the main controller are necessary. Digital magnetic encoders mounted to the motors generate digit position information. This quadrature signal is decoded and input to the DSP in the form of a 16bit count, ranging from zero at full extension to 0x2748 at full flexion. This provides a resolution of approximately 120bits per degree of digit rotation (thereby assuring far greater control accuracy than is actually achievable given the non-linearity of the mechanical system).

Supplementary hardware also includes interfacing between the acoustic slip sensors [2], the motor current sensors, and the controller. The analogue current signals are filtered through a low-pass 500Hz Bessel filter to reduce the noise interference generated by the switching of the H-bridge power electronics (Figure 4).

CONCLUSION

The Southampton-Remedi hand is a novel six axis hand prosthesis capable of stable prehension through the use of multiple independent digits. The mechanical design has been based on biomechanical studies of the natural hand.

The control of multiple degree of freedom devices can prove an extreme burden to the user. The Southampton Adaptive Manipulation Scheme has been illustrated to alleviate this burden by transferring low-level grasping control from the wearer to the hand itself. In order to maximise the potential of various grip scenarios, the UNB myoelectric controller is to be used to initiate a lateral, power or precision grip prior to triggering the SAMS control system.

Multiple, accurate sensing systems are an integral and essential component within the hand to achieve autonomous grasping control, by providing information on digit position, motor current and object slip.

Although the Southampton-Remedi hand is not expected to undergo clinical trials at this stage, the functional range of the device is to be assessed using the Southampton Hand Assessment Procedure [10]. The abstract tasks within this evaluation procedure are designed specifically to assess various prehensile patterns, and thereby provide initial assessment criteria for examining the functional range of the device.

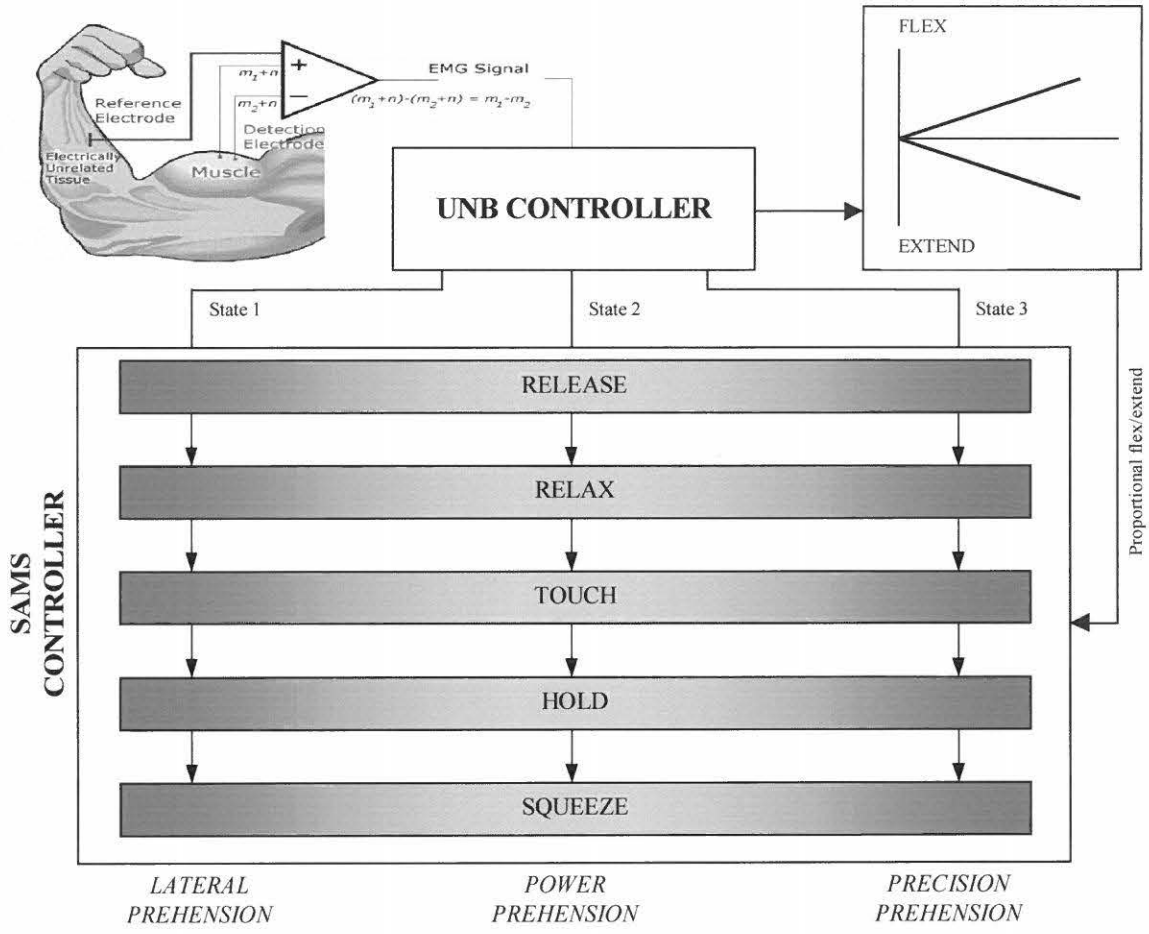

Figure 3 - Direct Prehensile Pattern and Adaptive Manipulation Control with the UNB/SAMS Controllers

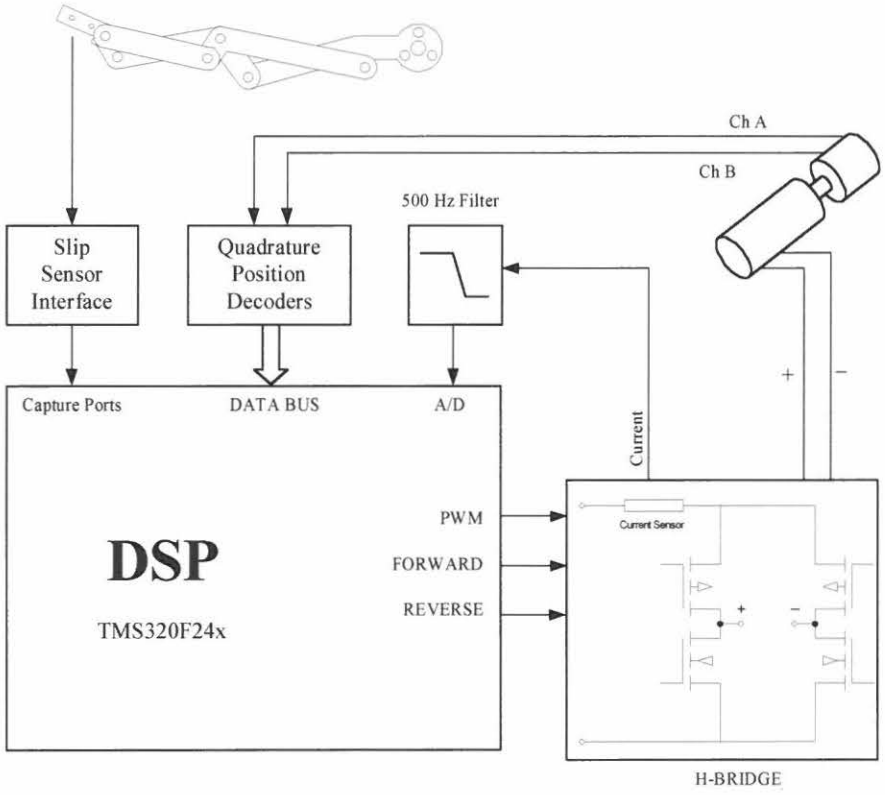

Figure 4 - Overall Input/Output System Control

ACKNOWLEDGEMENTS

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TELEASSISTANCE FOR ELECTRONIC COMPONENTS IN REHABILITATION TECHNOLOGY

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INTRODUCTION:

The synergisms of control and telecommunication systems have paved the road towards increasingly interesting fields such as teleassistance and telemedicine, and one other interesting application of these technologies may be in the field of prostheses. Nowadays, we are used to high-tech prostheses whose sensors or microprocessors, on one hand, enhance the performance of the prosthesis itself, while, on the other, can also cause problems which require the intervention of a specialised technician. The numerous benefits provided by telecommunication for prosthesis wearers are obvious for those patients who would otherwise have big problems in getting to the specialised centers, even just for a simple check-up. It is also important to consider the savings in money and time, since many patients bear not only travelling expenses, but are sometimes required to stay for days at a time at the specialised centers. The ideal solution would be to carry out as many telematic operations as possible through audio-video-data connection, which entails a much lower cost and much less inconvenience, even when considering a trip of only a few kilometres. This project originates from the necessity of INAIL, which has numerous centers spread out throughout the entire Italian territory, to concentrate specific technical services in a limited number of centers to which reference is made for each clinical case. By going to the closest INAIL Center, equipped with video-communication apparatus, the patient will be able to carry out operations that he normally would have had to carry out at the Prosthesis Center, ranging from the periodic check-up of the condition of the prosthesis, such as: the status of the batteries, the condition of the cosmetic sleeve, a control of the sensors and their calibration, to more complex cases of calibration or modifications in the function of the prosthesis itself, by means of the variation of specific parameters. Even the analysis of the electromyographic signal can be carried out at a distance, thus allowing the technical-rehabilitative crew to evaluate various solutions before asking the patient to return to the Prosthesis Center.

A fundamental partner in this activity is the OttoBock company, who has fully participated in all the objectives of the project for the purpose of providing both the patient and the orthopaedic technician all the instruments necessary in order to allow the patient to have increasingly advanced and reliable devices and the technician to carry out his job more quickly and efficiently. Various solutions were examined during the project, all of which were fully workable with different levels of quality and costs.

The only choice made and maintained from the initial phase of the project was that of utilising point to point connections though digital ISDN telephonic lines. In any case, this choice is not binding, although it is highly advisable, particularly in view of the fact that this service is available in almost every city, at increasingly economic prices. The reliability of this network in comparison to the telephone network and the availability of a larger band of transmission, allows for an efficient transmission of not only the audio signal, but also the video signal and a data channel.

The operations of calibration and monitoring made possible locally, by connecting the prosthesis with a PC, can be carried out between two local and remote locations that communicate through the ISDN network.

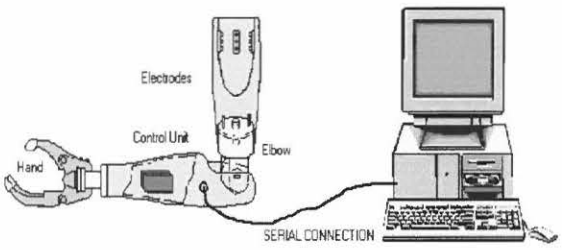

Fig.1a Control software for prostheses with microprocessor

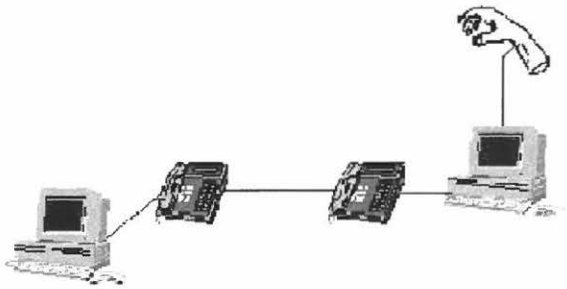

Fig.1b Connection diagram between local and remote locations

POSSIBLE SOLUTIONS FOR VIDEOCOMMUNICATION

The analysis of the problem and the evaluations carried out during its development, have allowed for identifying two groups of solutions which will be examined further ahead.

In any case, problems exist that are common to all the solutions, particularly in relation to the temporal synchronisation of the events between the local and remote locations and the delays caused by the width of the video signal band to be transmitted. In addition, in view of the need to communicate with the prosthesis, a synchronism is necessary between the images: for example, the hand that moves with the data that are transmitted or received during that action. This is the only way a diagnosis of the prosthesis itself can be carried out.

The elimination of the video surely simplifies the problem, but in the thirty-year experience of the Prosthesis Center, human contact between the technician and the patient has definitely proven to be one of the main prerogatives for a correct prosthetic treatment and the possibility of seeing each other favours this human approach.

The two groups of solutions identified for the problem are illustrated in fig. 2.

The first choice that has to be made for videocommunication is between the "Consumer" type systems, with the characteristic of low prices and typically medium-low performance and the "professional" type systems, characterised by a better performance and considerably higher costs (in our case, videocon Aethra kits were utilised).

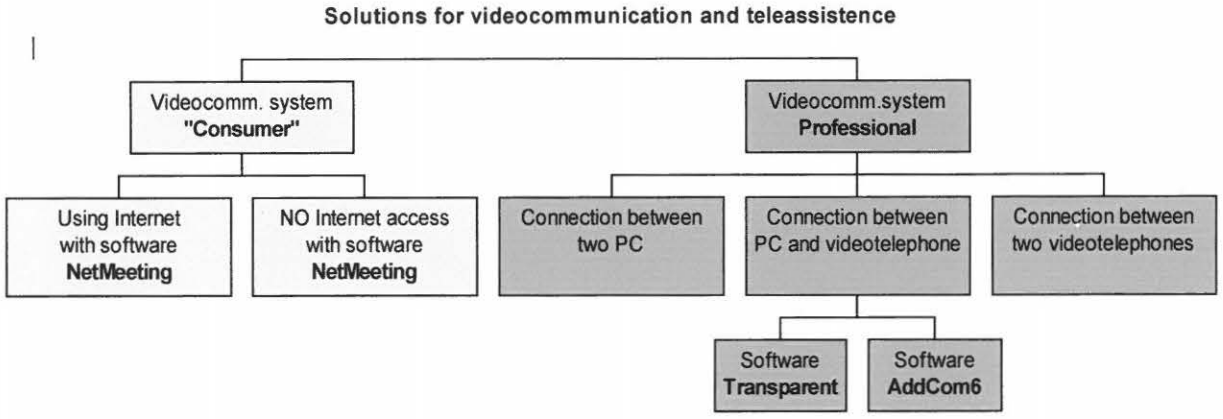

Fig. 2 Analysed solutions for video teleassistance

CONSUMER SOLUTIONS BY MEANS OF APPLICATION SHARING

For this group of solutions, the concept of Application Sharing, which consists in the utilisation of an instrument that allows the operator to act and manipulate on the prosthesis control Software at a distance, takes on fundamental importance. Substantially, the patient who goes to a peripheral center connects his own prosthesis to a PC containing the control/monitoring software, which is remotely shared with another PC on which the expert technician, who is capable of correctly interpreting the meaning of the gathered data, is operating.

The expert operator, from the main center, shares this application just as though it were directly present in his own terminal.

Different types of software are presently available that have similar features and in this study, it was decided to use NetMeeting Microsoft, also in view of the fact that it is distributed without charge.

NetMeeting is a *Microsoft* program that is very easy to use for video communication which allows connection through the local network (LAN), via Internet or by directly utilising a modem. If the connection is made through LAN or Internet, the calls can be carried out by specifying the name of the network or the TCP/IP address of a specific computer. If a connection is made to a computer that carries out a USL (User Location Server) via Internet, it is also possible to call and specify the name of a User in the file managed by the User Location Server. Once the call is made, NetMeeting allows both to dialogue by means of written messages (even with more than one User at a time) and to share the applications, which is the essential characteristic of the teleassistance project. Another important feature that this software offers is the audio-video remote connection with the User, according to current standards.

This solution requires that the software relative to prosthesis management be distributed to all the peripheral locations and run locally from the PC, with the patient's prosthesis connected by means of serial gate. The operator, from the main center, can share this application just as though it were available on his own terminal.

Therefore, on both sides, the system requires NetMeeting software and a videocommunication kit[1] and, on the user's side, the addition of the application package relative to the prosthesis worn by the patient

The advantages of this solution are fundamentally linked to its low cost, since both the required hardware and software are products that are easily found at low prices in any computer store. The above solution may foresee, as a further development, prosthesis monitoring directly from the home of the patient and no longer in the centers equipped with sophisticated (and costly) professional systems.

The disadvantages of this solution, on the other hand, are due to extremely long delays in results caused by the lack of hardware codifiers and de-codifiers for video and audio in these video communication kits. With the solution that foresees the connection via Internet, it is also important to keep in mind that the TCP/IP protocol does not guarantee a fixed delay in the transmission of the packages, first of all because they aren't transmitted in a group and therefore arrive at destination with very different speeds and secondly because the delays depend on the amount of traffic on the network. Another limit regards the quality of the picture: such devices guarantee a quality that is sufficient for seeing a person, but certainly not sufficient for examining details. The suggested window format for these devices is 160 x 120; when increasing in size, the detail of the image worsens. Although personal computers are used daily by an increasing number of people and their easy utilisation makes them accessible to people who would have never used them up until a few years ago, there is still a category of elderly people that finds that this instrument makes everything more complicated.

Furthermore, the enormous flexibility representing one of the strong points of the PC, may also turn out to be the weak point of the solution in the case one desires to use the computer for other tasks, from which the unavoidable problem of software viruses or instability could arise.

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[1] For testing, the multi-media Digicom Galileo kit was utilised, consisting in: video camera, headphones and microphone, integrated with a SoundBlaster 32W card

PROFESSIONAL VIDEOCOMMUNICATION SYSTEMS

The videocommunication apparatus utilised in the tests are systems manufactured by the AETHRA company in Ancona (Italy), a European and world-wide Leader as far as systems of videocommunication and teleconferences are concerned. Solutions are possible for various operative requirements (Fig.3a, 3b, 3c) which utilise from the simple but practical video telephone for instalment at the patient's home, to the more sophisticated systems of videoconference, for use by businesses.

Fig.3a Videocom system PC based.

Fig.3b High quality video conf.system

Fig.3c Videotelephone

In this analysis, the PC Based systems will be considered, i.e. those consisting in a PC with a HW dedicated to systems based on videotelephones.

By adopting these types of solutions, one may benefit from a complete system consisting in hardware and software the allow for management of remote connection via ISDN network. Typically, these kinds of systems include a PC card that encodes and decodes the audio video signal, also permitting the utilisation of professional monitors and video cameras for those cases where high quality is required. The systems are capable of utilising 1 to 3 digital lines, associated in such a way as to have a useful transmission band up to 384Kbit/sec. One important characteristic of the system is the availability of a *remote serial,* called *Com6,* which allows to receive and transmit the data without having the serial gate physically present on the PC, although present on the remote video communication apparatus to which one is connected. Thanks to a low-level software interface, the data transiting on a physical serial (com1 or com2) are transferred to the data channel of the ISDN network and, vice-versa, the ones arriving from the network are directed to the PC bus as though they were being received from a physical serial. In a totally transparent manner, the expert operator who has the task of tuning up the prosthesis, selects a remote serial (com6) by means of the options provided by the prosthesis management software and elaborates the signals of the remote patient's prosthesis just as though they were directly connected to a serial gate of his PC. For this additional serial gate (defined as a remote or virtual serial) provided by the software application, the characteristics of the communication protocol, such as the speed of transmission, the start and stop bits, the equality bits, etc., may be fixed.

This type of solution avoids the network transportation of video maps which characterise Application Sharing, thus limiting the flow of data from and to the prosthesis, which, in the worst of cases, is 19200, while obtaining a distinct improvement in the speed and quality of the image and a drastic decrease in the delays between the audio visual reception and the data relative to the prosthesis.

Two PCs are present in this solution: one at the patient's home and one used by the technician who manages the video communication and the data transfer and elaboration.

The software layer (Caronte.vxd) delegated to detecting the data of the physical serial connected to the prosthesis for transfer on the ISDN network, and the relative PC with the video communication apparatus at the

patient's home, may be replaced by a special video telephone which is entirely and automatically occupied with the management of the data channel, with the same features of the Caronte application. In this case, the apparatus found at the patient's home consists in a video telephone that can also be used as a normal telephone and which is no doubt more useful for people that aren't familiar with computer systems. The only flaw in this solution is the high cost of these devices when compared to a normal telephone.

In the hypothesis of the use of a PC and a videotelephone, two alternative software solutions have been developed for the purpose of making the system more flexible. In fact, with the previous structure (Fig. 4a), a software layer exists that provides access to the com6 serial (Addcom6) at an operative system level, although, in turn, the prosthesis software application must foresee a com6 serial gate in its configuration options in order to be able to activate the teleassistance. This forces the producers of prostheses to have to modify the existing software, already distributed. The reason for this is that the PC managing the videocommunication is the same one working with the prosthesis software. One solution to the problem (Fig. 4b) may be to separate the two functions of prosthesis management and videoconference/data transmission, by leaving an application called "Transparent" on the PC delegated to the videocommunication carrying out the function of recopying the data arriving from the com6 virtual serial to a physical serial of the same PC: com1 or com2. At this point, the physical serial can be connected to the physical serial of the PC in which the prosthesis management software is installed.

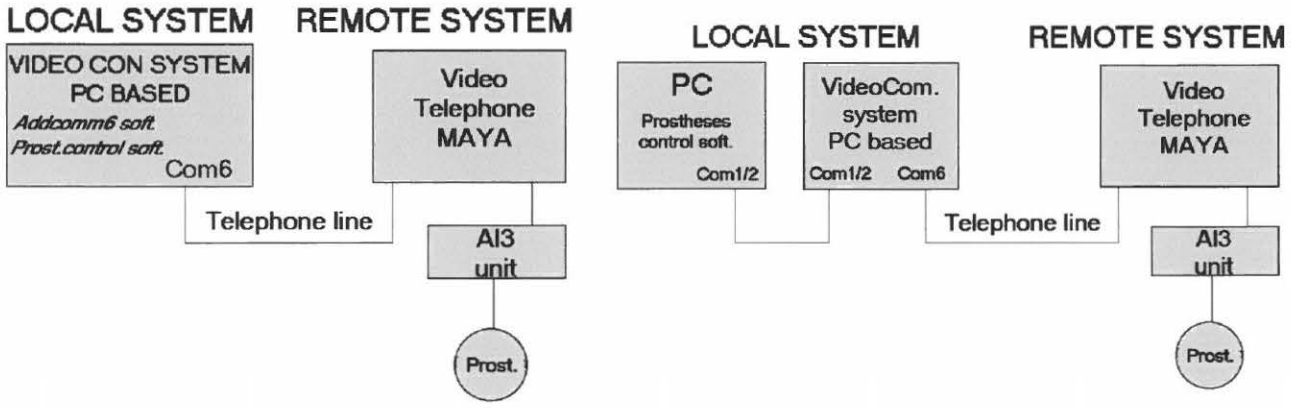

Fig.4a Solution with AddComm6 software

Fig. 4b Solution with Trasparent software

As described in the following Table, by using a second PC, the same software can be utilised without any modifications.

Remote Connection Application	Local Hardward Required	Necessary Software Modifications
Transparent	**2** PCs with Pentium 133 Mhz or more, one of which equipped with Aethra kit.	**None**
AddCom6	**1** PC with Pentium 133 Mhz or more, equipped with Aethra kit.	An additional serial gate is necessary: *Com6*

As a last solution, one could foresee the connection of two videotelephones that carry out all the functions of videocommunication and opening of the data channels for the virtual serial. With such a solution, however, the

problem remains that no intervention on the communication protocol is allowed and therefore it is impossible to carry out an optimal synchronisation of the video with the data.

RESULTS AND FUTURE DEVELOPMENTS

The various teleassistance systems based on professional equipment or on typically "low cost" equipment, with different solutions, have surely proven to be capable of carrying out an important role in the remote management of myoelectric prostheses.

The following Table lists the essential information concerning the various solutions, including economical aspects. Each solution requires the identification of certain hardware and software requirements for the local system and remote system and emphasise should be made that one absolutely preferable solution does not exist: in fact, the choice is based on a compromise between performance, costs and the typology of the patient. For example, in the case a choice must be made for an elderly patient, at home, who is not at all interested in purchasing a PC, the solution with the Maya telephone is surely the best, even though the costs are still fairly high.

ADOPTED SOLUTION	COST	NOTES
Low cost PC-based system	LOW	Ideal for patients having good familiarity with computers.
Professional system with HW PC (SDV8000) 128 KB	MEDIUM	Ideal for Assistance Centers
Professional system with MAYA 128KB videotelephone	HIGH	Ideal for elderly patients or other people with very little familiarity with computers.

The tests were carried out by connecting both an Inail center with the Prosthesis Center and the Prosthesis Center with the OttoBock Austria headquarters in Vienna, Austria. During these experiments, a test of the sensors installed on the SensorHand and their calibration was carried out. Verification was also made of the possibility of remotely diagnosing malfunctions due to the batteries or to a bad electrode contact. Another very interesting test, giving truly excellent results, was relative to an electromyographic examination of the patient at a distance.

One other important field in which this system could be extended is that of electronic wheelchairs, which may undergo functionality tests by means of special dedicated switchboards or software that can be personalised as far as maximum speed and acceleration, etc. are concerned. In this case too, teleassistance will make it possible to solve some of the wheelchair problems directly at the patient's home and offer him a series of additional services to make its utilisation and management more simple. Just think how much it would cost to ship an entire wheelchair in comparison to replacing only the broken part that has been identified by means of remote connection.

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FOREQUARTER PROSTHESIS WITH INTERCHANGEABLE ELBOW, FOREARM, AND HAND

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ABSTRACT

Fitting of individuals with forequarter amputations is often a challenging and fruitless effort. In order to provide optimum function for the patient, it is necessary to fit these individuals with either externally powered prostheses or hybrid prosthetic designs. These prostheses are oftentimes heavy and bulky and are not accepted by the patients for full time wear. There exists another population of individuals with forequarter amputations that are more interested in the cosmetic restoration of the prosthesis rather than its functional capabilities. Passive prostheses are typically fabricated for these patients in order to fulfill their particular desire. In an effort to increase acceptance and wearing time, the prosthetic design being discussed in this paper is an attempt to meet both criteria: provide function and acceptable cosmetics. By utilizing existing components, this prosthetic design enables the user to readily switch from a heavier, externally powered prosthesis to a lightweight, passive prosthesis

INTRODUCTION

"Conventional" prostheses for individuals with forequarter amputations have consisted primarily of two designs. The passive prosthesis is typically the lightweight cosmetic alternative, although it has many merits beyond just "filling the shirt sleeve." Most forequarter passive prostheses have a universal shoulder joint or a combination of bi-axial shoulder joint and internal rotation and external rotation available at the elbow turntable to simulate the tri-planar joint motions. The elbows can be either free-swing or friction, with or without a locking feature. The most crucial part of these passive prostheses is the terminal device (TD). The TD is almost always a lightweight passive hand with formable fingers. These hands can be covered with a relatively inexpensive glove which provides fair cosmetic results, can be covered with custom silicone gloves which offer excellent cosmetic restoration, or something in between these ends of the "glove spectrum."

It is possible to fabricate and fit completely cable driven or hybrid prostheses to individuals with intrascapulothoracic amputations. The difficulty, however, arises from the amount of excursion that the user can generate. Most cable driven elbows require 2½" to 3" of excursion to flex through the entire range of 125°, while voluntary opening and voluntary closing TD's require approximately 1¾" and 2¼" of excursion respectively. The addition of excursion amplifiers such as the APRL Sheave™[1] or a simple class three lever system are often utilized to decrease the amount of excursion that the user must generate. This, however, is at the expense of increased force required to operate the prosthesis. Although this is a viable option for completely cable driven prostheses, it is often difficult to find an appropriate placement for the excursion amplifier. Exoskeletal prostheses, covering a large amount of the body had been historically used for this design. These designs enabled the prosthetist to place the excursion amplifier in a variety of locations. The problem with this is that these types of prostheses are heavy, uncomfortable for the wearer, and cover a large surface area of the body. With the forequarter amputation itself, the body already has less surface area, thus there is a decrease in the body's ability to dissipate heat. This massive covering further complicates matters by adding to the heat retention. When the socket is fenestrated, i.e. Sauter frames, the wearer is more comfortable because the prosthesis is lighter in weight and cooler; however,

placement of the excursion amplifier is somewhat limited. The hybrid prostheses help to eliminate the need for as much excursion as the completely cable driven design requires. Several prosthetists have had good success with this type of design incorporating a cable driven elbow and externally powered TD or an electric elbow and a cable driven TD.

The other obvious option is to fabricate a prosthesis that is entirely externally powered. The Utah Artificial Arm™[2] and the Boston Elbow™[3] are "systems" which easily allow for this combination. Either of these systems can be utilized for a forequarter prosthesis very successfully. The difficulty with them is that they are heavier than the cable driven or hybrid designs, and much heavier than the passive designs. This issue of added weight contributes to the wearer's limited use of the prosthesis and in some instances, an ultimate rejection of the entire prosthesis.

DISCUSSION

The design that we are using is intended to increase the patient's acceptance and utilization of the prosthesis. This is being attempted by allowing the user to easily interchange a passive prosthetic design with an externally powered prosthetic design with the "push of a button" and plugging in of two wires. These two designs are incorporated into a single prosthesis which maintains a consistency with the feel of the socket as the user changes from passive to externally powered components and vice versa. The advantages of each system have been previously described, and we feel that these will contribute to the wearer's increased acceptance of the device.

The components of the prosthesis are many and can be seen in figure 1. Beginning proximally, is: a Servopro and wire harness™[2], an endoskeletal model, shoulder flexion abduction joint™[1], pylon tube kit™[1], passive wrist™[1], knurled plate™[4], and quick disconnect insert™[1]. These components are germane to both the passive and externally powered prosthesis. Below the quick disconnect insert for the passive components are: quick change wrist™[1] laminated to a 2" diameter, UCLA CAPP Delrin Body Friction Wrist™[1], passive CAPP elbow, ™[5], threaded cpvc tubing, knurled plate™[4], passive hand and glove™[4]. For the externally powered prosthesis the components below the quick disconnect insert are: quick change wrist™[1] laminated to the lamination collar™[2], Utah Artificial Arm™[2], system electric hand and glove™[4].

The selection of these components is designed to enable the user to easily convert his or her prosthesis from a passive prosthesis to an externally powered prosthesis. First, the user is required to unplug two pre-amp cables™[2] coming from the Servopro controller™[2] to their corresponding connectors on the Utah Artificial Arm. In addition to their pin connection pattern, the cables have been color coded to prevent inadvertent attempts to plug them together inappropriately. Secondly, the wearer pushes the button on the quick change wrist, in the same manner as it is designed for terminal devices, which will release the electric elbow and terminal device. The passive prosthesis is then installed by pushing the insert into the quick change wrist that is on the proximal end of the "passive components." Obviously, the reverse order would hold true for interchanging the passive components back to the electric components.

RESULTS AND CONCLUSION

At present, it is too early for us to tell whether or not this design will prove beneficial in the quest for acceptance of forequarter prostheses. The patient using this prosthesis still has it in the temporary stage but has demonstrated the ability to easily switch from the electric design to the passive design. It will be some time before we can determine whether or not she will accept the prosthesis. Also, each individual's ability to tolerate forequarter prostheses differs. This design, however, offers the ability for people requiring these types of prostheses to have both passive and electric designs in one. This factor, in addition to the weight difference of approximately 2½

pounds between the two sets of components should prove to increase the user's acceptance and wearing time of the prosthesis. This, or similar designs may also prove beneficial for people with transhumeral amputations to have different designs within one prosthesis, provided their residual limbs are short enough to allow for the components.

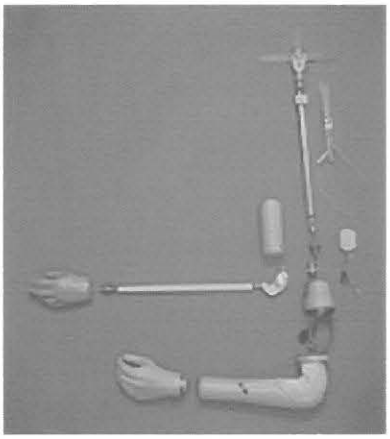

Figure 1

1. Hosmer Dorrance Corporation, 561 Division Street, PO Box 37, Campbell, CA 95008
2. Motion Control, Inc., 2401 South 1070 West, Suite B, Salt Lake City, UT 84119
3. Liberty Technology, Prosthetics and Orthotics Group, 71 Frankland Road, Hopkinton, MA 01748
4. Otto Bock Orthopedic Industry Inc., USA, 3000 Xenium Lane North, Minneapolis, MN 55441
5. United States Manufacturing Company, 180 North San Gabriel Boulevard, Pasadena, CA 91117

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

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ABSTRACT

The typical prosthetic wrist has a limited range of motion. Although a rotating device may be used for pronation/supination, flexion/extension and radial/ulnar deviation generally are set in a fixed alignment by the prosthetist. The prosthesis wearer compensates for the lack of wrist motion with abnormally large motions of the shoulder and elbow. This may give an awkward appearance, and also increase the risk of joint injury. The goal of the prosthetist is to choose an alignment that reduces these compensating motions. Unfortunately, people disagree as to what the optimal alignment is. The goal of this study is to quantify the resulting arm motions for various alignments, and determine which alignments allow near-normal motion.

This study examined ten subjects performing three activities of daily living (ADLs). The ADLs were: 1) drinking from a cup, 2) eating with a spoon, and 3) eating a sandwich. The subjects were all normally-limbed, so hand and wrist splints were worn to imitate a powered prosthesis. Five different alignments were tested using five wrist splint settings (0-10 degrees) and "normal" condition was tested with no wrist splint worn. Motions were recorded using a VICON™ 140 motion analysis system. The height to which the subject raised his/her elbow was used as a measurement of shoulder range of motion. Alignments were considered "acceptable" if they resulted in the elbow being lifted less than or only slightly greater than "normal".

FINAL PAPER NOT RECEIVED AT PRESS TIME

Magnetic Resonance Imaging of Congenitally Deficient Upper Limbs

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Introduction

Knowledge of the anatomic location, size, and contractility of muscles within a person's congenitally deficient upper limb is useful in prescribing and fitting a myoelectric prosthesis. In 1998, at University of Texas' Hermann Hospital, magnetic resonance imaging (MRI) was used to image both arms of five volunteers with congenital unilateral below elbow upper limb deficiences. Imaging both arms of each subject enables a direct comparison of normal and residual limb anatomy. The volunteers included one adult and four teenagers (Table 1). This paper summarizes findings on residual versus sound side musculature size, limb size, and selected residual limb features.

Table 1: Five volunteers with **congenital** upper limb below elbow limb deficiencies participated in this study.

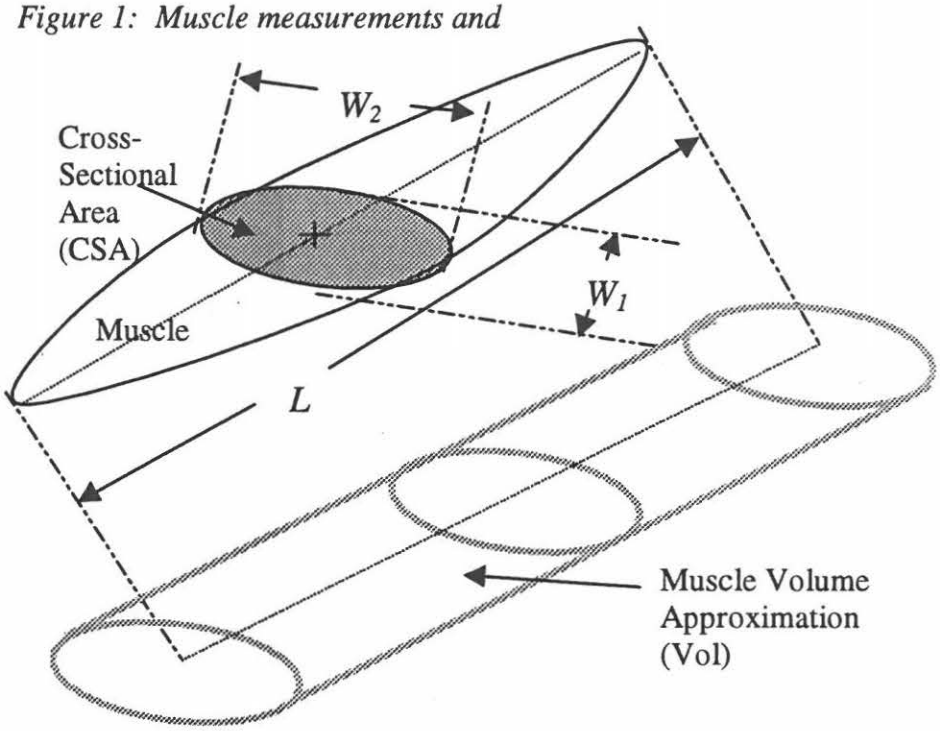
Study ID	Sex	Age	Residual Side	Residual Length	Comments
A	Female	32	Right	6.0 cm	Regular body-powered hook user since early childhood. Occasional body-powered hand use.
B	Male	12	Left	18.4 cm	No prosthesis use since age 6. Cannot fully straighten elbow of residual limb. Some carpal bones present. Residual forearm radius and ulna fused in pronated position. Sound and residual side humerus bones not equal in length.
C	Female	12	Right	7.0 cm	Moderate myoelectric user, hand only. Mother's ultrasound at 14-16 weeks of pregnancy showed two hands developing normally.
D	Male	13	Right	9.6 cm	Moderate myoelectric user, hand only.
E	Male	12	Left	10.4 cm	Occasional myoelectric user, hand only.

Technique

For each volunteer, the residual limbs were imaged from about 6 cm proximal of the elbow to the limb's distal end. Axial (transverse), coronal, and sagittal sections were imaged using an extremity coil on the first volunteer, subject A. Since the MRI process requires that the subject remain motionless for long periods in a small space through loud noises, the younger subjects' (B-E) imaging was limited to transverse sections, with other views synthesized. Sound limb images were limited to a length corresponding to their residual limb.

Working from the MRI films, University of Texas--Houston Health Science Center radiologists identified and measured the volunteers' forearm muscles. It was not possible to exactly measure muscle length for those muscles originating above the elbow joint or terminating beyond the field of view. This occurred on the sound limb only, where the volunteer's residual limb was short and only one extremity coil placement was done on both sides (subjects A, C-E). Muscles originating at the common flexor or extensor origin were measured from these origins. Muscles that terminate in a long tendon were measured only to the last visible muscle tissue. Cross-section dimensions (width of a minor axis, W_1, and width of a major axis, W_2) were measured at maximal cross-sectional mass, as Figure 1 shows. At this point,

Figure 1: Muscle measurements and



$$CSA = \pi(W_1/2)(W_2/2) \quad (1)$$

approximates the cross-sectional area (CSA) as an ellipse.

$$Vol = (CSA)(L) \quad (2)$$

approximates the muscle volume (Vol).

Maximum and minimum limb diameters and skin thicknesses were measured at a transverse plane including the radial head. Figure 2 shows these limb cross-sections

Results

Tables A-E (Appendix) show dimensions of each volunteers' residual musculature and their corresponding sound limb musculature. Table 2 summarizes all the volunteers.

For the most part, all muscles that could be identified and measured in the sound limb (within the length of the residual limb) could also be identified in the residual limb. The exception was *palmaris longus*. This muscle is missing in 10% of normal limbs [1], so its absence or presence in these volunteers' limb may not be related to their limb deficiencies. All muscles on the residual side were located approximately in their normal location with respect to other muscles. Muscles not originating in the common flexor or extensor tendons originated closer to the elbow in the residual limbs than in the sound limbs, suggesting shortened rather than truncated forearm bones. The residual limb muscles were considerably less developed than their sound limb counterparts. The average total residual limb muscle cross-sectional area was 15.7 cm^2, compared to 34.2 cm^2 (Table 7). Also interesting is the fact that the overall diameter of the residual limbs varies much less than that of the sound limbs--the standard deviation of the residual limbs' diameters is 0.26 cm versus 0.48 cm for the sound limbs.

The residual limb muscles were also separated from the skin surface by more electrically inactive tissue than their sound side counterparts (at a plane transversing the radial head, an average of 0.82 cm on the residual limbs versus 0.64 cm on the sound limbs). Toward the distal end of the residual limbs of subjects A and C-E, the thickness of the inactive tissue (apparently scar tissue as well as fat) increased greatly. This has implications for myoelectric prosthesis control, since this inactive tissue filters out the higher frequencies of the myoelectric signal [2]. Simulations by Farry [3] show that 0.8 cm of tissue attenuates a muscle action potential's amplitude by 80-90%. The action potential is almost undetectable through 1.5cm of tissue. The MRIs suggest that electrodes placed on the most distal 3-4 cm of the residual limb will not record useful myoelectric activity.

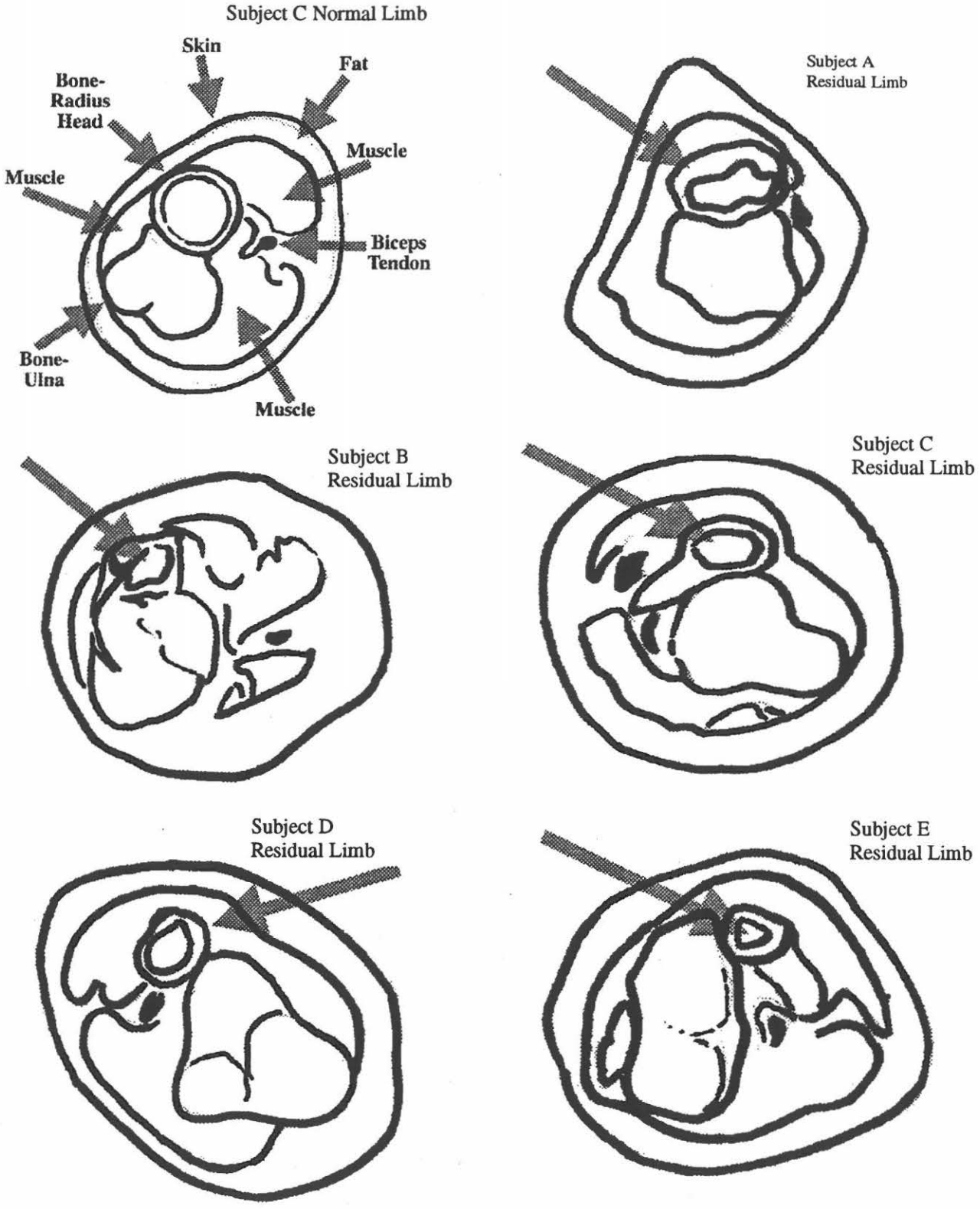
Table 2 also gives the correlation between these MRI measurements and preliminary results from a multifunction myoelectric control study, described in detail in [4,5]. The candidate controller discriminated between six motions--close grasp, open grasp, flex wrist, extend wrist, rotate palms up (supinate), and rotate palms down (pronate)--from four myoelectric signals collected from the residual limb during contra-lateral stimulation (i.e., the volunteers simultaneously moved their sound hand through these motions while imagining moving their missing hand). The best predictor of controller performance is the residual limb muscle volume. Interestingly, some sound limb characteristics correlate more highly with the controller performance than their corresponding residual limb characteristics.

The volunteers' elbow joints varied considerably from sound to residual side. Note, in particular, the dysplasia of the radial heads in all five residual limbs (Figure 2)--none are circular. All the study volunteers achieve the equivalent of pronation and supination of their forearms (with or without a prosthesis) using abduction and adduction of the shoulder and humeral rotation (depending on the degree of elbow flexion) rather than moving their proximal radio-ulnar joints. This suggests that the proximal radio-ulnar joint does not develop without use. In contrast, individuals with traumatic below elbow amputations retain some mobility in the proximal radio-ulnar joint, even though it is usually not useful in moving a prosthesis where less than 50% of the forearm remains.

Table 2: A summary of sound and residual limb statistics for the five subjects. No volume averages across subjects are given since sound side muscle lengths used in volume calculations are truncated by image field of view for some subjects.

Subject	Average Limb Diameter (cm) in Plane of Radial Head (cm)		Average Thickness of Tissue between Electrode and Muscle in Plane of Radial Head (cm)		Total Muscle Cross Sectional Area (cm^2) (Sum of Areas at Maximum Mass)		Total Muscle Volume (cm^3) (Sum of Muscle Areas at Maximum Mass Multiplied by Muscle Length)	
	Sound	**Residual**	**Sound**	**Residual**	**Sound**	**Residual**	**Sound**	**Residual**
A	6.1	5.2	0.51	0.67	39.6	14.1	562.9	50.0
B	7.1	5.9	0.81	0.89	47.6	15.5	717.9	120.2
C	5.8	5.4	0.64	0.89	17.8	9.4	112.4	31.9
D	6.2	5.6	0.58	0.97	34.7	17.3	341.4	67.5
E	6.3	5.6	0.65	0.69	31.6	22.3	227.5	77.3
Average	**6.26**	**5.54**	**0.64**	**0.82**	**34.3**	**15.7**		
Standard Deviation	**0.48**	**0.26**	**0.11**	**0.13**	**11.0**	**4.7**		
Correlations with Controller Accuracy	**0.91**	**0.81**	**0.63**	**0.00**	**0.79**	**0.70**	**0.61**	**0.97**

Figure 2: All volunteers had elbow abnormalities in their residual limbs. Upper left tracing is of a sound limb for comparison, transverse plane through the radial head. The unmarked arrows identify the radial head.



Conclusions

This five-person MRI study:

- quantifies muscle development and location in congenitally below-elbow deficient limbs;
- reveals extensive myoelectric signal attenuating scar tissue;
- suggests ways to predict success with a multifunction myoelectric controller; and
- shows bone abnormalities far from the end of the residual limbs.

Five subjects constitute a small sample and is a starting point rather than the basis for sweeping generalizations, but this sample suggests that congenitally deficient residual limbs are quite different from traumatically amputated residual limbs. This study raises many questions. Future work should investigate a routine role for MRI in prescribing prostheses for congenitally limb deficient individuals, especially its use in locating electrodes for myoelectric control.

Acknowledgements

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Appendix

Table A: Measurements of musculature in subject A. Residual forearm length from distal humerus is 6.0 cm. Muscle dimensions: L=length; W1=minor width; W2=major width; CSA=cross-sectional area; and Vol=estimated volume. All measurements are in centimeters (cm).

Muscle	**Sound Limb (Left)**					**Residual Limb (Right)**				
	L*	**W1**	**W2**	**CSA**	**Vol**	**L**	**W1**	**W2**	**CSA**	**Vol**
Extensor Carpi Ulnaris	18.7	1.2	1.3	1.2	22.9	3.2	0.5	1.0	0.4	1.3
Extensor Digitorum	18.3	1.5	2.0	2.4	43.1	3.6	0.3	2.1	0.5	1.8
Anconeus	7.8	0.8	1.6	1.0	7.8	3.2	0.8	1.3	0.8	2.6
Brachialis	3.0	1.2	3.6	3.4	10.2	1.6	0.7	2.7	1.5	2.4
Brachioradialis	11.4	1.4	2.0	2.2	25.1	3.6	0.5	2.2	0.9	3.1
Extensor Carpi Radialis	14.1	1.3	2.4	2.5	34.6	3.6	1.4	2.5	2.7	9.9
Pronator Teres	11.7	0.8	2.3	1.4	16.9	4.0	0.5	2.0	0.8	3.1
Supinator	9.3	0.8	2.9	1.8	16.9	3.6	0.6	2.3	1.1	3.9
Palmaris Longus	13.0	2.3	7.0	12.6	164.4	2.1	0.6	0.8	0.4	0.8
Flexor Carpi Radialis	>20	1.2	2.1	2.0	39.6	4.3	0.9	1.1	0.8	3.3
Flexor Carpi Ulnaris	>20	1.0	2.2	1.7	34.6	2.9	0.5	1.2	0.5	1.4
Flexor Digitorum Superficialis	>20	1.3	3.3	3.4	67.4	5.1	1.4	1.6	1.8	9.0
Flexor Digitorum Profundus	>20	2.2	2.3	4.0	79.5	3.6	1.2	2.2	2.1	7.5
Totals		**Sound Limb**		**39.6**	**562.9**		**Residual Limb**		**14.1**	**50.0**

* ">" indicates that the muscle continues beyond the imaged area.

Table B: Measurements of musculature in subject B. Residual forearm length from distal humerus is 18.4 cm. Muscle dimensions: L=length; W1=minor width; W2=major width; CSA=cross-sectional area; and Vol=estimated volume. All measurements are in centimeters (cm).

Muscle	**Sound Limb (Right)**					**Residual Limb (Left)**				
	L	**W1**	**W2**	**CSA**	**Vol**	**L**	**W1**	**W2**	**CSA**	**Vol**
Extensor Carpi Ulnaris	22.0	1.5	1.5	1.8	38.9	10.4	0.6	0.7	0.3	3.4
Extensor Digitorum	22.0	1.3	2.2	2.2	49.4	13.4	0.8	1.6	1.0	13.5
Anconeus	8.0	1.0	2.4	1.9	15.1	5.4	1.2	1.3	1.2	6.6
Brachialis	4.4	1.6	4.2	5.3	23.2	5.4	0.6	2.1	1.0	5.3
Brachioradialis	13.4	1.2	3.2	3.0	40.4	7.0	1.7	2.4	3.2	22.4
Extensor Carpi Radialis	17.4	1.2	2.3	2.2	37.7	7.8	0.7	2.0	1.1	8.6
Pronator Teres	16.6	1.1	3.4	2.9	48.8	3.2	0.8	1.0	0.6	2.0
Supinator	10.0	0.9	3.6	2.5	25.4	2.0	0.7	2.7	1.5	3.0
Palmaris Longus	10.0	1.2	2.2	2.1	20.7	tendon only				
Flexor Carpi Radialis	20.4	1.7	1.8	2.4	49.0	15	1.1	1.1	1.0	14.3
Flexor Carpi Ulnaris	24.6	1.8	2.0	2.8	69.6	6.8	0.6	1.3	0.6	4.2
Flexor Digitorum Superficialis	23.6	1.6	2.5	3.1	74.1	9.8	0.7	1.0	0.5	5.4
Flexor Digitorum Profundus	23.2	1.5	4.9	5.8	133.9	14.2	0.8	2.5	1.6	22.3
Pronator Quadratus	4.6	1.4	4.2	4.6	21.2	--	--	--	--	--
Extensor Pollicis Longus	15.4	1.0	1.8	1.4	21.8	2.6	0.8	1.0	0.6	1.6
Flexor Pollicis Longus	16.0	1.1	1.8	1.6	24.9	9.0	0.9	1.0	0.7	6.4
Abductor Pollicis Longus	12.2	1.3	1.9	1.9	23.7	2.6	0.7	0.9	0.5	1.3
Totals	**Sound Limb**			**47.6**	**717.9**	**Residual Limb**			**15.5**	**120.2**

Table C: Measurements of musculature in subject C. Residual forearm length from distal humerus is 7.0 cm. Muscle dimensions: L=length; W1=minor width; W2=major width; CSA=cross-sectional area; and Vol=estimated volume. All measurements are in centimeters (cm).

Muscle	**Sound Limb (Left)**					**Residual Limb (Right)**				
	L*	**W1**	**W2**	**CSA**	**Vol**	**L**	**W1**	**W2**	**CSA**	**Vol**
Extensor Carpi Ulnaris	>5.8	0.4	0.8	0.3	1.5	2.8	0.3	0.4	0.1	0.3
Extensor Digitorum	>7.0	0.9	1.6	1.1	7.9	2.1	0.5	0.8	0.3	0.7
Anconeus	6.6	0.8	1.4	0.9	5.8	2.6	0.6	1.5	0.7	1.8
Brachialis	4.4	1.8	3.4	4.8	21.1	1.4	0.5	2.0	0.8	1.1
Brachioradialis	>7.8	0.9	1.2	0.8	6.6	3.0	0.5	1.8	0.7	2.1
Extensor Carpi Radialis	>7.8	0.3	1.9	0.4	3.5	3.0	0.4	2.2	0.7	2.1
Pronator Teres	>7.8	0.7	2.7	1.5	11.6	4.2	0.6	1.7	0.8	3.4
Supinator	5.4	0.5	2.9	1.1	6.1	3.0	0.8	2.1	1.3	4.0
Palmaris Longus	>6.8	0.9	1.3	0.9	6.2	3.0	0.6	1.1	0.5	1.6
Flexor Carpi Radialis	>7.6	0.7	1.8	1.0	7.5	5.0	0.4	1.3	0.4	2.0
Flexor Carpi Ulnaris	>7.6	0.9	1.8	1.3	9.7	2.4	0.5	0.6	0.2	0.6
Flexor Digitorum Superficialis	>7.0	1.0	1.6	1.3	8.8	5.2	0.8	1.8	1.1	5.9
Flexor Digitorum Profundus	>6.6	1.1	2.8	2.4	16.0	3.8	0.8	2.7	1.7	6.4
Totals	**Sound Limb**			**17.8**	**112.4**	**Residual Limb**			**9.4**	**31.9**

* ">" indicates that the muscle continues beyond the imaged area.

Table D: Measurements of musculature in subject D. Residual forearm length from distal humerus is 9.6 cm. Muscle dimensions: L=length; W1=minor width; W2=major width; CSA=cross-sectional area; and Vol=estimated volume. All measurements are in centimeters (cm).

Muscle	Sound Limb (Left)					Residual Limb (Right)				
	L*	W1	W2	CSA	Vol	L	W1	W2	CSA	Vol
Extensor Carpi Ulnaris	>11.4	1.0	2.7	2.1	24.2	4.8	0.6	1.0	0.5	2.3
Extensor Digitorum	>13.4	1.2	2.2	2.1	27.8	7.0	0.4	1.8	0.6	4.0
Anconeus	6.4	0.6	1.8	0.8	5.4	3.2	1	2.3	1.8	5.8
Brachialis	4.6	2.2	3.7	6.4	29.4	1.4	0.6	2.8	1.3	1.8
Brachioradialis	11.0	1.0	2.2	1.7	19.0	3.4	0.7	1.8	1.0	3.4
Extensor Carpi Radialis	13.2	1.7	2.8	3.7	49.3	3.6	1.0	1.5	1.2	4.2
Pronator Teres	11.0	0.7	3.6	2.0	21.8	3.2	0.8	2.1	1.3	4.2
Supinator	9.8	0.8	3.7	2.3	22.8	4.0	0.8	3.2	2.0	8.0
Palmaris Longus	--	--	--	--	--	--	--	--	--	--
Flexor Carpi Radialis	>11.6	1.1	2.2	1.9	22.0	4.0	0.5	3.9	1.5	6.1
Flexor Carpi Ulnaris	>11.4	1.2	1.8	1.7	19.3	4.4	0.4	2.7	0.8	3.7
Flexor Digitorum Superficialis	>11.4	2.0	2.3	3.6	41.2	6.0	1.1	3.2	2.8	16.6
Flexor Digitorum Profundus	>10.4	1.4	4.6	5.1	52.6	3.2	1.4	2.0	2.2	7.0
Flexor Pollicis Longus	4.8	0.8	1.3	0.8	3.9	1.4	0.3	0.7	0.2	0.2
Extensor Digitorum Minimi	>5.8	0.7	0.8	0.4	2.6	0.8	0.3	0.4	0.1	0.1
Totals	**Sound Limb**			**34.7**	**341.4**	**Residual Limb**			**17.3**	**67.5**

* ">" indicates that the muscle continues beyond the imaged area.

Table E: Measurements of musculature in subject E. Residual forearm length from distal humerus is 10.4 cm. Muscle dimensions: L=length; W1=minor width; W2=major width; CSA=cross-sectional area; and Vol=estimated volume. All measurements are in centimeters (cm).

Muscle	Sound Limb (Right)					Residual Limb (Left)				
	L*	W1	W2	CSA	Vol	L	W1	W2	CSA	Vol
Extensor Carpi Ulnaris	>7.0	1.1	1.3	1.1	7.9	3.8	0.8	0.9	0.6	2.1
Extensor Digitorum	>7.8	1.2	1.6	1.5	11.8	5.0	0.5	1.7	0.7	3.3
Anconeus	7.6	1.0	2.2	1.7	13.1	4.6	0.7	1.0	0.5	2.5
Brachialis	4.4	1.8	4.5	6.4	28.0	1.8	1.4	3.2	3.5	6.3
Brachioradialis	>8.6	1.1	3.8	3.3	28.2	3.2	0.7	3.2	1.8	5.6
Extensor Carpi Radialis	>8.6	1.2	2.4	2.3	19.5	3.2	1.6	2.2	2.8	8.8
Pronator Teres	>8.6	1.9	2.8	4.2	35.9	2.8	1.4	1.6	1.8	4.9
Supinator	>6.2	0.8	3.4	2.1	13.2	4.0	1.1	3.3	2.9	11.4
Palmaris Longus	--	--	--	--	--	1.2	0.3	0.4	0.1	0.1
Flexor Carpi Radialis	>8.8	1.0	2.2	1.7	15.2	4.6	0.9	2.9	2.0	9.4
Flexor Carpi Ulnaris	>7.6	1.1	2.4	2.1	15.8	5.2	0.4	2.0	0.6	3.3
Flexor Digitorum Superficialis	>8.6	0.9	3.0	2.1	18.2	5.8	1.2	2.4	2.3	13.1
Flexor Digitorum Profundus	>7.4	1.2	2.8	2.6	19.5	2.2	0.9	3.7	2.6	5.8
Extensor Digitorum Minimi	>2.4	0.5	1.2	0.5	1.1	2.0	0.4	0.7	0.2	0.4
Totals	**Sound Limb**			**31.6**	**227.5**	**Residual Limb**			**22.3**	**77.3**

* ">" indicates that the muscle continues beyond the imaged area.

The use of the Hilbert transform in EMG Analysis

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1.1 INTRODUCTION

The Fourier transform has traditionally been used for the detailed analysis of EMG signals. This has yielded many useful results, none more so than the descriptions of the energy produced at differing frequencies. This has been invaluable in the development of robust EMG controllers and the analysis of active or diseased muscle. Recently, *Wavelet* analysis has been applied to the study of EMG signals and it has provided additional insight into the underlying structure of the signal. Both these methods have drawbacks, the Fourier transform relies on analysis of complete wavelengths to describe a signal. Wavelet analysis cannot resolve any event less than the length of the fundamental Wavelet. These factors manifest themselves as a smudging or broadening of the spectrum and therefore they lead to inprecissions in the results.

Empirical Mode Decomposition (EMD) and the Hilbert Transform (HT) have been applied to analyse the the EMG signal. This is a method that is used extensively in the fields of seismology and meteorology and is now being applied to biological data [2]. It is particularly good a resolving signals that are not based on continuous sinusoids. It has been used on EMGs to show that the energy in the signal is significant at frequencies up to 2KHz.

The paper will present the results of a study of signals derived from a range of prosthesis users and non-users. The results from the Hilbert transform will be compared with results obtained using conventional methods of analysis.

1.2 COMPARISON WITH OTHER METHODS

In the field of evolutionary signal analysis the Spectrogram (Fourier transform derived) has become the de-facto standard. There are many reasons for this, it is developed from an established technique (Fourier Transform) therefore people are accustomed to it. It has some major shortcomings, the Fourier transform is very bad at representing discontinuous signals. It also gives very poor frequency response or poor localization in time. These facts are well known but still they are used to analyse discontinuous signals. Attempts have been made to ameliorate this problem but these are far from satisfactory. Wavelet analysis was developed to address some of the shortcomings of the spectrogram but again this creates new problems. The concept of Wavelet analysis is similar to the Fourier transform, in that a prototype wave is used to describe the signal. In the Wavelet case this wave is usually symmetric and dilateable

in time therefore providing a degree of localisation. The major problem with Wavelet analysis is that there is no fixed way to select the fundamental (basis) function, and that events shorter that this are not well localised. These problems invariably manifest themselves as a smudging of the spectrum around the point of interest. The Hilbert Spectrum does not share this problem. Figure1.1(left) is a plot of the two individual sine waves (the first 200 samples),

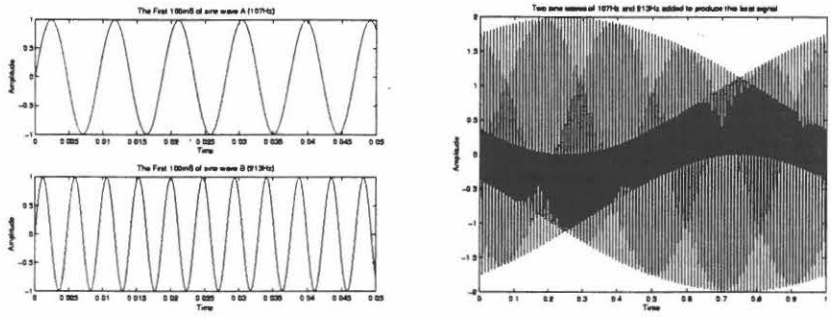

Figure 1.1: The individual sinewaves (first 200 samples.) and the result of summing them

the right shows them added together. This compound signal was analysed by a conventional Fourier based Spectrogram and by the EMD based Hilbert Spectrum. Figure1.2 illustrates the

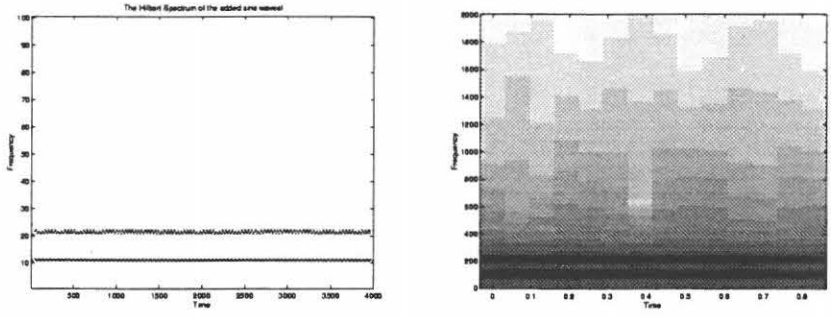

Figure 1.2: The Hilbert Spectrum and the Spectrogram of the Summed Signal

differences between the Fourier derived Spectrogram and the HT and EMD derived Hilbert Spectrum. The two frequencies are clearly defined by the two separate lines (on the left hand plot) unlike the conventional Fourier Spectrogram where there is a vast amount of smudging around the two lines..

1.3 THE HILBERT TRANSFORM

The Hilbert transform is a multiplication of the phase of all frequencies in a signal by 90^o("$\imath$"). For example a $\sin(\theta)$ becomes $\cos(\theta)$. From this transformation it is possible to construct an analytical signal, of the form $Z(t) = X(t) + \imath Y(t)$ where $Y(t) = \frac{1}{\pi} P \int_{-\infty}^{\infty} \frac{X(t')}{t-t'} dt'$ for the signal

under test [1]. For the HT to be well behaved there have to be certain constraints imposed on the signal, the signal must have an equal number of zero crossings and extrema, and be symmetric relative to the local mean ([1]). To get a measure of the frequency of the signal we define the instantaneous frequency as $\omega = \frac{d\theta(t)}{dt}$. Then from the analytic signal it is possible to resolve frequency information.

1.4 EMPIRICAL MODE DECOMPOSITION AND INTRINSIC MODE FUNCTIONS

To satisfy the conditions for a well behaved HT the Intrinsic Mode Functions (IMFs) must be symmetric with respect the local mean and they should have equal numbers of extrema and zero crossings [1]. Having a well behaved HT is a starting point. Invariably some of the data will not be an IMF; it will have many oscillatory modes. The data needs to be sifted to find the IMFs with one oscillatory mode. This is achieved through the fitting of *Splines* to the positive and negative maxima and minima of the signal. These two splines are then averaged to check if they satisfy the requirement for equal numbers of zero crossings and extrema and local average constraints. If it does not then splines are fitted to the extrema again and the process repeated until it does. Once it conforms this is a IMF. This signal is then subtracted from the original signal and remnant is then used as the starting signal. Splines being fitted to this and the above process repeated until the remnant is a trend or the total error is below the pre defined threshold. The set of individual signals if summed would produce the original signal with in the defined tolerance. These are the IMFs, they can now undergo the Hilbert Transformation.

1.5 INSTANTANEOUS FREQUENCY

In Figure 1.3 are two sets of figures, the two on the right are a section of the Basis functions, zoomed to the first 200 samples. The tapering effect is due to a window that was applied to the data to reduce the end effects. The two one the right are the total error and again the zoomed total error for the first 200 samples. The initial fluctuation is due to the spline fitting end effects. This problem is currently being addressed. Towards the center of the wave the error is minimal. This being set in the program, it is possible to achieve lower total error but at this time the end effects become pronounced. From the transformed IMFs it is now possible to attain the instantaneous frequency of the IMFs (as defined in section 1.3). With the frequency information it is possible to plot an image of frequency vs. time with amplitude calculated from the analytic function, represented as colour.

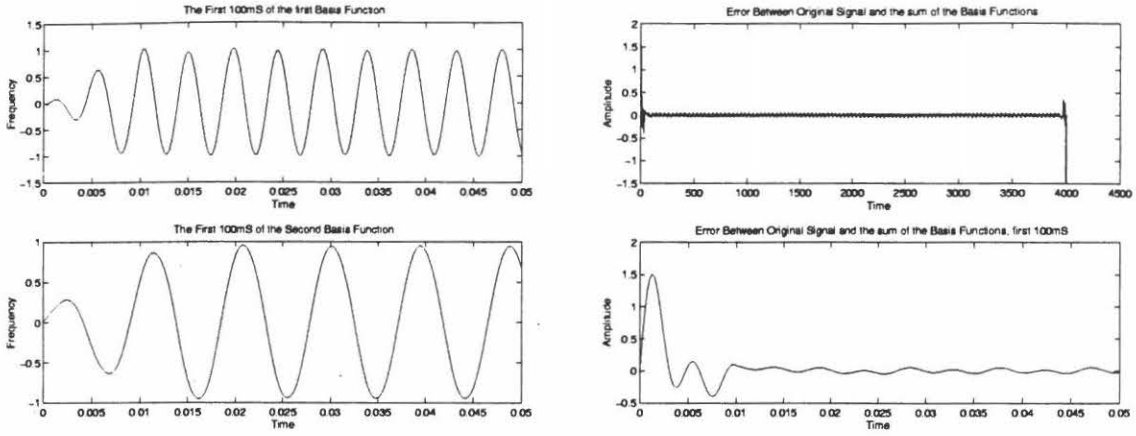

Figure 1.3: The resolved basis functions (first 200 samples) and the total error

1.6 DISPLAYING THE DATA

To construct a the Hilbert Spectrum from the data it has to be sorted into bins across the frequency range stepwise in time. This enables an image to be built up over successive time steps. The resolution is limited to the size of the bins. On the frequency side the maximum resolution is about 1/3rd the sampling rate. It is possible to produce a marginal Hilbert spectrum to compare with the Fourier spectrum. The marginal spectrum is an integration of the transformed IMFs with respect to time. Figure 1.2 is an example of the Hilbert Spectrum.

1.7 RESULTS OF PROCESSING EMG DATA

The following figures are images of the analysis of a real EMG signal from a user of a myoprosthesis. The bandwidth of the Hilbert transform is about 2/3rd of the sample frequency, with rounding errors this drops to about 1200Hz for a signal sampled at 4000Hz. Figure 1.4 is the Hilbert Spectrum and Power Spectral Density(PSD) of an EMG signal. The top plot (Hilbert) shows that a signal exists during the duration of the sampled data. The dotted line represent an analysis over the active portion signal. The bottom plot (PSD) conversely shows signal probably existed during the sampled data, again the dotted lines represent an analysis of the active portion. The technique requires added refinment, currently there are errors due to problems associated with the spline fitting program. The Hilbert Spectrum should be zero at zero frequency for EMG signals.

One of the interesting aspects of the signal when viewed as the Hilbert Spectrum is the high frequency energy in the signal when there is no contraction (left hand side of Figure 1.5). The right hand side shows the Fourier spectrogram, it is very hard to discern from this how much energy there is the in periods of little EMG activity. This high frequency energy causes

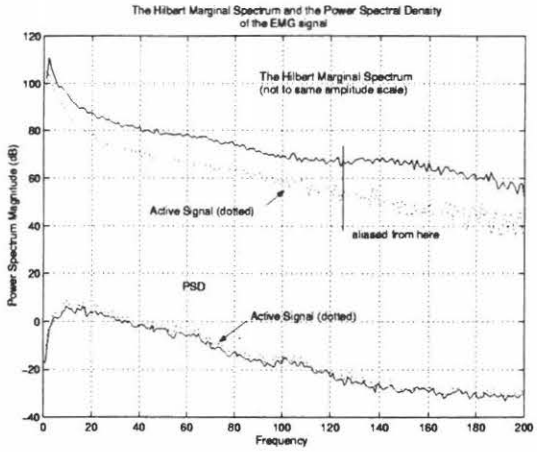

Figure 1.4: The Marginal Hilbert Spectrum (solid) and the PSD (dashed) for the active portion of an EMG signal

the PSD of the signal to become slightly skewed. In Figure 1.6, the first image on the left is calculated from the whole signal, whilst that on the right is only calculated for the active portion of the signal. The difference in magnitude is about 5db across the spectrum. This effect is more obvious in the Marginal Hilbert Spectrum of Figure 1.4. The lower graphs are the PSDs of Figure 1.6 overlayed, the difference is not clear, whilst the in the plots of the Marginal Hilbert Spectrum the difference is clear.

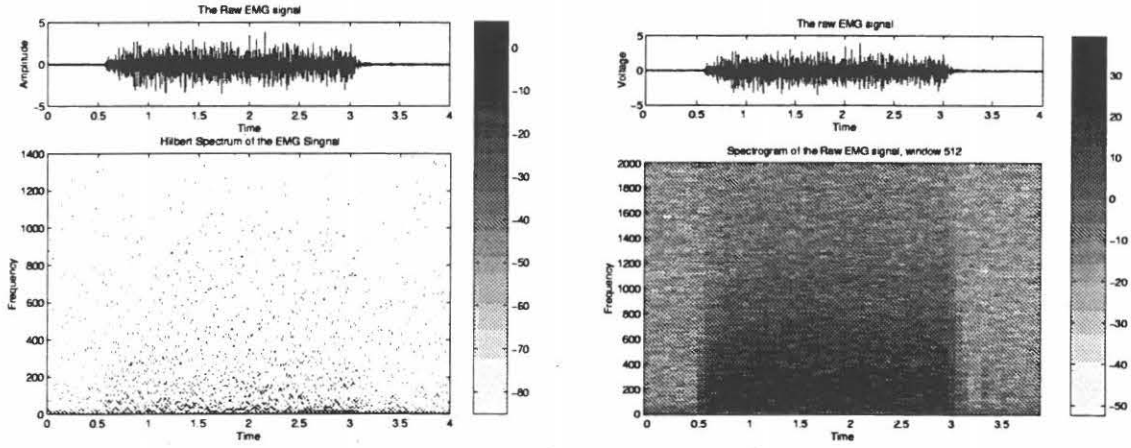

Figure 1.5: The Hilbert Spectrogram and a Fourier Spectrogram for the same signal

There are still some problems that need to be addressed with this technique, notably the spline fitting process which introduces errors. This tool is now being applied to many examples of EMG data, it is hoped that it will reveal some interesting insights to the structure of the Electromyographic Signal.

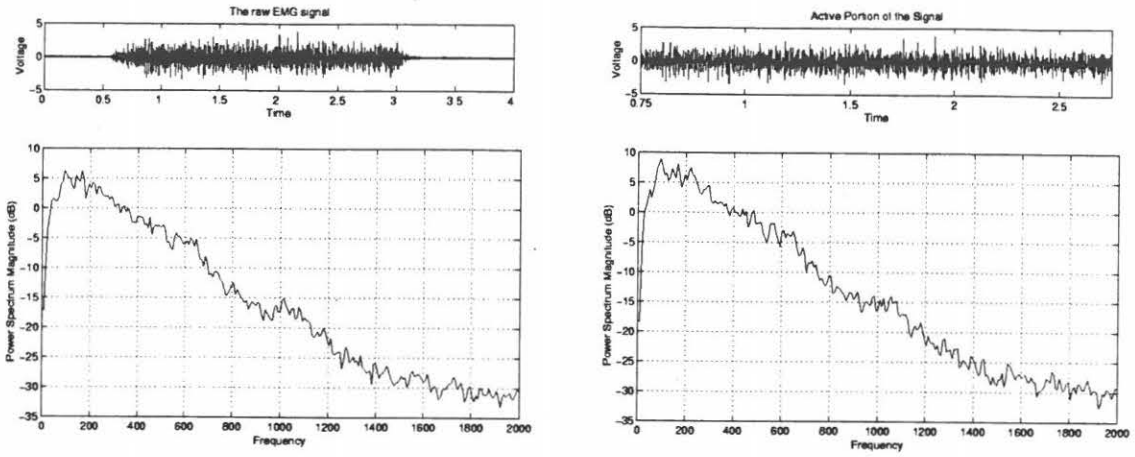

Figure 1.6: The PSD for the whole signal and just for the active signal

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The Use of Fuzzy Logic In the Processing Of Myoelectric Signals

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1.1 ABSTRACT

This paper describes the use of Fuzzy logic for the processing of EMG signals. This can increase the recognition rate and significantly reduce the number of computations required to generate an output. The initial placement of the Fuzzy sets was accomplished with the use of neural network techniques, these are not required for in the final system, only for setting up. The effectiveness of the features extracted from the EMG signals has been assessed using Principal Component Analysis (PCA). The developed system exhibits good generalisabilty but performs better when tuned to the intended user.

1.2 THE PRINCIPALS OF FUZZY LOGIC

Fuzzy logic is the a relatively new technique for computing using vague linguistic concepts. It was developed in 1960 by Lofty Zadeh [4]. It is now becoming accepted as a method for solving difficult engineering problems. Fuzzy systems excel in complex well defined problems, for instance the problem of target tracking. An implementation of this is designed with a Kalman filter and has a very rough transfer plane where as one that is implemented in Fuzzy Logic has a much smoother transfer plane and also uses much less computational power.

Conventionally most computing is performed on real, precise concepts. A computer has no concept of "tall" or "short", this is a human construct. In spite of the fact that they are inherently vague we still manage to communicate significant amounts of information using these terms. Fuzzy logic encodes this vagueness. It allows a value to be a "member" of a group. The "membership" of the group is defined as how much the value belongs. It is usually in the range 0 to 1. So using the example of membership functions in Figure 1.1a a value of 20 would be a member of Zero 0.5 and Low 0.5. The value is therefore a member of two groups, this fact is retained until the final defuzzification. While the inputs are "Fuzzified" they are operated on by a set of rules. The rules define the behavior of the system they map the input sets to the output sets. At this stage there maybe multiple active output sets it is not until the outputs are defuzzified that the output becomes real valued. This is a simple overview of a fuzzy system for a more detailed approach see [2]

1.3 THE EXPERIMENTS

Data was initially collected from normal healthy volunteers, usually from the forearm around the region of the Flexor carpi radialis, Palmaris longus and the Flexor carpi ulnaris. Later, users of upper limb prosthesis were also recruited. The volunteers were required to produce

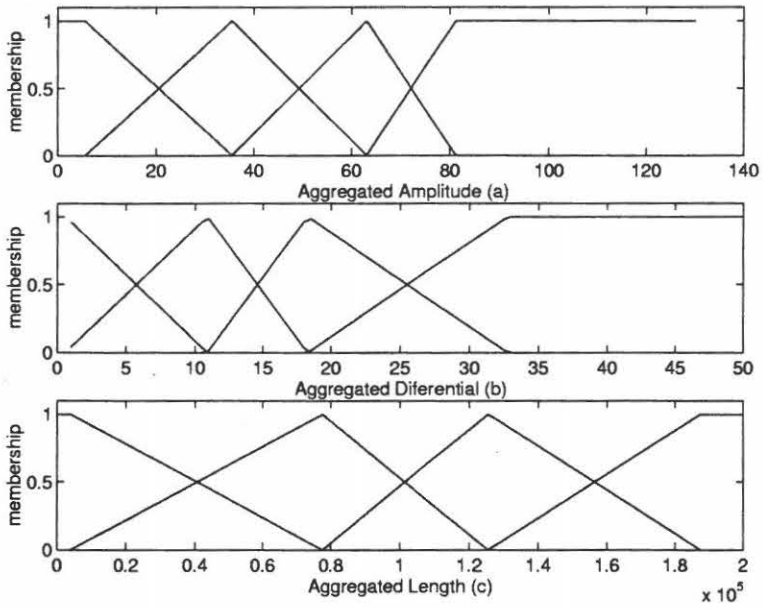

Figure 1.1: The Fuzzy Membership Functions (a) Average amplitude, (b) Averaged Differential, (c) Waveform Length

three distinct contraction levels and a zero. All subjects managed this will little or no training. For the able bodied volunteers the force level was also recorder using a simple force bridge. This enabled and a verification of the force level in the early stages of development. Twenty contractions of two seconds duration were recorded from each subject for each level. For the processing of the data the start of contraction was ignored. This avoids the problems associated with determining and detecting the start of contraction. As the target system is contraction level driven this is unlikely to present a significant problem in a real time system.

1.4 SIGNAL PROCESSING

The system decoded four levels, of EMG activity (including rest). These levels corresponded to distinct, increasing force levels in the able bodied users. In order to process the data, the amount of information in the signal has to be reduced. This was achieved through feature extraction. Similar features to those used by Hudgins [1] were extracted from the signal. The amount of information contained in these features was verified using Principal Component Analysis (PCA). This allows the amount of information contained in the data to be visualised easily, therefore making the assessment of the features' information content simple.

1.4.1 Placement Of The Sets

The optimal placement of the Fuzzy sets was achieved with Kohonen Clustering. This is a learning algorithm that moves the "center node" to the centre of a cluster. This is a standard

neural network technique which produces optimal clusters in N dimensional spaces [3]. The cluster centres were then used as the focus points for the fuzzy membership functions.

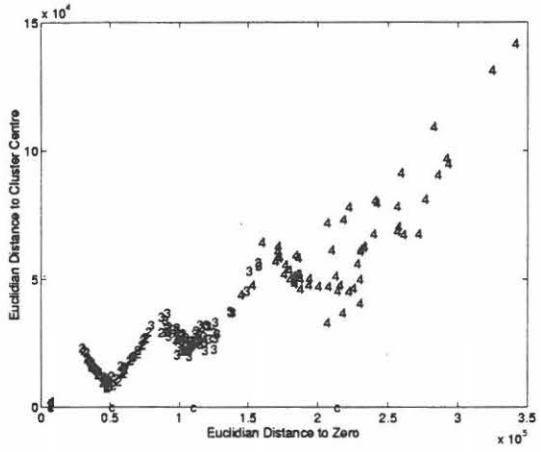

Figure 1.2: Kohonen centres with 4 nodes and the classifications of the Training data set

1.4.2 Fuzzy Analysis

The Fuzzy system aggregates the results from the fuzzification of the different features, but still retaining essential information. In comparison a Kohonen Network or Radial Basis Function Neural (RBFN) network would have to process one long vector. This would be composed of the individual feature vectors, resulting in long processing delays. Fuzzy logic provides a shorter route from the combination of the information extracted from the signal to an output. The production of a real valued output or defuzifiaction is relatively simple as the output is discrete steps. The simplest form of diefuzification was used, the maximum value from the Fuzzy processing was taken as the output.

1.4.3 The Features

The features extracted from the signal are as follows: Average Amplitude, Average Differential and Waveform Length. These are computed over a 200ms window of EMG data.([1]).

1. *Mean Absolute Value.* This gives an indication of the amplitude of the signal.

$$\overline{X}_i = \frac{1}{N}\sum_{n=1}^{N} abs(x_n) \qquad \text{for } i = 1, \ldots, I \tag{1.1}$$

where x_n is the n^{th} segment and I the total number of segments in the entire sampled signal.

2. *Waveform Length.* This is an indication of the activity of the signal, it is the sum of the absolute values of the signal over the window length.

$$L_i = \sum_{n=1}^{N} (abs(x_n - x_{n-1})) \qquad \text{for } i = 1, \ldots, I \tag{1.2}$$

3. *Zero Crossings* The number of times the signal crosses the zero axis within each window.

$$Zc_i = \sum_{n=1}^{N} \text{sgn}(-x_i * x_{i+1}) \qquad \text{for } i = 1, \ldots, I \tag{1.3}$$

$$sgn(x) = \begin{cases} 1 & \text{if } x > 1 \\ 0 & \text{otherwise} \end{cases}$$

4. *Differential.* This is the average differential within the window. Hudgins uses the difference between successive windows (equation 1.4) but as the signals that the system were trained on were constant, the average differential within the window was used (equation 1.5). The difference between successive windows was minimal when using Equation 1.4.

$$\Delta \overline{X}_i = \overline{X}_{i+1} - \overline{X}_i \qquad \text{for } i = 1, \ldots, I-1. \tag{1.4}$$

$$D_i = \frac{\sum_{n=1}^{N} (x_n - x_{(n-1)})}{N} \qquad \text{for } i = 1, \ldots, I \tag{1.5}$$

Equation 1.5 gave a better indication of the activity of the signal.

1.5 RESULTS

The results are displayed in table 1.5. When trained on three subjects the system performed well, yielding a 84% classification rate including the test signals. The test signals were not used in the training of the system. The system was optimized for one subject, (yielding a 100% classification rate) the total classification for the test data set was 95%.

	Classification Rate			
	Training Data		Test Data	Overall
Method	individual	all	all	all
Kohonen Network	84%	97.1%	93.3%	95.2%
RBFN	94%	Na	79%	86%
Fuzzy Nwk	Na	92%	76.7%	84.4%
Opt Fzy Nwk	100%	Na	93.3%	96.7%

Na = Not applicable

Table 1.1: Classification Performance

1.5.1 Feature Analysis

The extracted features were analysed with PCA, figure 1.3 is a plot of the first Principal component vs. the second PC. The figure demonstrates that the features are separable in the higher order PC therefore they are useful for separation of different force levels. As it is not possible to completely separate the features other features have to be used to add information. All the features were assessed in this way. However, Zero crossings when analysed this way, does not contain any useful information.

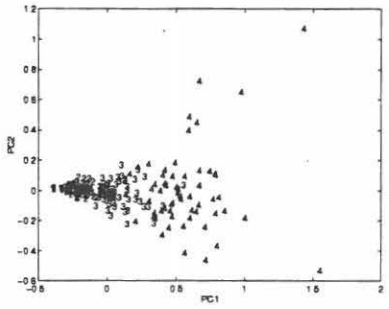

Figure 1.3: PC1 vs PC 2 for Average Differential

1.5.2 Computation Time

To evaluate the increase in computation efficiency the number of Floating Point Opperations (FLOPs) required to produce a result was recorded for the different algorithms. This gives a relative measure of the native performaces of the algorthms. The number of FLOPs is detailed in Table 1.2. The Fuzzy system performs nearly ten times faster than the Kohonen Network. The RBFN though having a higher classification rate requires many more FLOPS to achieve the same result.

	One Feature	Three Features
Kohonen Nwk.	480	1440
RBFN	34200	88000
Fuzzy Nwk.	Na	366

Table 1.2: Number of Floating Point Operations to produce an output

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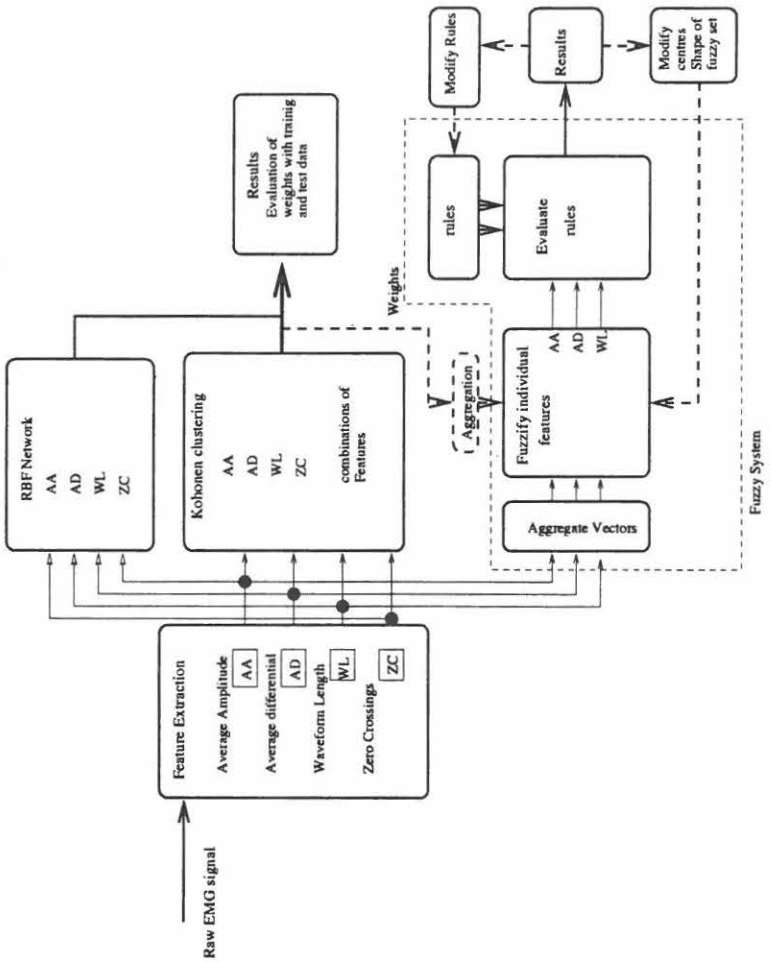

Figure 1.4: The Flow Diagram of the Fuzzy System and Neural Network Based Components

[4] L. Zadeh. Fuzzy sets. *Information and Control*, 8(3):338–353, June 1965.

NATURAL CONTROL OF KEY GRIP AND PRECISION GRIP MOVEMENTS FOR A MYOELECTRIC PROSTHESES

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ABSTRACT

Hand prosthesis function is augmented when the user can employ lateral grasp as well as traditional palmer grasp. Our goal in this investigation was to enable the below-elbow (BE) prostheses user to switch between and use these grasp modes in a natural and reliable manner. We recorded the EMG from residual muscles (*flexor dig; ext. dig; flex. pollicis longus; ext. pollicis longus*) involved in these grasp activities in an adult subject with below elbow (BE) amputation while she contracted her residual forearm muscles to mimic computer animations of different hand movements. To reduce crosstalk between the recordings from separate muscles, and to enhance the stability of the recording interface over the 30-day duration of the experimental sessions, we used chronically implanted percutaneous coiled wire electrodes implanted for 30 days (12 one-day sessions). Artificial Neural Network (ANN) pattern recognition techniques were used to extract voluntary command signals from the EMG signals. The mean absolute value (MAV) of the EMG signals was selected as a feature for training multilayer perceptrons. Initially, we trained ANNs having 5 hidden neurons using data from the 10th and 12th session individually (3 training sessions each). Three additional ANNs (sizes 4:7:4, 4:8:4, 4:9:4) were designed and trained (3 training sessions each) with combined data from experimental sessions 10 and 12. Subsequently, we separately tested the performance of these ANNs with data from the 9th, 10th and 12th experimental sessions. While the results showed that data from different experimental days were substantially consistent, more reliable recognition of the grasp mode from any arbitrary test sample (i.e. taken from test sessions 9,10 or 12), was achieved when we used an ANN that was trained with representative samples from more than a single experimental day (e.g. using 10th and 12th experimental days data for training). This produced mean rates of recognition (averaged over the results from the three ANN training sessions with network size 4:8:4) of 97.6% key grip closing, 83.3% key grip opening, 85.7% precision grip closing, 96.4% precision grip opening, for the combined evaluation data from all test sets.

We conclude that intuitive operator selection, between key grip and precision grip modalities, is feasible for cases of BE amputation using recorded myoelectric signals.

INTRODUCTION

Loss of an arm transforms former simple tasks into difficult and tiresome challenges. Current prosthetic arms include body-powered devices and myoelectrically controlled hands. Although body powered devices have an advantage in providing the user with rough sensorial feedback through a control cable attached to a harness the bulkiness of this harness presents a serious drawback to the user [1]. Moreover, this approach has serious limitations for control of a multiple-degree of freedom artificial arm. Myoelectric prostheses offer better promise to achieve multiple-axis control in spite of inherent drawbacks such us maintenance problems, high weight, difficulty in operating and learning myoelectric control, inadvertent operation, and unreliability [1]. Myoelectric control is performed by means of the electrical activity of contracting muscles and, therefore, can be highly intuitive. Myoelectric multiple-

axis control is feasible because the limit to the number of prosthesis functions is the amount of different contraction patterns produced by the user. However, control of present multiple axis prostheses is usually not natural from a motor-control point of view. To obtain more reliable and distinct contraction patterns, prostheses users are frequently requested to produce patterns which do not correlate in a natural way with the movement replaced by the prosthesis [2,3]. As the number of prosthesis functions increases so does the amount of training required for its operation. Additionally, separability of EMG patterns (regardless of the type of feature space being used) tends to decrease, as more patterns are required. Consequently, unfamiliar patterns of contraction may not be able to replace the normal motor control strategies for manual dexterity. Previous work [4,5] suggested that phantom limb phenomena are evidence of complex motor control skills still present in amputees and might be useful for prosthesis control. However, the issue of whether intuitive and reliable motor patterns can be conditioned through training is still unsettled. The present work utilizes a novel training protocol based on biofeedback and visual tracking tasks in a below-elbow (BE) amputee. The goal of the training was to rehabilitate the subject's motor control skills to control finger opening and closing for two different grip modalities, key grip and precision grip. Due to the expected day-to-day variation in the evoked contraction patterns, we employed a pattern recognition system based on Artificial Neural Networks to discriminate the users intended grasp motion.

EXPERIMENTAL PROTOCOL

The subject was an adult female (age 33) with BE left arm amputation (which was her non-dominant hand) with strong phantom limb sensation meaning that she could perceive the movement in her phantom hand and fingers when contracting her residual muscles. Eight of the subject's residual muscles were chronically implanted with bipolar pairs of percutaneous coiled wire electrodes for the duration of the 30-day study. Prior to the implantation protocol, the condition of the selected residual muscles was assessed by Magnetic Resonance Imaging techniques (MRI). Based on the MRI evaluation, localization strategies were developed to identify each of the target muscles. A pair of electrodes for bipolar recording was inserted with 2cm tip separation into each of the following muscles: (1) Flexor Digitorum, (2) Extensor Digitorum, (3) Flexor Pollicis Longus, (4) Extensor Pollicis Longus, (5) Pronator Teres, (6) Supinator (7) Flexor Carpi Radialis, and (8) Extensor Carpi Radialis, though only the first 4 muscles were studied with regard to the present investigation. Subsequently, the experimental protocol was conducted as 12 one-day sessions distributed over the 30-day period. We used data obtained during 3 of the last experimental sessions (9^{th}, 10^{th}, 12^{th} sessions) for analysis since any benefits of training could be expected to be achieved nearer to the end of the protocol. Results from session 11 were omitted because the subject reported during the recordings that on some occasions she could not move the fingers of her phantom hand as if they were "asleep". This phenomenon did not appear to be present during the other experimental sessions, however.

The subject was seated facing a PC visual display (Fig.1), and she was requested to mimic with her phantom hand (as a kind of tracking task) animations of a hand performing key grip (closing-opening) and precision grip (closing-opening). She did this by contracting her residual limb muscles. The phantom hand movements were performed in 4 sets of 25 repetitions of the same grip task before the task was shifted to the other grasp type, In every case, the movements started with the fingers extended (hand open position). A pause of 1s was included between each consecutive movement repetition. Additionally, the subject was permitted to rest between sets at her request. The multiple-channel EMG signals together with a reference signal with information about the phasing and progression of the animation were collected and stored digitally. The EMG signals were amplified between 1000 and 10000 and band-pass filtered (10Hz-1KHz) before being sampled (2KHz).

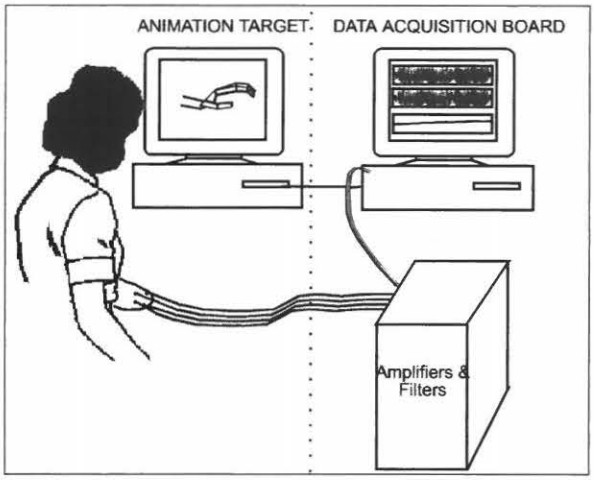

Fig.1. Experimental setup

OFF LINE DATA ANALYSIS

EMG signals were further processed to remove DC offset and motion artifacts (Digital high pass, Butterworth order 4, 20Hz). An initial visual examination of the EMG signals revealed which muscles presented reciprocal activity for opposing motor tasks (i.e. an increase in the activity of one muscle while a decrease occurs in the other). The muscles that presented the best signal/noise ratios and the greatest change of the EMG activity in the opposing motor tasks were selected as targets for detection of movement onsets. For precision grip closing and opening the selected target muscles were Flexor digitorum and Extensor digitorum, respectively, while for key grip closing and opening they were Extensor pollicis longus and Flexor digitorum.

Examination of the data from sessions 9 and 10 revealed a substantial reduction of differences between contraction patterns for some grip trials. Consequently, we applied a set of simple criteria to exclude some of those trials: (1) a highly deteriorated reciprocal activity during opposing motor tasks; (2) an insufficient level of contraction (peak value) of the agonistic muscle (less than 2 times the amplitude of the background activity). Additionally, in those cases where reciprocal activity could still be observed but was significantly reduced, the trials were excluded if they did not fulfill at least one of the following conditions: (a) existence of a clear delay from the first EMG spike of the agonistic muscle to the first EMG spike of the co-activating muscle, (b) a greater than 50% change in the peak to peak value of one of the muscles during the reciprocal movement.

After applying these criteria, 42 and 33 trials were excluded from session 9 and session 10, respectively. Furthermore, the last 25 trials of the precision grip movement in session 9 were also excluded because the subject complained of a sensation of "tiredness" in her phantom fingers during the execution of the precision grip motor tasks. Therefore, the number of precision grip trials passed onto the next stage of processing for onset detection were 33 for session 9 and 65 for session 10.

Onset detection

An arbitrary threshold was determined for each target muscle to be used in the detection of the movement onset. The mean value plus 2 times the standard deviation of the rectified background activity was used to specify the threshold value. To determine the existence of an onset event, the mean value of the rectified EMG signal from

the target muscle was computed (over a 25ms sliding window) and compared to the appropriate threshold. The condition for the existence of an onset event was based on the spatial (amplitude) and temporal (density of spikes) characteristics of the moving average signal. The algorithm applied a 200ms sliding window to the moving average signal. If the number of samples in this 200ms window exceeded the threshold for at least 175ms (not necessarily consecutively) then the first point of this 200ms window was labeled as the onset. The algorithm then began a search for an offset event, again using a sliding window of 200ms. An offset event was detected when the number of samples below the threshold exceeded 150ms. Following each onset detection, we extracted 200ms of raw EMG data from each of the muscles. These 200ms of data included an epoch of 25ms that preceded the onset event. These data were stored for subsequent feature extraction.

To train and test an ANN the amount of samples has to be the same for each movement. We detected 8 onsets for precision grip closing in session 9; 44 onsets for precision grip opening in session 10; and 45 onsets for precision grip opening in session 12. Although the other movements yielded a greater number of detected onsets, the amount of data available for training and evaluating an ANN was then limited to 8 samples per movement in the case of session 9; 44 per movement in session 10; and 45 per movement in session 12. Because there were so few input samples from session 9, we decided to use this set only for evaluation of the ANNs.

ANN studies protocol

The mean absolute value (MAV) was selected as the primary EMG feature for this study because of its computational simplicity and its extensive use in myoelectric prosthesis [3,6]. For every set of data collected at each detected movement onset, MAV was calculated for each of the 4 muscles. These values produced sample MAV vectors (input data samples) of four components. Input samples were normalized between [-1,1] independently for each day, using as boundaries the maximum and minimum values extracted from that day's session. Subsequently these samples were randomly assigned to training and test sets (80% and 20%, respectively, of the total amount of samples).

ANN training was performed under two paradigms: with individual data from each of the sessions 10 and 12, and also with combined data from sessions 10 and 12. NeuroSolutions™ software was utilized exclusively for these studies. ANN sizes were [4:5:4] for the ANNs trained with only one session's data and the sizes were [4:7:4, 4:8:4 and 4:9:4] for the ANNs trained with the combined 10^{th} and 12^{th} sessions' data. ANN architecture was feedforward (complete connectivity between adjacent layers) with hyperbolic tangential and softmax transfer functions for the hidden and output layers, respectively. The latter function permits interpretation of the results as probabilities. Fahlman's quickpropagation algorithm [7] was used for gradient descent. Three ANNs were trained (batch learning) with each training set and stored for posterior evaluation. The stopping criterion was that of the *Average Cost* between [0.003 0.006]. As an additional requirement, the errors (i.e. the Euclidean distances between the output and the target values) in the training set had to be from the same statistical population for each of the three ANNs trained with the same data. This last condition together with the stopping criterion warranted that all three ANNs were exposed to the same amount of knowledge. The rates of recognition in the training sets for all the ANNs were above 94% per movement, which is sufficient to assure local generalization. The evaluation of each ANN was performed with each individual test set (i.e. test sets 9,10 and 12).

RESULTS AND DISCUSSION

Results of the evaluation for the three ANNs trained from session 10 presented the recognition results (mean recognition rates $\pm$ the standard deviation) shown in Table I. The results of the three ANNs from session 12 are presented in Table II.

Session	Key grip closing	Key grip opening	Prec. grip closing	Prec. grip opening
9	**100** ± 0	**66.6** ± 7.2	**95.8** ± 7.2	**100** ± 0
10	**96.6** ± 5.7	**80** ± 10	**100** ± 0	**100** ± 0
12	100 ± 0	23.3 ± 5.7	86.6 ± 5.7	90 ± 0

Table I. ANN trained with set 10 and evaluated with test sets 9,10, and 12

Session	Key grip closing	Key grip opening	Prec. grip closing	Prec. grip opening
9	91.6 ± 7.2	62.5 ± 0	37.5 ± 0	0 ± 0
10	**96.6** ± 5.7	**70** ± 0	**90** ± 0	**80** ± 17.3
12	**80** ± 0	**96.6** ± 5.7	**90** ± 0	**90** ± 0

Table II. ANN trained with set 12 and evaluated with test sets 9,10, and 12

The best performance was obtained with the ANNs trained using the set from session 10 and evaluated with the test set from session 10. Additionally, those ANNs showed excellent recognition rates for 3 out of 4 movements when evaluated with the test set from session 9. However, the recognition rates when the session 12 data were used as the test set were unexpectedly low for the case of Key grip opening. In marked contrast, the ANNs trained with the session 12 data presented the opposite results: Rates of recognition were acceptable for the test set from session 10, but the results were unacceptably low with the test set from session 9, for both precision grip opening (0%) and closing (37.5%).

The limited success described above for the ANNs to perform consistently when presented with test data other than those they were specifically trained with, led us to expand the training set to include representative data from two different sessions, specifically 10 and 12.

The results of the evaluation for the ANNs trained using the combined data are presented in Tables III, IV, and V for network sizes 4:7:4, 4:8:4 and 4:9:4, respectively. ANNs trained using combined data with a hidden layer size greater than eight, showed acceptable performance (i.e. rate of recognition above 80%) for all movements over the three sessions with the exception of Key Grip Opening. In this latter case the session 9 data yielded a mean recognition rate of 66.6% for the ANNs with size 4:8:4. However, the general performance (i.e. the rate of recognition for the combined evaluation data from all of the test sets) yielded recognition rates above 80% for each movement (table VI).

Session	Key grip closing	Key grip opening	Prec. grip closing	Prec. grip opening
9	87.5 ± 0	62.5 ± 0	66.6 ± 14.4	100 ± 0
10	**93.3** ± 5.7	**80** ± 10	**93.3** ± 5.7	**100** ± 0
12	83.3 ± 5.7	80 ± 0	86.6 ± 5.7	90 ± 0

Table III ANN (4:7:4) trained with combined data and evaluated with test sets 9,10 and 12

Session	Key grip closing	Key grip opening	Prec. grip closing	Prec. grip opening
9	100 ± 0	66.6 ± 7.2	87.5 ± 12.5	100 ± 0
10	**96.6** ± 5.7	**90** ± 10	**90** ± 0	**100** ± 0
12	96.6 ± 5.7	90 ± 10	80 ± 0	90 ± 0

Table IV ANN (4:8:4) trained with combined data and evaluated with test sets 9,10 and 12

Session	Key grip closing	Key grip opening	Prec. grip closing	Prec. grip opening
9	95.8 ± 7.2	62.5 ± 0	91.6 ± 14.4	100 ± 0
10	**93.3** ± 5.7	**90** ± 0	**90** ± 0	**100** ± 0
12	96.6 ± 5.7	83.3 ± 5.7	86.6 ± 11.5	90 ± 0

Table V ANN (4:9:4) trained with combined data and evaluated with test sets 9,10 and 12

Size	Key grip closing	Key grip opening	Prec. grip closing	Prec. grip opening
4:7:4	88.0 ± 4.1	75 ± 3.5	83.3 ± 2.0	96.4 ± 0
4:8:4	**97.6** ± 4.1	**83.3** ± 2.0	**85.7** ± 3.5	**96.4** ± 0
4:9:4	95.2 ± 5.4	79.7 ± 2.0	89.2 ± 7.1	96.4 ± 0

Table VI Performance for (4:7:4), (4:8:4) and (4:9:4) ANNs evaluated with the combined data from all of the test sets

The improvement in the recognition rates obtained by training the networks with combined data can be better observed in figure 2

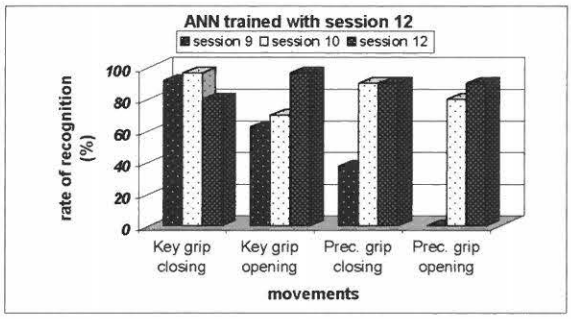

Fig 2.a Mean recognition rates for ANNs trained with 12 session's data

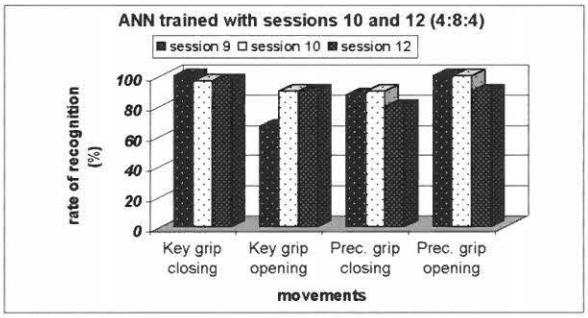

Fig2.b Mean recognition rates for ANNs trained with data combined from sessions 10 and 12.

CONCLUSIONS

The ability of a person with BE elbow amputation to produce intuitive EMG patterns to control the initiation of two different grip modalities has been demonstrated. Conversations with the subject revealed that she relied heavily on feedback from her phantom sensation throughout the training sessions. It is unknown to what extent the presence of phantom limb sensation is necessary for the success of the EMG command production. However, since the presence of non-painful phantom limb sensation in amputees has been reported to be between 80% and 100% [8], the training protocol might be workable for the majority of the amputee population. Moreover, an enhancement of phantom limb sensation might be achievable using a virtual reality training platform [9] to expedite the learning process. On the other hand, the amount of training required to achieve acceptable reliability of the EMG prosthesis control must still be determined. In the present studies, we have observed a certain degree of inconsistency in the generation of EMG activity. At least 3 different causes may contribute to this situation: (1) insufficient conditioning of the residual muscles, (2) insufficient training of the grip motor tasks and (3) disturbances in the user's phantom limb perception that may interfere in the generation of the motor commands. Nevertheless, for those signals that were processed, the ANN pattern recognition studies revealed a high level of consistency (MAV feature space) across different days. However, it was beneficial to combine data from two different sessions in order to achieve more uniform rates of recognition across all of the 4 hand functions studied. Future efforts include the extension of the combined EMG and ANN approaches to additional hand functions, such as voluntary wrist movements.

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[9] Ramachandran VS, Ramachandran D, "Synaesthesia in phantom limbs induced with mirrors". Proc R Soc Lond B Biol Sci 1996 Apr 22;263 (1369):377-86

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A PC BASED SYSTEM FOR SELECTING AND OPTIMIZING MYO CONTROLS TO THE PATIENT'S NEEDS.

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Contemporary systems for myoelectric fittings of upper limb amputees allow a variety of options for selection of a control system. The practitioner has to choose which system is the most appropriate for the patient's needs and demands.

A new device, consisting of an interface connecting the myo-electrodes to the PC and a Windows-based software, allows the practitioner to test the patient's capability for control of a myo-system. Based on this test, an appropriate control mode can be selected and evaluated by the patient. The PC system can emulate several modes.

During this phase of the session the EMG-signals of the patient are either shown as curves or they control a video animation of the selected myo-hand. In the first case all relevant levels of control are displayed. The practitioner therefore can evaluate if the selected control is right for the patient or not. Additionally, if more complex controls are selected – like those including co-contraction – every attempt at switching, if successful, is registered and displayed. If unsuccessful, the reason for failure is shown and the corresponding section of the graph is highlighted. Additionally possibilities for documentation are offered and a video game to motivate training is also included.

Training and selecting the right means of control for the myoelectric prostheses has shown that in general the amputee "adopts" and makes easier use of his prosthesis Patients recognize improved functionality and due to the higher motivation a much smaller risk of rejection is experienced.

Several modes of controlling a prosthesis are known. The most popular and the first to be positively accepted is the digitally controlled myo-hand. The more sophisticated control modes developed since then as e.g. the Dynamic Mode Control, better known as DMC, demonstrated the growing demand for devices that help adjust the electrodes and help verify the ability of the patient to build up enough EMG signal . Practitioners as well as health care providers wanted to be sure of the outcome of the specific fitting well ahead of time.

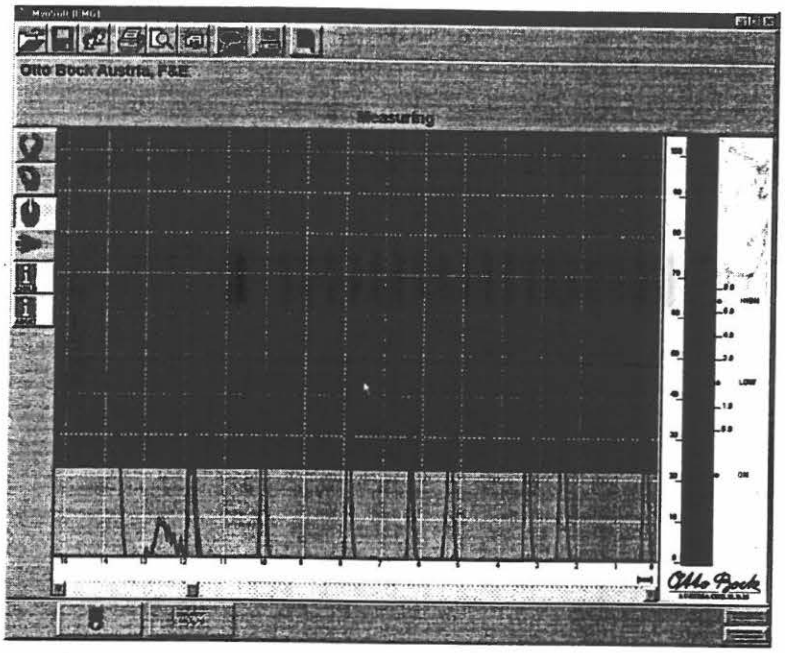
Practitioners sought for easy ways to fit different forms of control without needing to invest in several control relays or highly sophisticated programmable equipment. Therefore Otto Bock implemented standardized control modes which still can be individually adjusted to the needs of the amputee. Switching between these modes is done very comfortably by means of so called coding plugs. The color of these plugs corresponds to the type of control.

Nowadays with the many different functions offered by a prosthesis e.g. a myo-hand with proportional speed/force control and pro- and supination by either co-contraction or complex switching or the traditional 4-channel control, the practitioner together with his patient naturally wishes to gain the necessary comfort and experience in working with the chosen control. Documentation will help improve the relationship between health service, practitioner and patient.

Otto Bock has developed the MyoBoy, a set consisting of the hardware-interface, the electrode adapter cable, the grounding pin and a Windows-based software, all packaged in a soft padded case. The MyoBoy hardware a) shows the EMG-levels on a LED-display and b) digitizes the EMG-signals and transfers these via serial interface to the PC system.

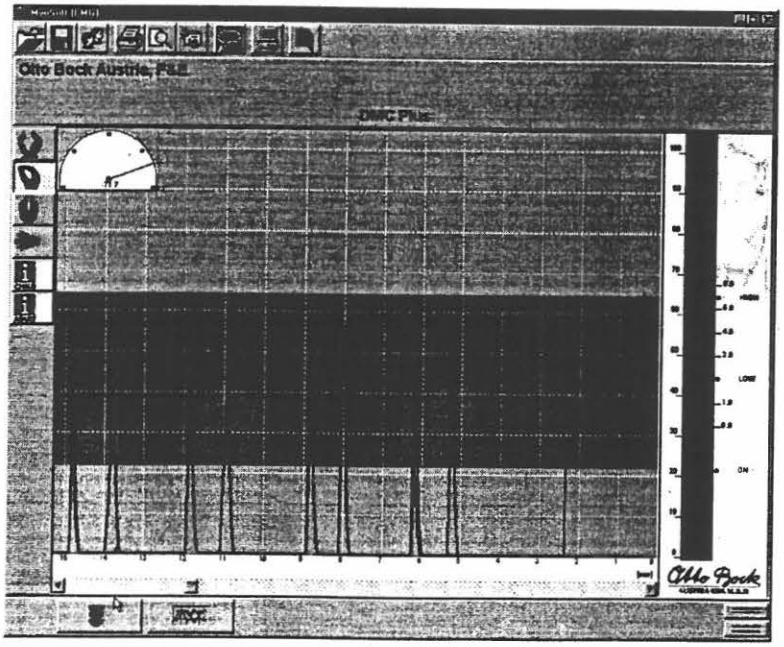
The software „Myosoft" shows these EMG-signals in different ways:

- Measurement mode



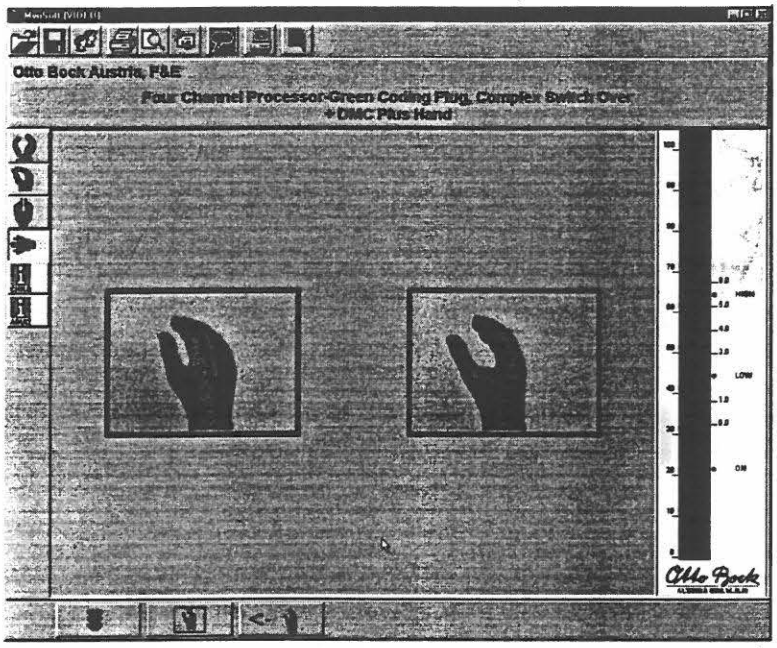
This display mode is used to locate the best electrode position(s). location for the electrodes. Furthermore the EMG-signal strength can be determined. The mark „ON" to the right side of the display must be exceeded.
Depending on how high the signal is (markings „LOW" and „HIGH") the appropriate control mode can be chosen e.g. Digital, DMC or DMC Low Input.

- Training mode



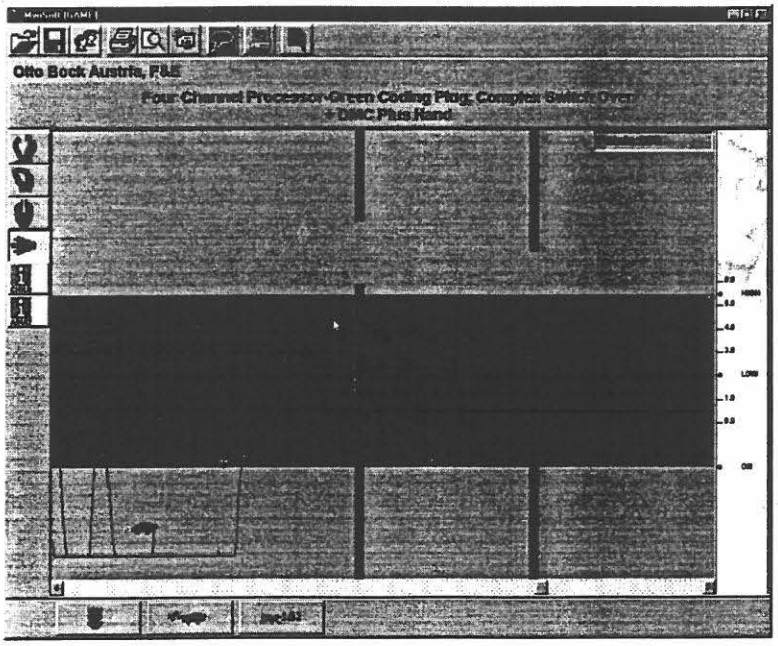
The practitioner will now adjust the electrodes in a way that the patient can best utilize his EMG-signals. He should easily reach the mark „ON" and only with strong effort reach the mark „HIGH". The proportionality of the system will thus be best used.
In the upper left corner the signal strength in percentage of the „HIGH" will be indicated. With this gauge one can prove the ability of the patient controlling DMC components.

- Animated mode



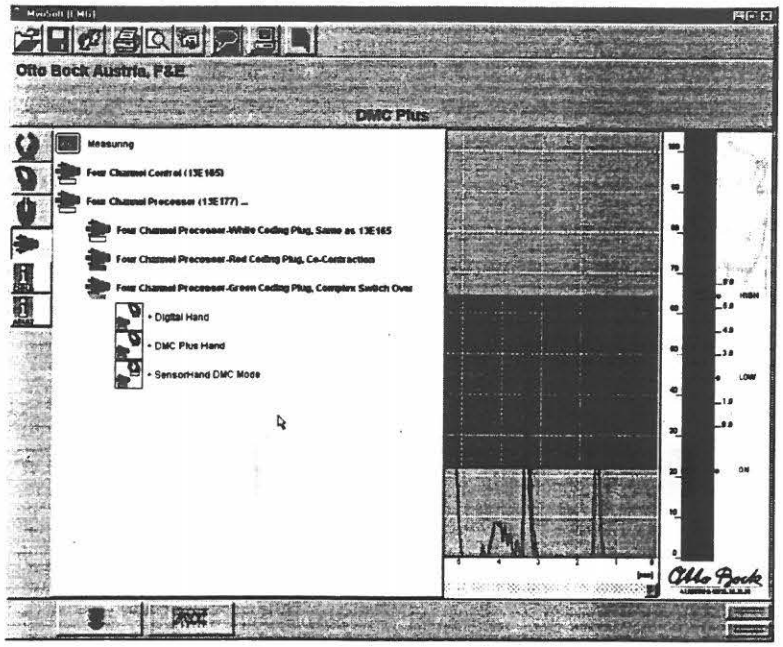
The display shows a sequence of pictures either opening/closing or pro- and supinating a hand prosthesis depending on the EMG-signals of the patient. The number and frequency of the individual pictures creates an animation similar to a video. The patient sees how he is controlling his prosthesis and how he is switching between the grasp and rotational modes.

- Game mode



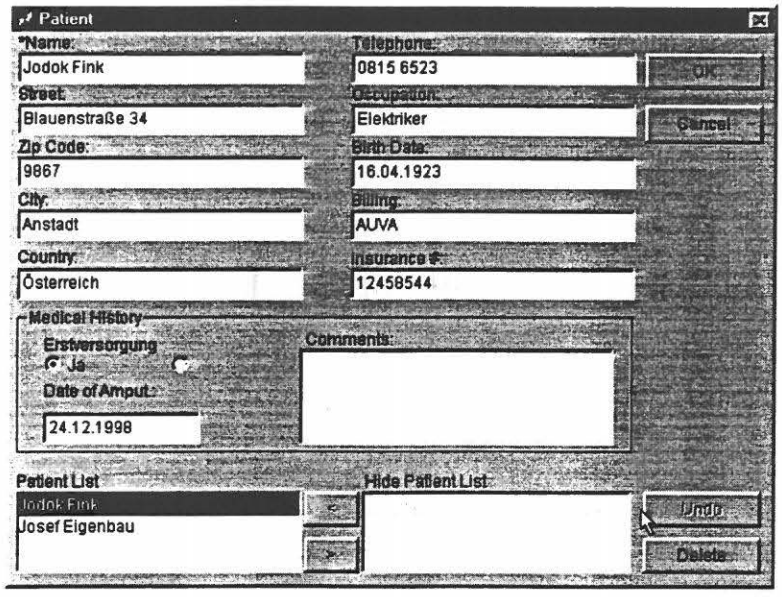
To especially motivate younger amputees a video game has been included. The car shown has to be driven through the barriers otherwise it will crash. In the upper right corner the amount of points are shown, adding 5 points each time the car passes the barrier, subtracting 5 points each time the car crashes.

- Indication



The practitioner can go through a process suggesting appropriate fitting and control possibilities depending on EMG-signal strength and number of electrode sites. Still simulating the control mode and closely watching and discussing the outcome with the patient on the MyoBoy a rather precise decision can be made.

- Patient Database



A patient data base helps keep track of the success of training sessions, chosen mode control type, date of fitting, et cetera

The goal to develop a flexible, upgradable, easy-to-learn and patient-motivating measuring as well as training instrument has been achieved. Nevertheless many other future „MyoBoys" will still have to be developed in order to improve the level of education and acceptance within the exciting field of myo-electric prosthetics.

LOW LEVEL RESPONSE OF BOCK AND STEEPER ELECTRODES

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ABSTRACT

For many years myoelectric fittings have been limited by the need for signals of at least 20μV when using convenient in-socket myoelectrode-amplifiers. With the introduction of the Bloorview-MacMillan MyoMicro™ technology, the controller is able to further amplify the patient myosignal. To evaluate the Bock 13E125 and Steeper Electrodes for providing suitable signals for control in the region below 20μV, both electrodes were tested at each gain setting. The Bock electrode gives the best signals in the 1 to 5μV range while the Steeper electrode is less sensitive. With suitable downstream amplification, both electrodes will provide proportional control with signals of 0-10μV.

BACKGROUND

When prosthetics mainly used threshold electronics, it was sufficient to have electrode-amplifiers (called just electrodes by most manufacturers) with a minimum threshold transition of about 20μV. Proportional control, however, requires a range of output values to control speed, grip force or other useful functions. Further, many patients have the capability of making maximal sustained contractions that generate as little as 5μV of signal. To handle this class of patient, Motion Control and Liberty Technology, but especially the former, have developed signal acquisition systems with the required sensitivity.

While the specialist may wish to use many different electrode systems, practitioners making only a few myocontrolled systems a year would like to use just one electrode for everything. With the introduction of the Liberty Technology VariGripII™ Controllers for one to three devices, it is now possible to add digital, post-electrode amplification to achieve control over a wide range of muscle input amplitudes. Bock and Steeper electrodes were evaluated alone and with the VariGripII system to see just how well they work as patient interfaces.

Three types of tests were done. First, each electrode was tested with a low level balanced differential input. Next, each electrode was used to generate output control signals for a Liberty Technology VariGrip Controller using the simulated input. Finally, systems were shipped to practitioners who have fitted patients requiring great sensitivity.

The Test Setup

An artificial input signal was fed to the electrode being tested from a Steeper Myoelectric Signal Simulator. This unit, which is no longer available, generates a simulated myosignal which has proven to work well for diagnosing prostheses. For each data point, the input signal was first calibrated by feeding it to an Otto Bock 757M5 Myotester. While this unit is known to give readings that vary somewhat from the most accurate possible electronic instrumentation, it generates numbers that everyone in the myoelectric control field is familiar with. Electrode outputs were measured using a Hewlett Packard 3456A Digital Voltmeter. The greatest source of error was the

setting of the input voltage especially at low levels in the 1 to 10 μV range. The electrodes evaluated were the Otto Bock 13E125 and the RSL Steeper SC01 (D12839). They were operated with an input of 6.0V from a power supply or from the internal voltage regulator when attached to a VariGripII.

What an ideal electrode looks like

To understand the data collected from the Bock and Steeper electrodes it is easiest to establish a working vocabulary by first describing an ideal electrode. One aim is to use these electrodes as sensors for the Liberty Technology VariGripII™

Controller. The analog-to-digital converter in this unit divides 6V into 1024 equal pieces giving a minimum signal of 5.9mV. There are many instances where a maximum signal of only 5mV has been used to operate proportional prostheses. Ideally the patient should be able to "divide" the space between 0 and 5μV into at least 10 parts. Thus the 0.5μV signal should produce a minimum of 10μV out of the electrode at a reasonable electrode gain setting.

Figure 1 shows input-output plots for an ideal electrode with six gain settings. Each plot is a straight line passing through the point zero, zero. As the gain setting increases the slope or steepness of the lines also increases. Since this is an ideal electrode, the change in this slope is uniform from one setting to the next. You can calculate the gains; 0.8V per 100 mV is a gain of 8,000. The other gains are 16, 24, 32, 40 and 48 thousand.

Figure 1. An Ideal Electrode.

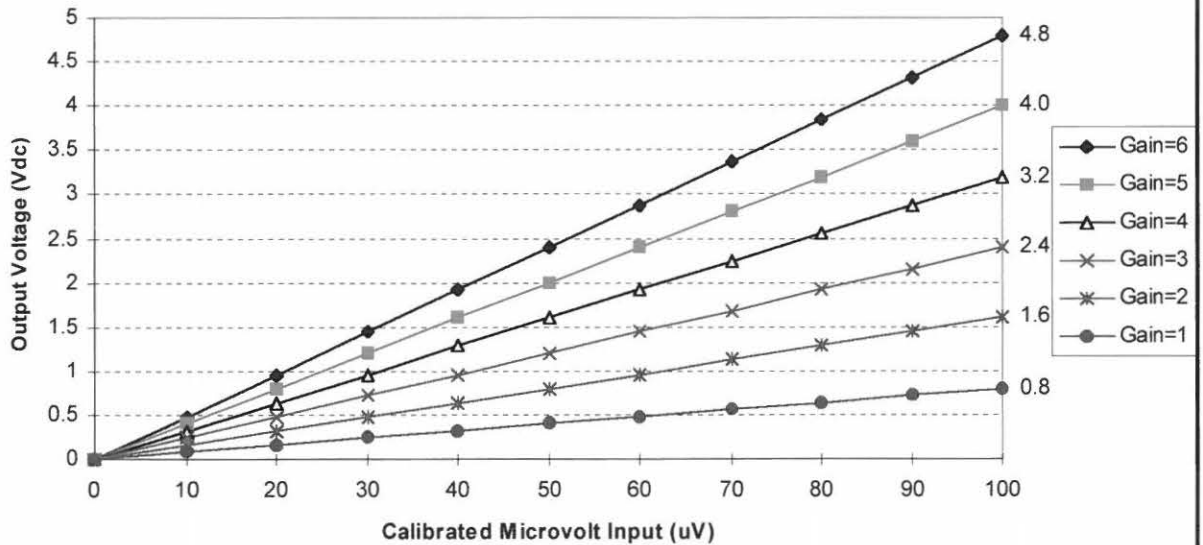


Figure 2. The "Knee" Problem.

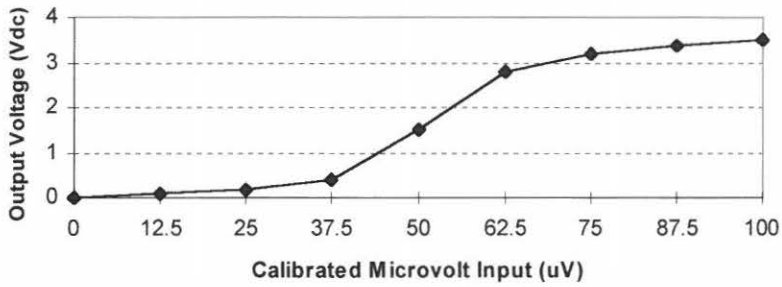


Figure 3. The Offset Problem.

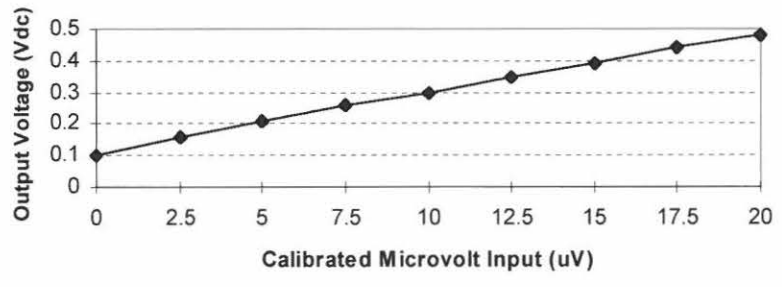
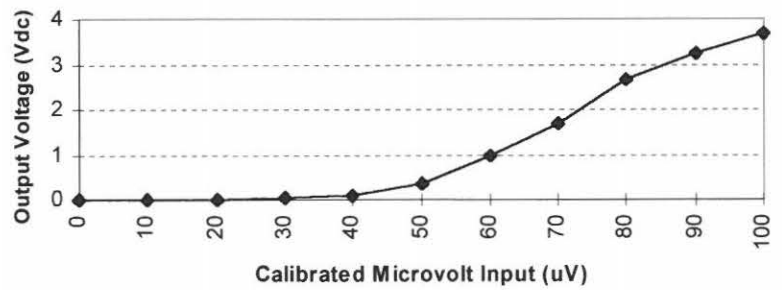


Figure 4. Low Input Failure.

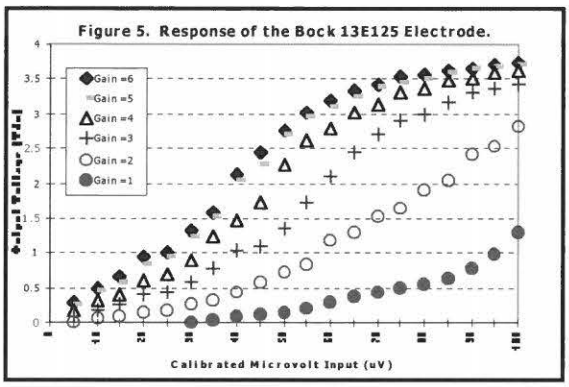


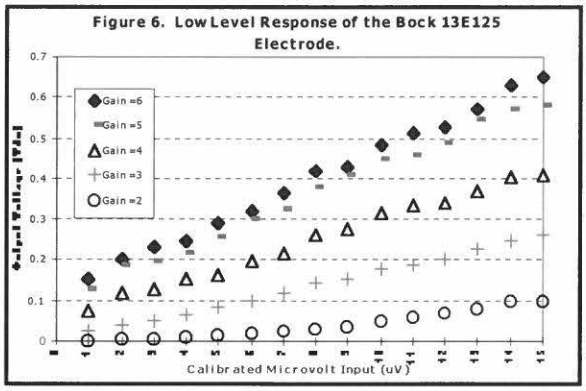
The less than ideal electrode

Figure 2 shows a typical problem electrode. The response curve has two knees, one at 37μV and one at 63μV. These sudden changes in the gain make it difficult for a patient to get good proportional control. On one side of the lower knee the gain is 1100 and on the other side 9000. The user has to work nine times harder to

change the output signal below the knee.

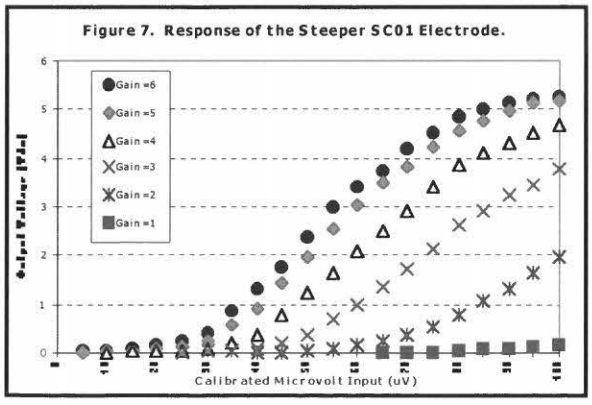
Figure 3 shows the offset problem. This electrode shows an output even with no input. With the MyoMicro software offsets can be cancelled out by giving the electrode a suitable "noise floor" setting. Figure 4 shows another problem. The electrode is essentially useless for inputs less than 40μV, and there is no way to get information from low levels of myoelectric activity.

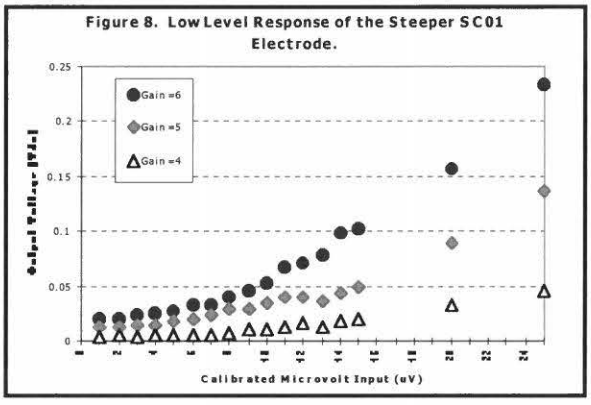




The Bock 13E125 data

Figure 5 shows the response of the Bock electrode from zero to 100μV while Figure 6 shows only the region from zero to 15μV. This electrode is not ideal, but it is good enough for all practical purposes. There is some knee effect. For instance, the low end of the G=3 curve has a gain of 16,000 while it is 60,000 at higher levels. The gain then decreases at still higher levels. However, there are no sharp changes in gain and we are not aware of any clinical feedback saying that the response is not acceptable. The low level response of the electrode is of particular interest. The response for a gain setting of 1 has become too small to measure on this graph, but the other plots show a uniform change of gain as one goes from G=2 to G=5. The actual gains are 7,000 16,400, 23,800 and 32,300. They are almost perfectly spaced 8000 apart. There is practically no improvement in gain above G=5. Finally, this electrode has considerable offset and the offset changes with gain. However, the offset can be removed if a deadband or noise floor is used. This is standard procedure with MyoMicro software and the VariGrip II Controller.





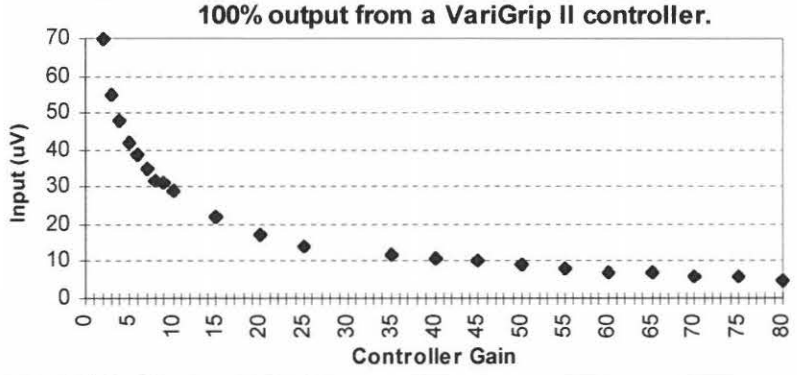
The Steeper SC01 data

An obvious deficiency in Figure 7, the response of the Steeper SC01 electrode, is the knee effect. If one extends the relatively-straight-line sections of the plots down to zero, one sees the following. G=1 is essentially useless; G=2 is useful from 55 to 100μV; G=3 from 40 to 100μV; G=4 from 30 to 100μV; G=5 from 20 to 100 μV; and 6 from 10 to 100μV. Figure 8 shows that only G=5 or G=6 might be useful in the 1 to 5 μV region. The actual gains for the G=4 to G=6 curves were 1700 at 14μV for G=4, 3800 at 11μV for G=5 and 5700 at G=6. This should be compared to the low level gains of 16 to 24 thousand for a comparable region of the Bock plot. The Steeper electrode also shows an offset voltage at low levels.

Selecting the right system gain.

Since the purpose of this study was to qualify electrodes for use with the VariGrip II Control System, it made sense to study how much system gain was needed with various levels of patient maximal signal. Figure 9 shows the result of setting the Bock electrode at G=3 and then adjusting the gain of the VariGrip Controller so that a given μV input would give a 100% signal to the motor controller. The chart can be used to select an appropriate gain if the patient signal is known, or it can be used to calibrate the patient input in μV if the gain for easily achieving a 100% signal is known.

Figure 9. Bock electrode input required to give 100% output from a VariGrip II controller.



Myoelectric control with weak muscles.

It is the nature of the myoelectric signal that it varies in a random way that requires some level of filtering when going from the raw myoelectric signal to a DC control voltage. The need for filtering is most apparent at low levels where only a few muscle fibers are producing the signal. The Bock and Steeper electrodes were tested at low levels with a live subject to evaluate the usefulness of the low level signals. The Bock electrode was set at 3 and the Steeper at 5.5 to produce similar outputs. With a gain of 55 programmed into the VariGrip II controller, the Bock electrode produced a jittery control signal. The Steeper electrode required a gain of 70 for a comparable signal. For practical use, both signals require the setting of some noise floor to reject inadvertent signals and noise. Without further filtering, the jittery signals degrade proportional performance. The MyoMicro program has recently been modified by the addition of a filter to address this problem. There was not time to evaluate the filter before writing this report.

CLINICAL RESULTS

The VariGrip II Controller has now been applied to two patients using system gains of about 55 and a Bock electrode gain setting of 3. These fittings are successful and the patients are well pleased. This makes it likely that more low level patients will be fitted with this system using the familiar Bock electrode. Similar success is possible with the Steeper electrode, but only at the highest gain setting. An additional benefit to setting the Bock electrode gain to G=3 is that the prosthetist can then make minor gain adjustments in the field by increasing or decreasing the gain at the electrode.

NOTES

NOTES

FRIDAY, AUGUST 27, 1999

8:45 am — *Daily Notices*

9:00 am — **Special Guest Speaker:**
Adele Fifield,
War Amputations of Canada

10:00 to 10:30 am — *Refreshment Break*

10:30 am to 12:10 pm — **Programmable Controllers:**

Programmable Control: Technical Aspects
I. Kurtz, W. Heim, H. Bauer-Hume, S. Hubbard, and S. Ramdial

Programmable Control: Clinical Experience at Bloorview-MacMillan Centre
I. Kurtz, W. Heim, H. Bauer-Hume, S. Hubbard, and S. Ramdial

New Clinically-Useful Multi-Device Control Strategies Made Possible by the VariGrip II Controller
C. Wallace and T.W. Williams, III

Clinical Experience with Programmable Controllers
G. Stevens and T. Farnsworth

12:10 to 1:30 pm — *Lunch*

1:30 to 2:45 pm — **High Level Fittings:**

Fourquarter Amputation: A Self-Suspending Shoulder Cap
D. Allen and W.G. Dykes

Adaptation of Locking Shoulder Joints to Increase Functional Range of Motion for Bilateral Upper Limb Deficiencies
G. Stevens and T. Farnsworth

MicroFrame Interface Design for High Level Myoelectric Prostheses
J. M. Miguelez

2:45 to 3:15 pm — *Refreshment Break*

3:40 to 4:55 pm — **Vendors Forum:**
This last session of MEC '99 will be devoted to any vendor that wishes to talk, show or comment on any aspect of prosthetic components

Participation by representatives from Hugh Steeper, Otto Bock, Liberty Technology, VASI, and Motion Control are welcomed.

5:15 pm — *Shuttle Bus from UNB to Sheraton*

THE WAR AMPUTATIONS OF CANADA ADDRESSING THE NEEDS OF CANADIAN AMPUTEES

Adele Fifield, CAE
Director, National Amputee Centre

I lost a leg to cancer at the age of 13. I became a member of The War Amps Child Amputee Program, or CHAMP as it is better known, shortly thereafter. For the last 10 years I have worked with The War Amps and am presently its Director of the National Amputee Centre and Director of Prosthetics and Counselling. In that time, I have attended over a dozen prosthetic conferences to keep up-to-date on artificial limbs and to pass that information on to Canadian amputees.

The War Amputations of Canada is a charitable organization that has been in existence for over 80 years and I will share with you an overview of our history and what we do.

HISTORY OF THE WAR AMPS

The Association was founded by national charter in 1920 as The Amputations Association of The Great War, a fraternal society to provide direction for its members while also seeing to their needs. Counselling, self-help and practical assistance were emphasized.

The organization's first President was Lieutenant Colonel Sidney Lambert, an army padre who had lost his leg in action during World War I. He helped galvanize his comrades in the early years and later spearheaded a drive to set up a sheltered workshop where war amputees could work for competitive wages, while providing a service that would generate funds for the organization. The name of the association was changed to The War Amputations of Canada, and as disabled veterans returned from World War II, the organization provided information, fellowship and employment opportunities to all of its members.

In 1946 the Key Tag Service was launched as a means of providing employment for war amputees. Since then the service has grown into a computerized program that has returned nearly a million sets of lost keys to their owners, but always is dedicated to providing employment for Canadian amputees and people with other disabilities. The War Amps has been able to initiate many programs over the years due to public donations to the Key Tag Service.

The War Amps has always had as its hallmark the philosophy of "amputees helping amputees." Because the Canadian public was so supportive, in 1953 The War Amps expanded its scope of assistance to "civilian" amputees in order that war amputees could share their knowledge with others who are missing limbs from causes other than war. Later, the program was divided into two, one for adults and a separate program for children.

PROGRAMS FOR ADULT AMPUTEES

The War Amps has two programs to assist adult amputees in Canada.

CHAMP Adult Program

When child amputees enroll in The War Amps CHAMP Program, which I will discuss later, a "lifetime commitment" is made to assist them. When Champs turn 18 or have finished their post-secondary education they become members of the CHAMP Adult Program. Through the CHAMP Adult Program, financial assistance is provided for artificial limbs (including repairs) - the remaining cost after funding from provincial plans or private insurance has been accessed - and limited funding for recreational fittings. Financial assistance is also available to cover travel costs to/from prosthetic centres.

Adult Prosthetics Program

The War Amps also provides assistance to thousands of other Canadian amputees who register in the Adult Prosthetics Program as adults. The Adult Prosthetics Program covers 20% of the total cost of a new limb, to a maximum of $2,000, every three years. We recognize, however, that a new amputee requires both a temporary fitting followed by a definitive fitting months later and we provide funding for both. These guidelines allow us to provide some assistance to as many adults as we can. As a charitable organization, available funding under personal insurance, social assistance or some other program, must be accessed before requesting assistance from The War Amps. We also provide information on prosthetics and amputation.

CHAMP PROGRAM

Cliff Chadderton, who lost his right leg during World War II, became the Chief Executive Officer of the organization in 1965. For more than 30 years he has tirelessly served the needs of Canadian amputees, both young and old. Under Cliff Chadderton, the organization grew dramatically and has become known around the world for its innovative programs and ideas. Although veterans' issues are still today a large part of the work of the CEO, Cliff Chadderton realized many years ago that war amputees were being well served by existing programs and that in the future their needs would decrease. So in 1975 he turned the organization's attention to child amputees and in 1975 started the Child Amputee (CHAMP) Program. Children missing a limb at birth, for medical reasons, or because of an accident receive the help they need to live full and active lives. Based on principles of fostering positive, winning attitudes, CHAMP offers child amputees specialized assistance - financial as well as emotional. The organization tries to reach all amputee children and their families as soon after an amputation as possible, providing artificial limbs, information and counselling to help the children and their families cope with their amputations.

The services and programs provided through CHAMP are extensive.

On a practical level, CHAMP provides financial assistance for artificial limbs and related expenses not covered by government or private health plans. Child amputees outgrow their artificial limbs faster than their clothing! They also need specialty limbs and devices which allow them to participate in recreational activities at school, in the community, and within the family.

On a personal level, CHAMP provides emotional support for youngsters and family members alike through a strong network of supportive counselling programs and regional seminars.

CHAMP Seminars

Regional CHAMP Seminars, held annually across Canada, bring together child amputees and their parents to meet their peers and learn about artificial limbs and new developments. They also derive great benefit from interacting with one another. In a country as expansive as Canada, many children live in communities where they know no other amputees, so can often feel alone - a CHAMP Seminar is often the first opportunity many of them have to meet other child amputees.

Each Seminar offers sessions in numerous areas:

* Parents gain support from others who have "been there" and CEO Cliff Chadderton shares his over 50 years of experience in handling situations they face in raising their amputee children.
* Junior Counsellors are older Champs who offer support to younger Champs and parents, providing practical advice and personal insight into growing up with amputation. They are natural role models.
* Teens helping teens - Champs in their teenage years have very different concerns. Sessions for older Champs provide a setting to discuss topics such as dating, driving and career opportunities.
* What's new in artificial limbs? Families at seminars learn about the latest developments in artificial limbs. Champs participate by proudly demonstrating their own artificial limbs, including special artificial limbs for activities such as swimming, skiing or playing the violin!

* Super Champ have unique needs - special seminars address the unique needs of Champs with multiple amputations.

Offshoot Programs of CHAMP

The War Amps also has developed offshoot programs within CHAMP.

In the 80s, The War Amps initiated a program called MATCHING MOTHERS whose theme is "parents helping parents." The Program matches parents of an amputee child with another "more experienced" family within CHAMP. The name of the Program exists for two reasons - MATCHING MOTHERS was the idea of a group of CHAMP *mothers* themselves who expressed how their own anxieties about their child's amputation would have been alleviated by early contact with a parent who had "been there." Plus, although more and more fathers are requesting matches, most requests through the Program are still for mothers. The program is, however, there to assist the whole family. Similarly, we have a JUNIOR COUNSELLING Program where older members of CHAMP, now in their teens and young adulthood, provide the peer support to new amputees or those going through a difficult time. CHAMP parents and the amputees who volunteer in our support programs are **not** professional counsellors. Their role is to share their own experiences and insights and to simply offer support. Requests for matches may come from families themselves or a doctor, nurse or social worker.

Numerous safety programs were initiated after it was noticed that many children enrolling in CHAMP had lost limbs in accidents. PLAYSAFE is aimed at cautioning Canadian youngsters of the dangers while at play and is a *kids-to-kids* approach to child safety awareness. DRIVESAFE was developed to promote safe driving and to prevent serious injuries due to accidents, and SAFETY WALK encourages parents to take their children on a walk to inspect potentially dangerous sites in their neighbourhood.

The JUMPSTART Program provides computers and computer training to Champs with multiple amputations. CEO Cliff Chadderton describes JUMPSTART as "the great equalizer." JUMPSTART allows children with multiple amputations to keep up with their peers at school and opens doors to future career opportunities. JUMPSTART focuses on the benefits of computers and early computer training, with the ultimate goal being future employment and independence for children missing multiple limbs.

The CHAMP JUMPSTART Training Facility is a specialized computer training workshop that provides excellent employment opportunities for amputees. Courses are individually tailored for each student in the different facets of computer work, classroom and practical experience.

WHY SUCH A COMPREHENSIVE PROGRAM?

The value of such a comprehensive program that addresses all the needs of an amputee is reflected in my personal story. As mentioned previously, I lost my leg to cancer as a young teenager. In the first moments after my doctor told me what was happening - once the few moments of utter disbelief wore off and I realized it was not a terribly bad joke - my greatest fears surfaced in the questions I immediately asked the doctor: *Am I going to die? Will I ever walk again? Will anyone still like me?*

Several issues are at the forefront here. It is a *physical, social and emotional* roller coaster on which amputees find themselves. When working with amputees at any stage of rehabilitation, you cannot treat one aspect of their rehabilitation and ignore the others and expect great results. In some cases the amputee's life has been threatened and dealing with that is of primary importance. Once an amputee's physical survival is assured (as much as it possibly can be), then his/her social and emotional needs must be met. Amputees want to become functional again and to take part in society as they used to (or as much as possible), and must be given realistic expectations in regards to prosthetic fittings, recognizing, of course, that the individual's motivation and perseverance plays an integral role in the results. While dealing with the fitting process, those working with an amputee must recognize that

strong emotions are at play as the individual tries to come to terms with an altered body image and fears of rejection by loved ones.

Oftentimes, amputees themselves are the ones who project their own fears onto others, sometimes without realizing it, and thus close themselves off to close relationships. Later when relationships fail or do not begin at all, the failure may be blamed on the amputation when in fact signals were being given out that the amputee was not approachable - creating a vicious circle. It is obvious that if amputees are struggling with emotional issues, these will be reflected in how they cope with other aspects of the rehabilitation process like prosthetic fittings. In short, as you have likely heard many times, do not forget the "whole" person you are working with, and ensure all those who can help an amputee are brought together in a "team" approach. Be careful though not to overwhelm the amputee - I have heard from many new amputees that they are often intimidated and fearful when they enter a clinic situation and are placed in a room with half a dozen professionals who use terminology they cannot understand.

In order to address the needs of Canadian amputees, The War Amps is continuously creating new programs and resources.

Most recently, to address the needs of teenage amputees we produced a series of *TEEN TALK* videos covering the topics of: the school years, relationships and body image, driving and employment. We have placed the series on one two-hour video tape that is used in part at our seminars, but is also available to members on loan.

Another resource that we created to address the needs of very young child amputees who are entering their school years is a Starting School Kit - this kit contains guidelines and suggestions for parents to actually go into the school or daycare to introduce the child to his playmates or classmates in a very positive way, and to provide an opportunity for the other children to ask questions about the child's amputation or artificial limb. Our families have high praise for the kit. The kit also includes two short counselling videos - a four minute cartoon called *Kinetic Prosthetics* featuring an amputee character named Fleetwood and a 15-minute puppet video, *Just The Way I Am,* featuring amputees, Luke and Michele, showing how they handle situations at school.

Plans are always ongoing for additional resources to be introduced.

THE FUTURE

The organization has put the groundwork in place for the time when war amputees will no longer be able to run the affairs of the Association. The Association plans to carry on under the name The War Amps. Through Operation Legacy, former members of CHAMP who are now grown will continue The War Amps' tradition of "amputees helping amputees." The War Amps will continue on into the future to provide many valuable services to its amputee members and the Canadian public.

PROGRAMMABLE CONTROL: CLINICAL EXPERIENCE AT BLOORVIEW MACMILLAN CENTRE

Isaac Kurtz, Winfried Heim, Heidi Bauer-Hume, Sheila Hubbard, Sandra Ramdial
Bloorview MacMillan Centre, Toronto, Ontario

INTRODUCTION

Advances in microprocessor technology in recent years have led to the introduction of programmable control systems for powered prosthetics. These systems allow amputees to try a variety of control schemes and choose the one that suits them best. Prosthetists, no longer limited to preprogrammed control schemes, can devise new schemes that are suited for the amputee's individual needs. Over the past few years, Bloorview MacMillan Centre has fit approximately 20 clients with programmable control systems. A retrospective analysis of this group, which includes amputation levels from below-elbow to shoulder disarticulation, demonstrates the benefits this approach. The benefits fall into four general categories: 1) evolution of the control system as the user's needs and abilities change, 2) the amputee's ability to choose their own preferred strategy, 3) accommodation of abnormal and noisy signals and 4) ability to accommodate high-level amputees. This paper will summarize our clinical experience with programmable control. Case studies illustrating this approach and its various benefits will be presented.

BACKGROUND

Developments in microprocessor technology in recent years are having a significant impact on the delivery of prosthetic systems. Microprocessors have been designed for battery operation at a wide range of voltages using little power. These processors are available in very small packages and incorporate all of the features needed for prosthetic control. Many microprocessors are available with on-chip field-programmable memory. It is the field-programmable feature of these chips that allows them to be adapted to the needs of a wide range of amputees.

Manufacturers of prosthetic hardware have taken advantage of these developments by creating prosthetic controllers that can be programmed and customized by prosthetists in the field. Animated Prosthetics, Liberty Technology, Motion Control and Variety Ability Systems Incorporated (VASI) have all introduced field programmable prosthetic controllers.

The powered upper extremity prosthetics team at Bloorview MacMillan Centre in Toronto has had extensive experience using MyoMicro, a graphical software programming tool that is compatible with the Liberty and VASI controllers to meet the diverse needs of a wide range of clients. This experience has confirmed the versatility of the microprocessor-based programmable approach.

CLINICAL EXPERIENCE

Evolving control systems

Choosing a control system for an amputee's prosthesis involves assessing the amputee's functional goals and needs, ability to control multiple myoelectric and mechanical inputs, tolerance of componentry and cost. Regardless of how thorough the clinical team is in making this assessment, the above factors are likely to change over time. Programmable control allows the clinical team to change the control system as the needs and abilities of the amputee change without incurring the cost purchasing and installing a new control system.

A common example of control system evolution is the clinical practice of using a voluntary-opening or "cookie crusher" control system for congenital amputees under 3 years of age and switching to 2-site myoelectric control when the child is ready [1]. Currently, this switch is done by purchasing new hardware for the child's prosthesis. Using a programmable controller, the change in control system can be done without changing the hardware, by reprogramming the prosthetic controller. Control system evolution is also common in the case of traumatic amputees whose tolerance for complex and heavy componentry increases as they become accustomed to their injury and to wearing a prosthesis. This is illustrated in the following case from Bloorview MacMillan Centre's recent clinical experience.

Case 1

The client is a 16 year old female, competitive figure skater who suffered a traumatic, very short AE amputation as a result of a boating accident. She was initially fitted with a lightweight passive cosmetic prosthesis. The initial goal was to condition her to wearing a prosthesis and accommodate her short-term need for cosmesis to be a bridesmaid.

Upon assessment for an active prosthesis only one weak EMG signal site on the deltoids was found but there was sufficient humeral flexion and extension to operate forse sensitive resistors (FSR's). The VASI programmable controller was programmed to pick up signals from 2 FSR's mounted in a check socket. The family was anxious to have an electric elbow incorporated but the clinical team felt it was not advisable for the first fitting since the client is very petite and still emotionally fragile. The team decided to use programmable control now so that the elbow option could be added at a later time if she is able to handle the additional weight and control system complexity.

Implementing control strategies that are not commercially available

Often clients with special needs require customized control systems because they have difficulty controlling commercially available systems. In the past, this type of customization was expensive and generally avoided by the clinical team. With programmable control, the prosthetic team can create virtually any control scheme that suits a client's unique abilities.

The Bloorview MacMillan Centre powered prosthetics team has taken advantage of programmable control for clients who had been fitted many years earlier with control systems that are no longer commercially available. These clients have become accustomed to their prostheses since early childhood and changing the control system would have resulted in degradation of control accuracy. This is illustrated in the following case.

Case 2

The client was a 31 year old male with a congenital short below-elbow deficiency, employed full-time as a forklift operator. He used a conventional, body-powered prosthesis until age 13 and was admitted in 1982 for myoelectric fitting. In spite of prolonged assessment and control training , it was not possible to overcome a muscle co-contraction problem. The client was eventually fitted with a Variety Village version of the UNB 3-state system and became a successful, full-timer user of his myoelectric prosthesis.

In May of 1997, he required a replacement of his myoelectric prosthesis but the UNB system was no longer available. Programmable control was used to assess his ability and his preference between rate and level sensitive strategies. The client preferred to continue to use a level-sensitive approach and the programmable controller was configured to emulate the system to which he was accustomed.

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1 Hubbard, S., Bush, G., Kurtz, I., & Naumann, S. "Myoelectric Prostheses for the Limb-Deficient Child", Physical Medicine and Rehabilitation Clinics of North America, 2:4, 847-866, November, 1991.

Overcoming interference problems

Interference, or noise, poses a significant challenge to successful myoelectric fitting. Interference may come from other muscles in the client's residual limb, external electrical sources and inadvertent cocontraction of the controlling muscle. Programmable control opens the possibility for a number of innovative ways to overcome interference in a myoelectric system. For example, digital filters and time delays can be introduced in the signal path to make the system more selective to the desired signals. The control logic can also be changed. For example, the system can be programmed to ignore a muscle signal if antagonist muscles are active at the same time. Alternatively, the system can react to the first muscle to contract and ignore the second muscle. In some cases, cocontraction itself is the most reliable way for users to communicate their intentions to the system. The following case illustrates the use of programmable control to overcome electrical interference due to an amputee's pacemaker.

Case 3

A male War veteran sustained an amputation above the left elbow. His arm was blown off in battle and severe tissue damage resulted. Burned tissue and skin grafting cover the residual limb and part of his upper body. This patient's circumstances are also complicated by the fact that he wears a pacemaker. Upon the initial assessment by the team, it appeared that he had adequate biceps and triceps control to operate a myoelectric prosthesis. A closer examination revealed otherwise. A strong biceps muscle signal was obtained only when the muscle gained a fair size muscle belly. The muscle would not achieve the same signal strength when the development of its muscle belly was hindered, such as is the case when containing the residual limb in a prosthetic socket. In addition to the fitting difficulties associated with stump pain and flaccid tissue, a major problem arose when interference, generated by his pacemaker, was picked up by the electrodes in a conventional two-site myoelectric control system.

In order to fit this man with a 2-site myoelectric prosthesis controlling an electric elbow and hand, a programmable control system was used. Through the Myomicro software, the electrode gains were altered to enable the muscle signal fluctuation of the biceps to be picked up when it was not able to gain its full muscle belly while contained in the prosthetic socket. The pacemaker signal problem was addressed by implementing time-delay filters in the input signal path. A biceps – triceps co-contraction selects the device to be activated with an automatic revert to hand default.

High-level amputees

The most challenging powered upper extremity fittings involve amputations above the elbow and higher. These clients have the most complex needs but they often have the fewest sites for control. The process of choosing the control strategy that is most appropriate for these clients and optimizing the control parameters for them is greatly enhanced by the programmable control approach that allows clients to try a variety of options and refine the one that best suits them. The following case illustrates the usefulness of programmable control for a high-level client.

Case 4

The client is a 46 year old woman who suffered a complete transverse amputation of her left arm. She was very motivated to attain restoration of bimanual function in order to resume her duties as a general physician in a busy family practice. Cosmesis and function were equally important and our goal was to provide her with a prosthesis which would optimize function while appearing as natural as possible both statically and dynamically.

The clinic team along with the client, decided that a prosthesis with a passive shoulder , an electric elbow and electric hand would best serve her needs. The programmable controller was included to give us the flexibilty to design the most effective control strategy possible. The major problem was a limited number of control sites, due in part, to her distorted proprioception in the shoulder region. We were able to identify only 3 distinct shoulder

movements; protraction, retraction and elevation. Using the MyoMicro controller we were able to explore various control options including rate and level sensitive 3-state control. It was finally determined that the best strategy for this client was a control system that permitted mode selection between the hand and elbow control.

In the client's prosthesis, two Force Sensing Resistors are used as input devices and were chosen because they are low profile and offer proportionality. The FSR's are positioned on the socket at specific points anterior and posterior to the acromion process. Protraction of the shoulder controls hand closing and elbow flexion, while retraction of the shoulder controls hand opening and elbow extension. By using the MyoMicro software, the sensitivity levels of each FSR can be independently adjusted to refine control.

Attempts to position the FSR's to allow shoulder elevation to activate both devices simultaneously as a mode selection signal were abandoned due to the difficulty of this approach for the client. Instead, the mode selection function is provided by a microswitch positioned over the acromion process and accessed by the third movement, shoulder elevation. To avoid mode switching due to inadvertent activation of the microswitch a time delay filter was introduced. The controller's capacity to fine tune the system resulted in a prosthesis which satisfies the client's need for subtle movements for hand and elbow control.

NEW CLINICALLY-USEFUL CONTROL STRATEGIES MADE POSSIBLE BY THE VARIGRIP II™ MULTI-DEVICE CONTROLLER

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ABSTRACT

The VariGrip II™ Controller can operate up to three variable speed motors. For inputs it uses two, three, or four variable voltages and one on-off switch. This makes a large number of control strategies available, but not all are clinically useful. To date the most popular controls have used two myoelectrodes to control two or three devices. The controller has been used with the two VASI elbows and the Hosmer and Boston elbows to solve clinical problems that were heretofore intractable. The system also permits proportional control of two-motor devices such as the Bock 2000 hands, the Steeper Gripper, and the Centri UltraLite hand. We will discuss a number of the strategies that have proven clinically useful.

Many devices can be controlled.

Since the VariGrip II system is designed to provide direct proportional control of electric motors, it follows that the devices controlled should be the manufacturer's device without any electronic circuit. Listed below you will find the one and two motor devices that have been successfully controlled. In the lists below the synergetic prehensor has two motors, but they are wired in parallel and drive like one. The VASI hands are also available with the controller built into the hand when no other device is to be controlled.

1-Motor Gripping Devices
Bock 8E37 hands
Bock 8E32 greifer
Steeper series 200 hands
Hosmer Michigan Electric Hook
Hosmer NU-VA Synergetic Prehensor
NYU Hosmer Prehension Actuator
VASI Linear Actuator
VASI Childrens electric hands
Bock SUVA hand

2-Motor Gripping Devices
Centri UltraLite Hands
Steeper Powered Gripper
Bock 2000 Hands

Wrist Rotators
Bock 10517 Electric Wrist Rotator
VASI VR Series Electric Wrist Rotators

Elbows
VASI 38 Small Childs Elbow
VASI 812 Large Childs Elbow
Boston Elbow II
NYU-Hosmer Electric Elbows

Other Devices
VASI Linear Actuator
Steeper Electric Lock Elbow
Liberty Electric Actuator for Shoulder

New Single-Input Capabilities.

A patient with a single good myoelectric site has always been a challenge. A single poor site is even worse. The VariGrip II system programmed with Bloorview-MacMillan's MyoMicro™ program offers new clinical choices.

The most popular choices for one-muscle control are the *cookie crusher* and *quick-slow* strategies. A third choice, the *UNB dual-threshold* or *level* control which is no longer available except as offered in the MyoMicro system, gives fixed speed control only. Each strategy has its limitations. *Cookie crusher* is perfect for a toddler, but its full-force closure is unsuited to an adult user. The Bock quick-slow circuit cannot be adjusted to suit the user, and finally the UNB *level* control only permits fixed speed operation. These three strategies are all offered with MyoMicro with the added advantage that delay times and other parameters can be [set for] the particular patient when the default value is not suitable. Further, the VariGrip II and VASI proportional hand circuits permit setting the extra gain that will accommodate a user who generates as little as a 5μV maximal contraction.

Because even the three strategies above may not suit all patients, a new strategy has been added to the MyoMicro system called *Alternate*. With this strategy every other contraction closes the hand while alternate contractions open it. A default can be set to always make the next contraction open the hand after a waiting period. Several practitioners have used *alternate* with their patients either alone or in combination with more complex strategies.

Differential control

Differential control has been used by Liberty and Motion Control for over thirty years. It is the default control scheme with MyoMicro. Simply, differential control subtracts one muscle signal from the other and uses the resultant signal to control motion. To work well, differential control requires the rejection of inadvertent signals from either muscle using a "dead band" or "noise floor" setting. Thereafter, the strength of the two signals is adjusted until the user feels that generation of the two signals requires equal effort. Differential control is recommended whenever a weak muscle can only generate a signal with some cocontraction of the agonist present. With suitable gain adjustments the inadvertent cocontraction can be ignored using differential control.

New mode selection schemes

To use two myoelectric sites to control more than one device requires a selection strategy. Mode selection was introduced by Motion Control with the Utah Artificial Arm and later with the ProControl system. Liberty Technology has also made extensive use of mode selection with the Boston Elbow and now with the VariGrip II. With analog circuitry all of these schemes were limited to what could be easily built into a single circuit. Software control permits the construction of new ways to change control from one device to another.

The most reliable mode selectors are a rapid cocontraction of two muscles or the use of a switch. Neither scheme will cause inadvertent motion of the prosthesis. With MyoMicro it is easy to set up a scheme where slow contraction of either muscle controls one device while a quick contraction controls the second. One can also use a rapid contraction held for a period of time to initiate change. This can be done with just one muscle. In all cases once a mode is selected one can either wait for a second signal to shift back or one can set a revert time to automatically shift back to the default control.

Unusual clinically-useful strategies

A recent challenge was a patient who had only one usable upper arm muscle. She did not want to use a switch for mode selection. The *quick-slow* scheme was first proven to be good for controlling one device. Then several mode selection schemes were tried. The best scheme turned out to be a rapid, strong contraction sustained for a full second. The key element was "strong." It was necessary to use a level higher than her normal quick-slow control levels. With this scheme, as soon as the signal reaches the "strong" level, it inhibits control of the direction normally selected by "quick." The hardware used was a VASI 8-12 Elbow and a Bock SUVA hand with the VariGrip II controller.

Another use of the VariGrip II has been to add new strategies to the Boston Elbow. The most obvious addition is the use of cocontraction to shift modes. With MyoMicro software the exact type of cocontraction can be tailored to the user. One can use cocontraction to select two or three devices and can also add automatic reversion to the default device. Cocontraction to select modes is now considered a standard offering.

The VariGrip II controller also permits any one-muscle strategy to be used with the Boston Elbow. For instance, a patient was recently fit using *alternate* as the control strategy and a switch for mode selection.

Making setup easy for the practitioner

Setting up a complex control scheme is work best left to people with a background in electronics, computers and control systems. Liberty Technology in the USA, VASI in Canada and a few trained practitioners all have the required expertise. For the average practitioner all that is required is the ability to evaluate the patient and to define the clinical challenge. New challenges will lead to new control schemes. Fortunately, once a scheme has been tried on one patient it is permanently available for use with others. At present the library of control schemes has at least 55 variations in it. An ongoing challenge will be to publish these schemes using vocabulary that is clear to the clinician rather than to the technical people who conceived of them.

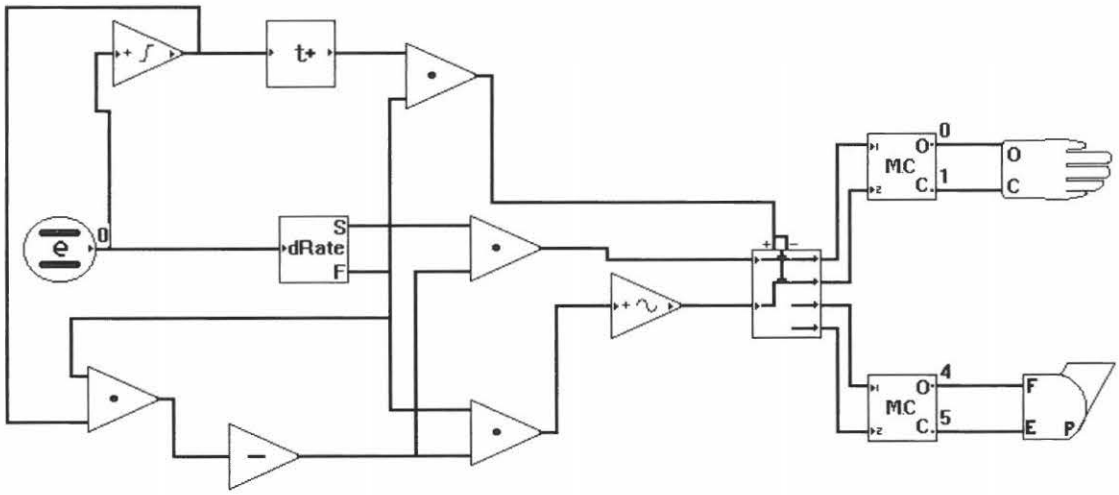

Figure 1. This elaborate MyoMicro screen is used to program the Varigrip II to control both a hand and elbow with one muscle. Mode selection uses a quick, strong, sustained contraction. There is no need for the prosthetist to learn how to set up the scheme. This is the manufacturer's job.

CLINICAL EXPERIENCE WITH PROGRAMABLE CONTROLLERS

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Upper Extremity Prosthetic Program

INTRODUCTION

One of the greatest advancements in powered upper extremity prosthetics has been the advent of "programmable" controllers. These programmable controllers are allowing new operational strategies to be designed and implemented for upper extremity patient populations. Previously, existing technology provided basic functional abilities but continued to limit maximum functional restoration of upper extremity use following amputation and prosthetic fitting. Any additional changes to the "control strategy" of the prosthetic system meant changing hardware with additional costs. This was not a cost-effective treatment protocol for the providers who were reluctant to explore available options. With these limitations, many patients were forced to adapt to decreased functional capabilities with inadequate prosthetic systems or without the assistance of any prosthesis.

Programmable controllers have enhanced the care and functional abilities of the upper extremity amputee by providing progressive adaptation throughout the rehabilitation process. Input, output and control parameters are optimized as functional demands are increased. For example, the system can be initially setup for ease of use and training and later optimized for function as signal strength and abilities increase.

Two patient cases will be presented. Case 1 utilizes a single-site myoelectric control, harness pull switch for mode select controlling an electric elbow, wrist and hand (or Greifer). Case 2 utilizes a single-site myoelectric control, nudge "rocker" switch for mode select to control the electric wrist and hand (or Greifer).

CASE STUDY #1

A thirty-one year old male patient presented with a left mid-transhumeral amputation, secondary to a complete degloving injury. For approximately one year prior to prosthetic evaluation he had multiple revision surgeries including a latissimus dorsi muscle transfer and extensive skin grafting. Secondary to complications from the trauma and surgeries, it was noted the residual limb had extensive scarring and limited range of motion. (Figure 1)

All upper limb prosthetic options were discussed with the patient as they applied to his condition. The prescribing physician, case manager, and occupational therapist were consulted. The rehabilitation team elected to fit the patient with a single-site myoelectric controlled externally powered prosthesis that included: Boston elbow system, Varigrip II programmable controller, Otto Bock electric wrist, and Otto Bock electric hand and Greifer.

At the time of the initial evaluation, the patient was evaluated for myoelectric signal strength and control. A single myoelectric site was located on the anterior surface of the limb at mid- to distal third level. The transferred latissimus dorsi muscle presented with signal strength of approximately 20-30 microvolts, although endurance was limited. Similar signal strengths were also found on the posterior aspect of the residual limb and the remaining portion of the latissimus dorsi still anatomically intact. The anterior site was chosen due to signal strength and skin condition at site. (Figure 2)

The patient received instruction on isometric exercises to strengthen and increase endurance at the proposed myoelectric sites. The patient was also referred to a rehabilitation center for pre-prosthetic training to increase

range of motion and enhance prosthetic fitting potential. As a result, myoelectric signal strengths increased to 40-50 microvolts and endurance and selective control was improved as well. No changes to range of motion were noted. The patient was referred back for an expedited prosthetic fitting procedure.

The patient was fit with a diagnostic prosthesis over a two-day period utilizing a rate-sensitive single site myoelectric system. (Figure 3) Initial fitting included a suction, flexible inner socket, DurrPlex frame, and chest harness with a flexible shoulder saddle. A combination of a slip pull-sock and wet fit were used for donning purposes. The Boston elbow, Otto Bock hand and Greifer were added using a single harness pull switch to mode select between elbow and terminal device function. (Figure 4) Prosthetic training was initiated the following week. Minor skin irritations due to scar tissue were resolved with prosthetic socket adjustments. The inner flexible socket was eventually changed from Bioelastic to Proflex with silicone. Once good control of elbow and hand function was achieved, the strategy within the Varigrip II programmable controller was changed to add an electric wrist rotator.

Stabilization of limb/tissue volume within the diagnostic prosthesis preceded fabrication of the definitive prosthetic design. No interruptions to wearing schedules or prosthetic training were experienced during the fitting and adjustment period. Ongoing outpatient training allows for continued development of prosthetic applications for the patient's lifestyle and activities of daily living.

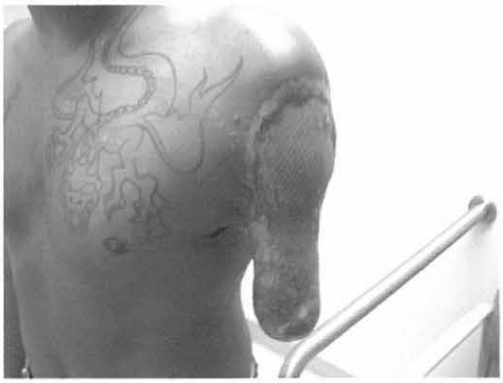

Figure 1. Patients residual limb.

Figure 2. Myoelectric testing.

Figure 3. Patient wearing diagnostic prosthesis.

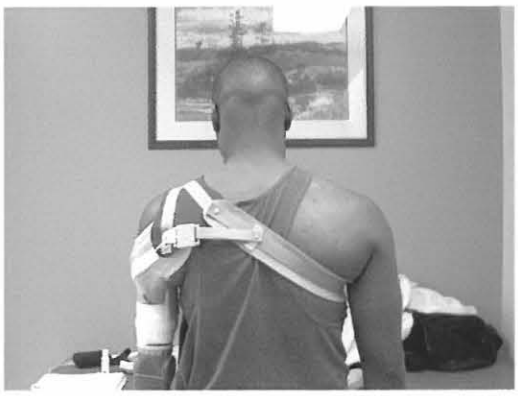

Figure 4. Pull harness switch for mode selecting.

CASE STUDY #2

A forty-one year old male patient presented with a right long transhumeral amputation and a left shoulder disarticulation secondary to electrical trauma. The right side presented with multiple complications from the trauma including limited range of motion, extensive scarring in axilla region, and complete loss of biceps. The patient had undergone three revision surgeries to increase range of motion of the right shoulder and for skin grafting to axilla region. (Figure 7) No complications were present on the left side. Decreased flexibility of torso/spine was also noted during this initial evaluation. (Figure 5)

Current prosthetic designs consisted of a left externally powered cap-style shoulder disarticulation prosthesis with limited function and success. The right side had been unsuccessfully fit with a full myoelectric prosthesis that he had been unable to wear. The potential myoelectric sites selected for use in the current prosthesis were over the triceps and anterior deltoid. All prosthetic options were considered and discussed with the patient and rehabilitation team. Due to the patients limited range of motion and strong desire to utilize a functional hand, a single site hybrid system was proposed for the right side. The left side was refitted with an anatomically contoured frame style socket for increased stability and control, using existing components. An electric unlocking mechanism was added to the left locking shoulder joint to increase range of motion and function. (Figure 6)

During the initial evaluation, the patient was myotested on the right side. No functional anterior muscle group was found and the patient was quite limited in both strength and endurance of the triceps muscle, generating a signal of approximately 5-10 microvolts. The patient was given isometric exercises to strengthen the triceps muscle that improved in strength to 10-30 microvolts over a four-week period. Outpatient therapy was initiated with continued pre-prosthetic training.

The patient was initially fit with a hybrid transhumeral prosthesis with an Otto Bock AFB elbow and an Otto Bock single site system to control hand or Greifer function. A nudge controlled rocker switch was used to control pronation and supination of an electric wrist rotator. The prosthesis is donned utilizing a "slip" pull sock over a custom compression garment. Limited success was achieved with the system due to weak and inconsistent EMG signal strength. The patient could only operate the system during optimum conditions but had difficulty under variable conditions. Other limiting factors were the continuous migration of the nudge control switch with movement, difficulty in reaching the switch in all positions, and the loss of the line of sight as the patient attempted to perform various activities.

The Varigrip II programmable controller replaced the Otto Bock single site system. Initially, the patient was fit using an alternating single site strategy with a nudge rocker switch (single position) for mode selecting between electric terminal devices and wrist rotation. The programmable controller amplified signal strength to allow for operation at such low EMG signals. The program was set to default to terminal device closing function after an adjustable time period of inactivity. The patient wore this system for a three-month period with great success. (Figures 6 and 8)

Once the patient achieved consistent EMG control and success with the prosthetic system, alternate control strategies were investigated to optimize the system. For example, a single site rate-sensitive proportional system was evaluated using the same single site electrode for mode selecting. This was accomplished by using a quick, hard prolonged contraction to mode select. Although the patient was able to operate this design, it required greater attention to operate and proved to be too difficult for the patient to operate. This design will be evaluated for possible implementation in the future care of this patient.

Currently, the controller has been reprogrammed for a rate-sensitive proportional control single site system with rocker switch for mode selection. This system allows the patient to have direct control over open and close

functions without any default mechanism in place. Continued evaluation of the system will occur over the next three months to maximize optimum patient function and abilities. The system will be reprogrammed and optimized accordingly.

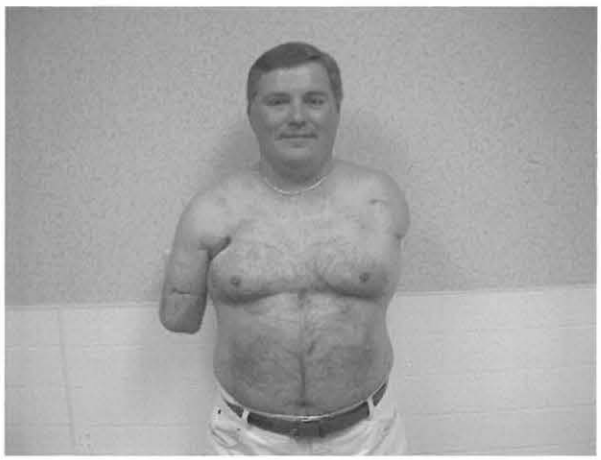

Figure 5. Patient residual limbs.

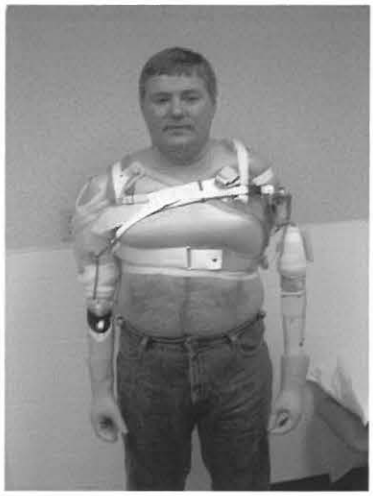

Figure 6. Patient wearing diagnostic prostheses.

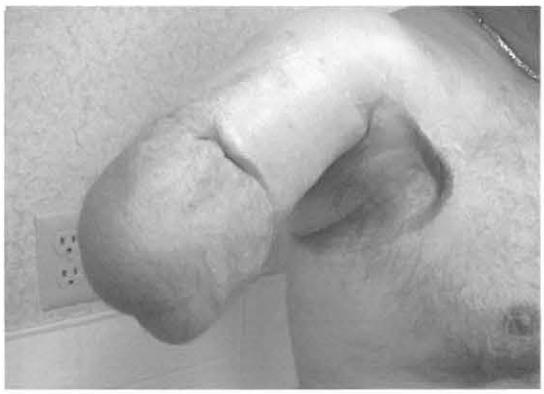

Figure 7. Patient's right residual limb.

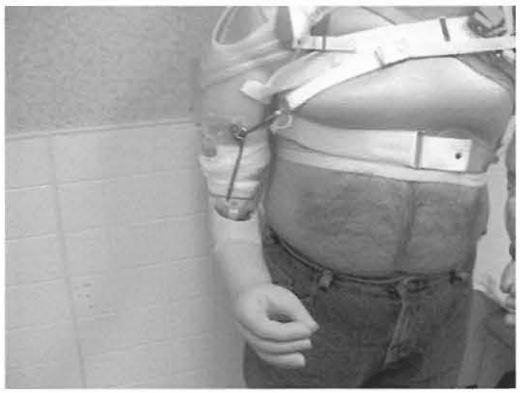

Figure 8. Right diagnostic hybrid system.

DISCUSSION

Programmable controllers can be utilized with most powered prosthetic components and can be easily adapted for varying conditions. The use of programmable controllers requires computer literacy and an extensive learning curve for programming on the part of the prosthetist. It is recommended that the system be initially set up by the distributor of the controller and then optimized by the prosthetist in the clinic.

The system is capable of controlling more than one device simultaneously. For example, a patient might have myoelectric control over a terminal device and switch or servo control of the elbow. Many different input and mode selecting options can be explored to suit the individual needs of the patient. A dual site myoelectric input

controlling elbow, hand and wrist might default to hand function and utilize a single co-contraction to initiate elbow mode while a double co-contraction might be used to initiate wrist operation. Another example of use include the fitting of a post-operative prosthesis in switch or servo mode utilizing the same components to advance to myoelectric control when the patient is ready for a definitive prosthesis. Again, an experienced prosthetist will be best able to maximize the prosthetic system for each patient.

CONCLUSION

Programmable controllers are now allowing more patients to be fit in prosthetic systems that were previously not ideal candidates due to decreased range of motion, limited strength, minimal EMG signals, bilateral deficiencies, and other complications. Many programmable controllers have been successfully fit to date. The control strategies used include switch, servo, and myoelectric inputs controlling multiple motor outputs. Both of the patients presented would not be candidates for myoelectric control of multiple devices without the use of a programmable controller.

In the future, the system will be improved when more successful fittings have occurred and strategies have been developed for use with alternative patient groups. Real-time adjustments and diagnostics will further enhance the use of programmable controllers. Data collection for EMG information, component cycles, motor current and velocity, and feedback systems could also be developed. Finally, remote diagnostics and adjustments could be used to optimize patient performance in off-site locations.

ACKNOWLEDGEMENTS

Special thanks to Craig Wallace and T. Wally Williams of Liberty Technology for their technical support and expertise.

FOREQUARTER AMPUTATION: SELF – SUSPENDING SHOULDER CAP

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For many years, suspension for most levels of upper limb prostheses was by web or nylon straps. Amputees looked upon these straps as a necessary evil but most longed for an alternative. In recent years, self-suspending sockets using either indirect skeletal attachment or suction/skin traction have been developed for trans-humeral and lower levels. Trans-radial levels are particularly suitable for self suspending sockets and at most enlightened prosthetic centres it now a rarity to come across harness suspended prostheses at this level. However, for higher levels, shoulder disarticulation and in particular forequarter amputation the residual stump shape does not lend itself to self-suspending methods.

Forequarter amputation involves complete removal of the arm and shoulder structure leaving an amputation site with little the prosthetist can utilise to assist in suspending the prosthesis. Additional difficulties arise, as no effective body power sources are available to operate prosthetic components. Externally powered components may be used but the additional weight of these components adds additional suspension problems. It can also be difficult to locate enough control sites to effectively operate these components. For these reasons unless extremely well motivated and determined most forequarter amputees resort to a lightweight shoulder cap prosthesis which extends to the axilla level. The sole purpose of this shoulder cap is to restore the patient's shoulder profile and improve their body image. However even this simple device requires suspension straps that extend around the shoulder on the sound side. Some amputees tolerate these straps but for others can be the major factor in rejecting the prosthesis. Comfort is a problem as is hygiene and especially for the female amputee, who may be wearing lighter clothing, it can be impossible to conceal the straps effectively under clothing. In addition, even with lightweight shoulder prosthesis it is often impossible due to the shear angle of the amputation site to prevent socket slippage. The socket may be extended around the base of the neck to the sound side in a halter shape but even this type of socket is not totally stable.

The idea for developing a completely different method of suspension for the forequarter shoulder cap came about after a discussion with Dave Allen of IDS Ltd, Dublin at the annual conference of the British Association of Prosthetists and Orthotists which is usually attended by around 1000 delegates. Dave had experimented with Amoena to assist the retention of a cosmetic partial shoulder restoration pad and found them to be very effective and thought that it might possibly to use the same material to suspend a complete forequarter shoulder cap.

Amoena pads are manufactured by the Coloplast Group for the attachment of breast prostheses. One side of the pad has a coating of medical grade adhesive and the other side loop Velcro.

Against this background and initial trial began at the National Centre's Prosthetic/Orthotic Clinical Unit at the West of Scotland Mobility and Rehabilitation Centre (WESTMARC), Southern General Orthopaedic Hospital, Glasgow.

The first patient fitted was a lady of 35 years of age who had an amputation two years previously as a result of a malignant soft tissue tumour. This lady had initially been provided with a lightweight laminated shoulder cap (fig 1&2) and a full endo-skeletal forequarter prosthesis (fig 3). She worked as a receptionist at a Health Centre and in the first few months would only leave her house if she were wearing her full prosthesis. As her self-confidence grew her preference was to wear the shoulder cap but still complained of the problems of socket slippage and discomfort from the simple retention straps.

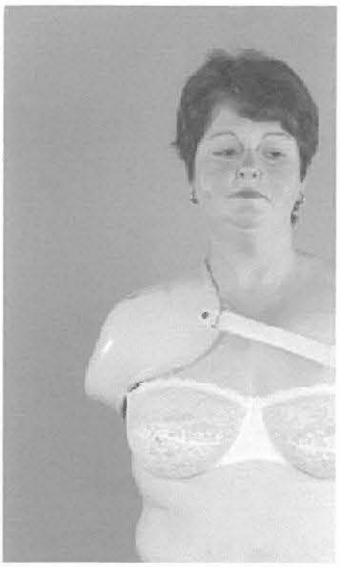

Figure 1

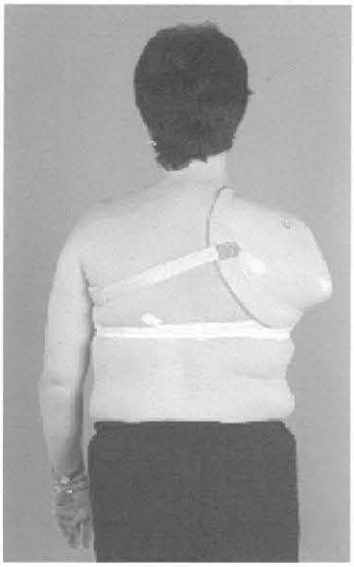

Figure 2

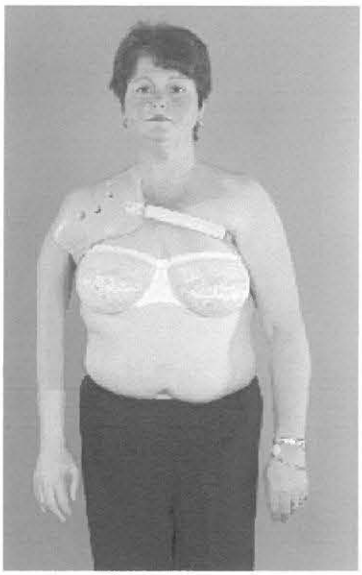

Figure 3

When approached to take part in the trial she was very willing and took the view that anything that might result in the demise of the straps was worth trying. Advice was taken from colleagues in mastectomy clinic at the hospital on the use of Amoena pads. The first stage was to check for any allergic reaction to the adhesive of the Amoena pads. A small patch was applied to the patient's skin and she was asked to check this regularly over a period of a week for adverse reaction. No problems were encountered and a cast was taken to begin a fitting. A shoulder cap was then produced using a custom mould and Otto Bock Pedilen Flexible Foam 150. To ensure a close fit inlays were moulded in to the foam to accommodate the thickness of the pads.

The method proved to be a success with the patient extremely happy with the immediate result. A review was carried out after one week and the patient arrived to report absolutely no problems. Retention was even better than had been anticipated, slippage was completely eliminated and the patient was enthusiastic about the improvement in comfort and cosmetic appearance without straps. She also felt the prosthesis to be lighter than previous models.

Initially we had been slightly concerned that adhesion of the pads (fig 4) might be a problem as the patient seemed to readily perspire. Our fears were unfounded as even with a daily shower the Amoena skin adhesive pads were staying in position for an average of four days before needing to be replaced. Since the initial fitting she has had a number of vacations to very hot countries without problems. It is now almost two years since this first fitting (fig 5) and the situation is unchanged, no skin problems have been encountered. The prosthesis is worn for an average of 15 hours every day.

Figure 4

Figure 5

Since this initial trial a small number of additional fittings of shoulder caps have taken place all with the same success. We are currently looking at the possibility of using this method to suspend lightweight endoskeletal shoulder disarticulation prosthesis.

Although the overall numbers involved in this trial are relatively small the results so far show promise in improving the quality of life for a number of forequarter amputees and we have no hesitation in recommending the method to others.

Construction method

The main concern for many forequarter amputees is to have their shoulder profile accurately restored and for a number of years we have used the method of using the inverted shape of the sound side to make the mould. This method has proved to be extremely accurate and although more time consuming in the early stages than paper profile methods pays dividends at fitting stage as is it very rare that anything other than minor modifications are required to the shape of the shoulder profile.

A slab plaster cast is taken of the complete upper torso and from this casts of both sound and amputated sides are duplicated. Various plumb and reference lines are marked on the plaster cast before removal. The amputated side cast is modified in the usual manner and smoothed while the sound side cast is only smoothed. A thin laminate of two layers perlon stockinet and 100% flexible acrylic resin is then made over the sound side cast. When cured this is cut from the cast and trimmed. Next heat is applied to the laminate allowing the mould to be easily inverted thus taking on the profile of the missing shoulder and acting as an outer mould.

The modified amputated side cast is then covered with PVC under vacuum and the laminated shoulder former placed on the amputated side cast utilising the plumb and reference lines for accurate placement. The former is securely taped in position and filled with lightweight rigid polyurethane foam. When cured the foam is then fitted to the patient and any minor modifications required to give a perfect shoulder profile carried out. The next stage is to produce finished outer mould. For ease and speed this is thermoplastic draped using polypropylene.

PElite or any other suitable material is then shaped to the size selected for the Amoena suspension areas and adhered to the PVC covering of the amputated side cast to allow for the combined thickness of the Amoena pad plus a layer of hook velcro. Both the cast and outer polypropylene mould are then sprayed with a suitable release agent. The polypropylene outer mould is then accurately positioned over the cast taped and filled with Otto Bock Pedilen Flexible Foam 150 (fig 6). When cured the foam cap is removed from the mould and flash areas removed (fig 7). Hook Velcro is then glued onto the foam cap using a suitable adhesive.

Figure 6

Figure 7

The most suitable method of positioning the shoulder cap on the patient is first to check the position without fitting the Amoena pads. If necessary light marks can be made on the skin to indicate the correct position using a suitable skin marker. The Amoena pads are then cut to shape and fitted to the Velcro in the shoulder cap but at this stage, the protective paper over the adhesive on the underside is left in place. The protective paper is then removed from the top pad and the shoulder cap placed in position on the patient. If the correct position is obtained the covers are removed from the other pads and pressed into contact with the skin.

Removal of the shoulder pad is achieved by placing a finger(s) between the skin and shoulder cap and gently easing the Velcro apart. If care is taken at this stage the skin Velcro will stay securely in place ready for the next application.

ADAPTATION OF LOCKING SHOULDER JOINTS TO INCREASE FUNCTIONAL RANGE OF MOTION FOR BILATERAL UPPER LIMB DEFICIENCIES

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Upper Extremity Prosthetic Program

INTRODUCTION

Patients with "high" level bilateral upper extremity deficiencies require maximum rehabilitation to increase functional independence and self-care skills. When the deficiency includes one or both shoulder joints, functional range of motion is compromised immensely. This limitation in shoulder joint range of motion reduces the overall level of function as well as the ability to perform basic activities of daily living (ADLs).

BACKGROUND

Current prosthetic shoulder joints allow for shoulder flexion, extension, adduction and abduction. Other joint types have been proposed in related research. While most joints utilize passive friction for positioning, a limited number allow for positive locking in flexion and extension. Friction type joints do not adequately support the weight of most prosthetic systems (body powered or externally powered) while in a flexed position. Positive locking type joints allow the prosthesis to be utilized throughout a greater range of motion.

The limitation of these joints has been the actuation of the locking mechanism. Typically, a manual release cable is attached to a mechanical lever switch for operation. The lever switch is very difficult for the high-level bilateral amputee to operate. Also, this requires the amputee to utilize the contralateral limb/prosthesis for operating the lever switch of the shoulder side. Chin nudge switches have been used with minimal success in this application as catastrophic failures have been experienced when high forces are placed on the control cable for operation. In addition, the amount of force required on the nudge to activate the unlocking mechanism is excessive and very difficult to operate with the chin. Some patients may not have the range of motion necessary to operate such a device.

DISCUSSION

The unique challenges presented by such patient profiles necessitate the development of a powered actuation system for the shoulder joint. The system utilizes a simple electronic switch to control an actuator that in turn operates the shoulder unlock mechanism. This actuator allows the patient to preposition the arm in flexion and extension independently, without utilizing the contralateral prosthesis.

Components

The powered shoulder unlock mechanism consists of several individual components discussed below and shown in Figures 1, 2, and 3. There are a variety of prosthetic options and combinations of externally and body powered elbows, wrists, terminal devices and controllers that will effectively address the needs of the patient. Specific component selection must be determined on an individual basis.

1. **Switch**. An Otto Bock rocker switch was used to activate the actuator. A variety of push or pull type switches would also be effective depending on the individual situation.

2. **Actuator/motor**. A Michigan hook actuator was modified for use in the shoulder by disengaging the breaking mechanism to allow for rapid return for shoulder locking. The actuator is mounted on the distal-lateral aspect of the shoulder socket to enable effective pull on the shoulder lever unlock mechanism.

3. **Power supply**. Power supply can be provided by standard 6 volt Otto Bock (or comparable) battery or accessing the power supply at the elbow.

4. **Pulley.** A force amplification pulley is attached to a cable from the unlock lever to decrease the force on the motor. This will increase the amount of excursion required. An excursion amplifier pulley was used.

5. **Cable**. Spectra cable was used for ease of adjustment and replacement. Spectra cable is available from TRS.

6. **Extended unlocking lever on joint**. The unlocking lever was extended to reduce the force required to operate the joint and to position the alignment of the pull cable.

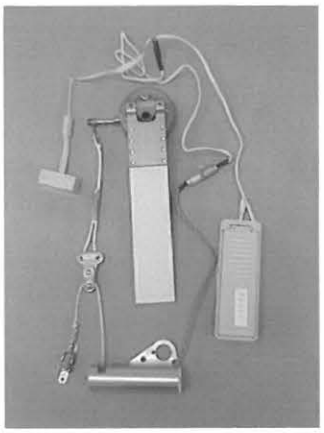

Figure 1. Unassembled Components

Figure 2. Diagnostic Test Prosthesis

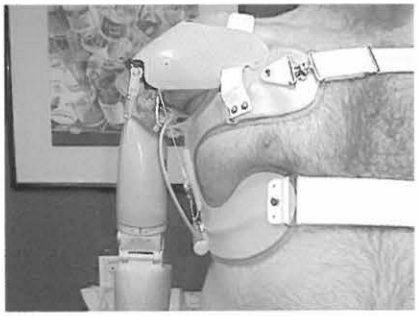

Figure 3. Finished System

Operation / Positioning

Patient applies continuous pressure to the rocker switch with their chin while bending forward and allowing gravity to position the arm in flexion or by positioning the arm against an external object. Once the desired position is obtained, the rocker switch is released and the shoulder joint automatically relocks. To reposition the arm back to anatomical position, the procedure is reversed by first decreasing the load to the joint (with gravity or against an external object) and applying continuous pressure to the rocker switch.

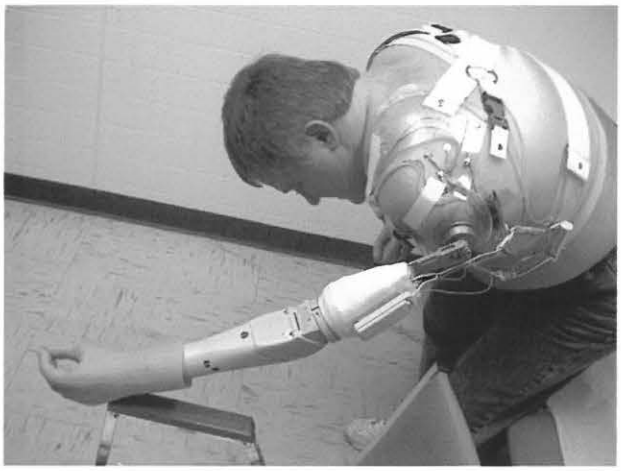

Figure 4. Patient using "object" to position prosthesis.

Figure 5. Prosthesis positioned for use.

FUTURE

Features for the next generation of shoulder joint locking mechanisms are outlined below:

1. An alternating locking/unlocking actuator which would allow natural free swing of the shoulder during ambulation. This would also reduce battery drain when maintaining the shoulder in the unlocked position.

2. Reduced size of both the locking actuator and actual shoulder joint.

3. Integrate control and power of the locking actuator into the control system for prosthesis using a programmable controller.

CONCLUSION

High-level bilateral amputees require systems to improve functional abilities for activities requiring full forward flexion and above-head orientations. The ability to initiate and sustain these alignments without dependence on the contralateral limb/prosthesis dramatically enhances the ease of use of the prosthetic system. As a result, greater independence leads to greater function.

Four powered shoulder joint unlocking mechanisms have been successfully fit to date. Each patient was able to independently position the prosthesis. Each patient experienced an increase in independence and function after being fit with this system.

Further research and development will continue to maximize functional independence and activities of daily living for such complex patients. Applications for unilateral amputees will also be addressed in future research.

MICROFRAME INTERFACE DESIGN FOR HIGH LEVEL MYOELECTRIC PROSTHESES

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ABSTRACT:

The provision of a myoelectric prosthesis for high level upper extremity amputees requires consideration of a myriad of factors. Although state-of-the-art componentry can dramatically effect the outcome, other factors left unaddressed can negate any advantages that new technology can offer. Many individuals fit with a prosthesis from the humeral neck level and higher often complain that the weight of the prosthesis, heat build-up while wearing the prosthesis, lack of stability, difficulty in independent donning, and reduced control of terminal device while in certain planes and body positions have resulted in reduced wearing times and in many cases discontinuation of prosthetic use all together. Of course, the factors listed above that lead to reduced wear and in some cases discontinuation are not comprehensive, but do overwhelmingly categorize the responses of over 250 high level amputees polled across the United States between 1993-1999. Although a panacea does not exist to completely eliminate the above factors that lead to discontinuation, substantial success at addressing these concerns can be found in the use of an interface design. Typical interface designs for high level amputees can generally be divided into three classes. 1) "Bucket Interface Style" that completely covers the effected shoulder and torso often to the midline. The major disadvantage of this style is overheating of the wearer due to excessive interface to skin coverage. 2)"Modified Bucket Style" which encompasses the basic design of the "Bucket Interface Design" as it completely encapsulates the shoulder girdle but does not extend to the midline or inferiorly to capture a majority of the torso. This design has partial success in the reduction of heat build up experienced by the wearer, but often results in poor stability and lack of skin to electrode contact which is manifested in poor terminal device control in certain planes and body positions. Additionally, auxiliary harnessing is required for this design which can have a negative impact on independent donning. 3) "Sauder Frame Design" which involves the use of aluminum struts to reduce heat build up and provide acceptable stability and donning effort. This design has not been universally adopted in the United States due to high degree of fabrication time and skill required. The purpose of this paper is to detail the MicroFrame Interface Design, a more effective alternative to the three current interface designs for high level myoelectric prostheses.

FINAL PAPER NOT RECEIVED AT PRESS TIME

AUTHOR INDEX

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