



**The Journal of the International Society  
for Prosthetics and Orthotics**

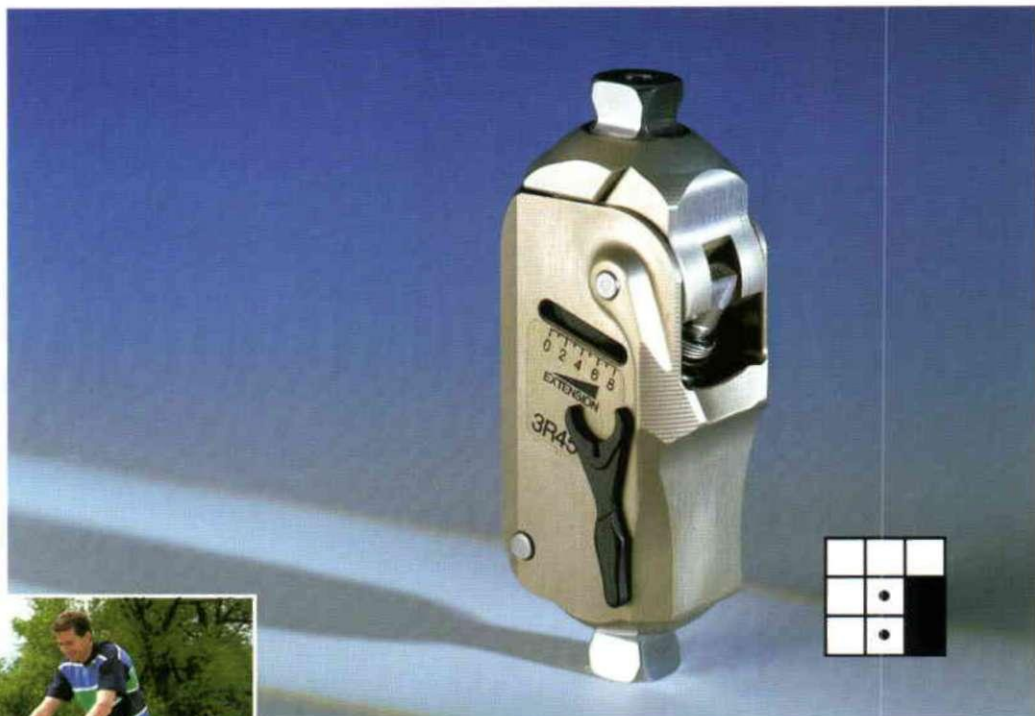
# **Prosthetics and Orthotics International**

**December 1995, Vol. 19, No. 3**

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N. A. Jacobs (President-Elect)	UK
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## *Secretary*

Aase Larsson	Denmark
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## Editorial

Over the years the Society has attempted to highlight particular subject areas in prosthetics and orthotics that require consideration, analysis or direction. This has been achieved by gathering together experts in a particular area and providing them the means of examining the topic in depth.

The first major area to be examined in this way was prosthetic and orthotic education and training, particularly that related to developing countries. A series of meetings were held: Prosthetics and Orthotics in the Developing World with respect to Training and Education and Clinical Services, Moshi, Tanzania 1984; International Symposium of Prosthetic-Orthotic Educators, Toronto, Canada 1984; Training and Education in Prosthetics and Orthotics for Developing Countries, Jönköping, Sweden 1985, and Up-grading in Prosthetics and Orthotics for Technicians from Developing Countries Trained on Short Courses, Glasgow, Scotland 1987. As a result of these meetings the Society was able to evolve a policy of education and training which recognised the needs of developing countries as well as their limited resources. This policy has been widely discussed and has been adopted by the major agencies involved in education and training in the developing world.

The Society has held three workshops which examined new developments in prosthetics and orthotics; Above-Knee Fitting and Alignment, Miami, USA 1987; Teaching Material for Above-Knee Socket Variants, Chicago, USA 1987; and CAD CAM in Prosthetics and Orthotics, Seattle, USA 1988. The reports of these workshops were widely disseminated and helped influence the practice at that time.

In 1990 the Society held its first consensus conference on the topic of Amputation Surgery in Glasgow, Scotland. As a result of that conference, a series of courses in Amputation Surgery and Related Prosthetics were held in the Netherlands (1992), Tanzania (1993), Thailand (1994), Slovenia (1994) and Panama (1994). Further courses are presently being planned.

A Consensus Conference on Lower Limb Orthotic Management of Cerebral Palsy was held in Durham, USA 1994, the report of which is presently being printed and subsequently will be distributed. As a result of this conference, it is intended to produce a standardised protocol to be used when dealing with patients with cerebral palsy. The protocol would cover such elements as patient assessment, treatment types and outcome measurements and would allow comparison of the work carried out in the different centres. In addition, consideration is being given to organising courses to disseminate the information from this conference more widely.

The latest consensus conference was on Appropriate Prosthetic Technology for Developing Countries which was held in Phnom Penh, Cambodia in June 1995 in association with USAID and WHO. The conference brought together the major agencies in this subject area. The report of this conference is near completion and will be printed by the turn of the year, in addition, a special issue of *Prosthetics and Orthotics International* based on the conference will be produced in 1996. As a result of the conference, a meeting of US Agencies involved in prosthetics in the developing countries has been organised in Washington in January 1996. The Executive Board is considering other ways of spreading the information agreed by this conference.

Such conferences have helped to advance knowledge and practice in many subject areas and efforts will be made to examine other topics in a similar way. A subject that has been neglected in the past is poliomyelitis. Planning has already begun to hold a consensus conference on this topic covering both the problems in the developing world and post-polio syndrome. It is hoped that the Society will find support for such a meeting and it will have as successful an outcome as its predecessors.

The Society is open to new thoughts on this area of its work and the Executive Board would be pleased to hear of any ideas which you may have on subject areas that can be treated in this way.

Norman A. Jacobs  
*President-Elect*

## **Executive Board Meeting**

**30 September-1 October 1995**

**Amsterdam, Netherlands**

The Executive Board welcomed Hans Christian Thyregod as the incoming Chairman of the Finance Committee.

### **Finance**

The Chairman of the Finance Committee, Hans Christian Thyregod, and the Honorary Treasurer, Steen Jensen, proposed a modified procedure for the investment of the Society's capital. This was in response to the expressed concern of the Executive Board regarding the investment, in 1994, of a substantial proportion of the Society's capital in long term bonds. Although the initial loss in value of the bonds has now largely been recovered, and their value continues to increase, the funds are not sufficiently accessible. It was proposed that the investments of the Society should be held as an appropriate balance of cash, short term bonds and long term bonds. The investments should achieve the highest possible gains with modest risk. This Capital Investment Policy was agreed by the Executive Board. A proportion of the existing long bonds will be converted to cash and short bonds when market conditions are suitable.

### **Executive Board Elections**

The Chairman of the Protocol and Nominations Committee, Mel Stills, introduced a schedule for the Executive Board Elections for 1998. This now includes an extra stage designed to increase the input from National Member Societies by giving them the opportunity to offer names of members or fellows for consideration prior to the formulation, by the Executive Board, of the slate of nominations. The deadline for receipt of these names, by the Honorary Secretary, will be May 1996.

### **Education**

The Chairman of the Education Committee, John Hughes, reported that the package of information on Category II practitioners (Orthopaedic Technologists), to include job description, learning objectives and content and level of final examinations should be ready for the next meeting of the Executive Board.

On the question of update courses on Amputation Surgery and Related Prosthetics, the Chairman reported that the Education Committee was currently exploring the possibility of two courses in the industrial world and two in the developing world. External funding is currently being sought.

Jean Halcrow was developing a programme for courses in Lower Limb Amputee Management and Upper Limb Amputee Management for nurses and therapists (and possibly doctors). It is intended that the first course be held in London in late 1996. It was anticipated that this would attract participants mainly from Europe. The possibility of such a course in the USA is also being considered.

John Hughes had recently carried out a three yearly inspection of TATCOT, the School for Orthopaedic Technologists in Tanzania. He reported that the standard of the school appears to have been upheld and recommended that the school should continue to be recognised for the training of orthopaedic technologists for a further period of three years with the recommendation that the school should seek to involve staff in clinical activities and institute a staff development programme. This was endorsed by the Executive Board.

A Certification Sub-Committee, chaired by Michael Schuch, had been formed to focus on the possibilities of developing a programme for the international certification of prosthetist/orthotists. It would undertake the task of developing a Category I package similar to that being developed for Category II and also intends to consider the idea of registration for prosthetist/orthotists.

It was agreed that Michael Schuch, as Chairman of the Certification Sub-Committee, should proceed to organise a meeting to take forward the analysis of the certification trial examinations.



**Membership**

The President-Elect, Norman Jacobs, advised the Executive Board that there were 28 National Member Societies and that the total membership of the Society was 2,690. He expressed concern over the number of cancelled memberships, some of whom were clearly active in the Society. This problem of non-payment involves the mailing of numerous reminders and is very time consuming for those involved. The International Committee will be asked, at its interim meeting, for comments and guidance.

**Publicity and Publications**

The Chairman of the Publicity and Publications Committee, Gerhard Fitzlaff, presented the recommendations of that Committee. It was agreed, following a recommendation from Amar Jain, that the instructional video on trans-tibial amputation surgery should be promoted through the WOC Newsletter possibly in collaboration with commercial companies. There will be continued efforts, by the editors, to reduce the mailing costs of Prosthetics and Orthotics International. Examples of different ISPO logos were shown. It was recommended that information should be given to all National Member Societies regarding the use of the ISPO logo and title and the availability of ISPO signs. The Publications Committee will continue to examine the possibility of providing further promotional material. It had previously been decided that the Directory of Members should not be printed because demand for the directory printed in 1992 had been very poor. However, it was felt that updated information should be available in some form. One method to consider was a computer disk and this idea will be examined.

**Standards**

David Condie updated the Executive Board on the work of ISO TC 168, ISO TC 173 and CEN TC 293. ISO TC 168 has working groups dealing with: Terminology and Classification, Medical Aspects and Physical Testing. Although the major focus has been prosthetics, there will now be a strong emphasis on standards for Orthoses. ISO TC 173 has sub-groups concerned with: Classification of Technical Aids for Disabled Persons and Wheelchairs. CEN TC 293 has sub-groups for Prostheses and Orthoses and Wheelchairs.

**Research and Evaluation**

David Condie discussed the Consensus Conference on the Orthotic Management of Cerebral Palsy. The final report was nearing completion and will be mailed to the participants and to all National Member Societies. There were two areas in which the conference felt ISPO could take action: firstly by producing a standardized protocol for such elements as patient assessment, treatment types and outcome measurements (might be reviewed in a few years at a follow-up conference); secondly by organising instructional courses on this topic.

Binks Day, as Task Officer, presented a preliminary report of the Consensus Conference on Appropriate Prosthetic Technology which had been held in Phnom Penh, Cambodia. A short report had been submitted to USAID from whom much of the funding had been received. It was anticipated that the final report would be completed by the end of 1995.

Michael Schuch presented initial ideas for a Consensus Conference on the Management of Poliomyelitis. It was agreed that he should chair a steering group which will organise the conference covering the two major areas, continuing polio in the developing world and post-polio syndrome. It was proposed that the conference should take place in May 1997. It was felt that, on balance, the logistics favoured holding this conference in the industrial world and probably in the USA.

**International Consultants**

The Honorary Secretary reported that several communications had been received from International Consultants. Daniel Suarez and Carolina Schiappacasse of the Argentinian National Member Society have been very active as International Consultants for Central and South America. They have organised several courses and have forged links with Uruguay, Chile, Brazil, Peru and Bolivia. They expressed the hope that a twinning arrangement with the US National Member Society would enable

them to fund the travel of six South American student prosthetists/orthotists to attend courses which were being provided, free of charge, in Buenos Aires. Juan Martina of the Caribbean National Member Society, International Consultant for Central and South America, reported that several events had been organised in 1995 by ISPO-Caribbean, with more planned for 1996. ISPO-Caribbean had strengthened ties with Ecuador, Venezuela and Suriname and there was a hope that Ecuador and Suriname might form National Member Societies. A congress was being organised jointly by ISPO-Colombia, Caribbean, Panama and Argentina in Bogota in July 1996. He was planning to invite all of the International Consultants in his area of the world to meet together to exchange ideas and develop a strategic plan to promote ISPO and encourage the formation of new National Member Societies.

Ed van Laar of the Dutch National Member Society, International Consultant for Central and Eastern Europe described activities in Hungary where he had been instrumental in the formation of a new National Member Society. In particular, he was proposing to hold three conferences, each of one day duration, for about 200 doctors/therapists/technicians.

The President and Eiji Tazawa of the Japanese National Member Society, International Consultants for the Pan Pacific had compiled a list of key persons involved in Prosthetics and Orthotics in the South East Asian countries. The President advised the Executive Board that he intended to write to these key persons with a view to forming a network for future conferences and courses on prosthetics and orthotics, and for the purpose of promoting ISPO.

Harold Shangali of Tanzania, International Consultant for Africa, reported on the aims and objectives of the African Rehabilitation Institute (ARI) and gave the location of its offices in Zimbabwe (head office), Congo and Senegal and planned offices in Uganda and in North Africa (location not yet determined). He had also listed key persons in Africa and gave statistics related to prosthetic/orthotic provision in Tanzania, Kenya, Uganda, Zambia, Zimbabwe, Ethiopia and Malawi. When asked about the impact of the twinning of the UK National Member Society and Malawi, he reported that it had been positively received but there was anxiety with regard to continuity as the current activity has a finite term.

### **International Organisations**

The Society continues to enjoy good relations with a number of other International Organisations: the International Association of Prosthetists and Orthotists (INTERBOR), World Health Organisation (WHO), Rehabilitation International (RI), the International Commission of Technology and Access (ICTA), the African Rehabilitation Institute (ARI), Internationaler Verband der Orthopädie Schutetechniker (IVO), World Orthopaedic Concern (WOC), United Nations (UN), the International Committee of the Red Cross (ICRC), World Rehabilitation Fund (WRF), the United States Agency for International Development (USAID) and the International Rehabilitation Medical Association (IRMA).

### **Congresses**

It was agreed that the recent World Congress in Melbourne, Australia had been a great success. Valma Angliss presented the report on the Eighth World Congress which contained a great deal of valuable information. Valma Angliss and the organising group were congratulated and thanked, on behalf of the Society, for their excellent work.

The Chairman of the International Congress Committee, the President-Elect Norman Jacobs, informed the Executive Board that preparations for the Ninth World Congress, to be held in Amsterdam, The Netherlands in 1998, were progressing well. The International Congress Committee had toured the venue, the RAI Congress Centre, and had been impressed by the facilities. The initial response from exhibitors had been very favourable.

### **International Committee**

The Executive Board considered possible venues for the Interim Meeting of the International Committee. It was decided that the Interim Meeting should be held in Denmark as this would be likely to minimise the cost of travel for most members attending.

Brendan McHugh  
*Honorary Secretary*

**The Brian Blatchford Prize for the triennium 1992-1995 was awarded to Össur Kristinsson. This paper was based on his acceptance speech which was delivered at the World Assembly in Melbourne, Australia on 6th April 1995.**

## **The Brian Blatchford Prize acceptance speech**

Ö. KRISTINSSON

*Össur hf, Reykjavik, Iceland*

The announcement that I would be awarded this honourable prize warmed my heart. At the same time, suddenly I felt old. You know, prizes usually are awarded to say thank you and goodbye. And then, just as suddenly, it dawned on me that I am indeed very old. In a few years, the year 2050 to be specific, I will turn 106 years old. Now, I do not look very much forward to that birthday. But let it come and I will face it like a man.

As a requirement of the prize it is required that I give a kind of Curriculum Vitae.

I started my education in prosthetics and orthotics in Sweden the year 1962. Eventually I came to work with a company called Een-Holmgren Ortopediska, a company that later became a part of the LIC concern, and there I worked with many excellent prosthetists and orthotists and other good people.

During this time I had a lot of fun. Most trans-tibial sockets were then already made from plastics, and we used lots and lots of polyester resin. But, we also made sockets from aluminium and wood. Creating a trans-femoral socket from a closed-end aluminium cylinder, shrinking, expanding and forming, or finishing a wooden socket into a beautiful piece of thin walled, lightweight but strong structure, reflecting the pride of the craftsman, was pure fun. I was young, life was a delight and the skills I was acquiring were pure art and I am proud and grateful to have had the opportunity to participate.

I also had some time to experiment with improvements and new designs in cooperation with the late Gunnar Holmgren. We were

working with the victims of the Thalidomide tragedy and a knee unit with lock, child's size, was called for. I do not remember if there was not any such unit available or if we just wanted one of our own. Anyhow, I designed a unit with a locking pin located in the tube and we made and applied quite a few of them. This knee was later scaled to full size and is still on the market in that guise.

During the seventies I spent my time building my company and working on improvements of existing techniques as means to better serve my customers. This eventually led to the flexible socket with a rigid external structure for support, later to be called the ISNY socket concept when the Een-Holmgren Company and the New York University School of Prosthetics became involved. This concept was presented at an AOPA meeting in Houston 1983. There was an ISPO instructional course organised in Jönköping Sweden, and instructors from schools all over the world came. From there the concept swept the world in very short time. We still use it in our practice, sandwiching the frame between two thin layers of very flexible polyurethane laminates for extremely comfortable sockets.

Parallel to the ISNY project another new concept was taking shape and was presented shortly thereafter. That was the roll on silicone sleeve that later, as a commercial product, got the name Iceross. Presented as a technique for custom making silicone sleeves it then took considerable time and effort to find the materials and production methods required for a high quality commercial product. This concept was also presented at an AOPA meeting and led to products like the 3 S and similar, even before we were ready to enter the marketplace with our Iceross.

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Then came the Masterstep foot, of which the most outstanding features are the adjustability and a truly progressive carbon fibre spring, then the pressurised casting instrument which I firmly believe in and, believe it or not, I am still at work on it. Referring to the opening sentences of this speech, I still may have a trick or two up my sleeve. What they might be remains to be seen, but they sure look exciting to us.

I want to take this opportunity to thank all the people who have helped along the way. I specially want to thank Bo Klasson for a friendship lasting for 30 years and of course my wife Bjorg Rafnar for her support through all

the years we have been together in this, through rain as well as sunshine.

Thank you for awarding me this prize. I am a proud recipient.

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KRISTINSSON Ö (1993). The ICEROSS concept: a discussion of a philosophy. *Prosthet Orthot Int* 17, 49-55.

## VIDEOTAPE ON TRANS-TIBIAL (Below-Knee) AMPUTATION

As an outcome of the Consensus Conference on Amputation Surgery, a videotape on Trans-Tibial amputation has been produced for ISPO by Amar Jain, consultant orthopaedic surgeon, and Worcester Videos. The videotape lasts for 18½ minutes and covers Indications, Assessment and Standard Surgical Techniques.

The videotape is available in PAL, NTSC and SECAM formats. Please state the format when ordering.

Copies of this videotape are now available at a price of USD 35 plus postage from:

ISPO  
Borgervænget 5  
2100 Copenhagen Ø  
DENMARK

## Amputation level assessment using lightguide spectrophotometry

D.K. HARRISON\*, P.T. McCOLLUM\*\*, D.J. NEWTON\*, P. HICKMAN\*\* and A.S. JAIN\*\*\*

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### Abstract

The aim of this experimental study was to investigate whether lightguide spectrophotometry in the visible wavelength range in skin could be used to predict stump healing viability in patients with critical lower limb ischaemia.

Remission spectra recorded at two sites (medial and lateral) on the line of a proposed trans-tibial amputation (TTA) and at 10mm intervals along the leg were analysed to give haemoglobin oxygenation ( $SO_2$ ). Degree of tissue hypoxia (DTH) along the leg was defined as the percentage of values along the leg less than 10%  $SO_2$ . DTH and mean  $SO_2$  values were compared with skin blood flow values ( $I^{125}$  4-Iodoantipyrine clearance technique) and clinical outcome of trans-tibial amputation, (TTA) or trans-femoral amputation (TFA), in 41 patients.  $SO_2$  histograms were also measured in 12 normal subjects for comparison.

The results of the study allowed the establishment of criteria for the accurate prediction of flap healing potential. Successful TTAs all displayed a minimum mean  $SO_2$  at the medial and lateral measurement sites of 30%, together with a maximum degree of tissue hypoxia of 15% along the limb. The combination of these criteria gave a sensitivity and selectivity of 1.0 for prediction of a successful outcome of TTA.

### Introduction

Peripheral arterial disease afflicts some 5% of  
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British males over 50. Its most common symptom is intermittent claudication which tends to stabilise soon after onset; nonetheless about 2% of such patients eventually need to undergo an amputation of the lower limb because of critical limb ischaemia (CLI) (Dormandy, 1991). Patients who have a trans-tibial amputation (TTA) rehabilitate better and have a better chance of leading an independent life than those who have a trans-femoral amputation (TFA) (Cumming *et al.*, 1987).

The Dundee vascular laboratory has over ten years' experience in using a combination of infrared thermographic imaging, skin blood flow recordings by a radioisotope clearance method at the site of a proposed skin flap, and Doppler pressure measurements in the pre-amputation assessment of stump viability (McCollum *et al.*, 1985). These combined methods have achieved a success rate of 93% for TTAs and a 75% TTA/TFA amputation ratio (McCollum *et al.*, 1988).

Despite the development of new techniques for the assessment of skin blood flow and metabolism, the Dundee group has not been able to improve on the TTA/TFA prediction technique since the above reports. Furthermore, there is still no reliable method for predicting the outcome of amputations in patients with CLI who may benefit from an amputation more distal than TTA. This applies particularly in patients who might be suitable for an amputation through the foot, which would allow greater mobility.

It has been shown that oxygen is one of the most important nutrients for wound healing

(Forrester, 1988). Whilst skin blood flow may be an indicator of the rate of delivery of those substrates required for the healing process, it is not necessarily a direct indicator of the adequacy of the rate of delivery of oxygen to the tissue cells (Harrison *et al.*, 1992). Furthermore, its direct measurement is invasive and exposes the patient, medical and technical staff to the hazards of ionising radiation.

Historically, hypoxia has been shown to impair both the synthesis of collagen and differentiation of fibroblasts in healing wounds (Niinikoski, 1969; Stephens and Hunt, 1971). Wound chamber investigations have revealed that the partial pressure of oxygen ( $pO_2$ ) in tissue falls steadily from 6.0 kPa (45 mmHg) in normal undamaged tissue to levels close to zero in the centre of wounds (Silver, 1980). Furthermore, the oxygen supply to the advancing edge of granulation tissue is diffusion-limited (Silver, 1969). Fibroblast activity is maximal up to 50-80  $\mu$ m from nutrient capillaries where  $pO_2$  is between 1.3 and 2.6 kPa (10 and 20 mmHg). Macrophages have a lower oxygen requirement and are to be found in the free edge of granulation tissue within wounds (Silver, 1980). However, despite being able to ingest bacteria in areas of low oxygen tension, it is uncertain whether they can kill ingested bacteria under hypoxia (Hohn, 1977). Thus it would appear that assessment of tissue oxygenation would help to predict the healing potential of ischaemic tissues.

The current method of choice for measuring tissue oxygenation is the polarographic  $pO_2$  electrode. Two commercially available techniques come into consideration for measurement in human skin: the transcutaneous ( $tcpO_2$ ) technique and needle electrodes. The former method relies on heating the skin in order to cause maximal hyperaemia and increase the permeability of the skin for  $O_2$ . The technique is time-consuming in that each measurement requires at least 20 (and ideally 40) minutes in order to obtain a stable reading and the results can be instrument dependent (Spence *et al.*, 1985). Furthermore, a reliable value for predicting viability of wound healing has not been established (Hauser, 1987). Despite the development of a skin  $pO_2$  microelectrode almost 20 years ago by Spence and Walker (1976), the only readily available system for measuring tissue  $pO_2$  with a needle

electrode employs a needle which, at 300  $\mu$ m in diameter, is too large and invasive for use in skin. It has, however, been quite widely applied to the measurement of  $pO_2$  in human muscle (Heinrich *et al.*, 1989).

A recent development has been the application of near infrared spectroscopy for the monitoring of brain oxygenation (Delpy *et al.*, 1987). Spectrophotometry has also been developed in the visible region in order to measure haemoglobin oxygenation in inflamed skin (Harrison *et al.*, 1992), and in the skin of claudicant patients during treadmill exercise (Hickman *et al.*, 1994). The possible advantages of the technique would be the fast, non-invasive measurement of tissue oxygenation using a parameter upon which local  $pO_2$  depends (via the haemoglobin dissociation curve) but which does not necessitate heating the skin. The aim of the present investigation was to determine whether lightguide spectrophotometry could be used to predict amputation level in terms of mean oxygen saturation or critical level of oxygen supply in skin and to compare the technique with the "gold standard" skin blood flow method.

## Methods

### Tissue spectrophotometry

The principle of lightguide spectrophotometric measurement of haemoglobin saturation in skin is illustrated in Figure 1. Light emitted from a transmitting fibre passes into the skin where multiple scattering events occur as photons encounter cellular and subcellular particles. The types of scattering

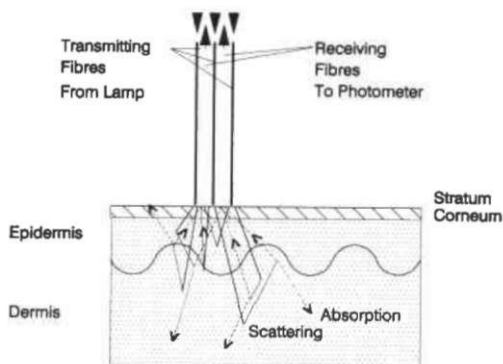


Fig. 1. Schematic diagram of light remission from skin in lightguide spectrophotometry. 3 transmitting and 2 receiving fibres are shown.

involved are discussed in detail elsewhere (Frank and Kessler, 1992). However, as seen in the figure, light will be absorbed by the haemoglobin present in the blood flowing within the catchment volume of the lightguide bundle before being eventually scattered back to a receiving fibre. Analysis of the remitted spectra (the term "reflected" is avoided due to the multiplicity of scattering events) enables the oxygen saturation ( $SO_2$ ) to be measured.

An MCPD-1000 (Otsuka Electronics, Osaka) lightguide spectrophotometer, employing a Y configuration lightguide consisting of 18 transmitting and 12 receiving fibres (each 200 $\mu$ m diameter), was used for the investigations. The light source was a 150 W Xenon lamp. Light entering the photometer is split by a diffraction grating before falling on a photodiode array. The signal from the detector, now split at 2nm intervals over the wavelength range 300-1100nm, is amplified, digitalised and transferred to a personal computer for storage and processing.

Analysis of the spectra was carried out in the visible range (500-586nm) using a 6 wavelength technique described in detail by Harrison *et al.* (1992). Briefly, the gradients between 5 experimentally determined isosbestic wavelengths (500, 520, 548, 575 and 586nm) were added to give an index which was related to the haemoglobin concentration. This index was used to normalise the measured tissue spectra.  $SO_2$  was calculated from the gradients between the absorption peak for deoxygenated haemoglobin (560nm) and the two adjacent isosbestic wavelengths (548 and 575nm) of the normalised spectra. The index thus obtained was calibrated in both *in vitro* and *in vivo* experiments to enable an accuracy of  $\pm 5\%$  to be achieved (Harrison *et al.*, 1992).

#### IAP clearance

( $I^{125}$ ) 4-Iodoantipyrine (IAP) has been in routine use for amputation level assessment for over 10 years and is preferred for skin blood flow (SBF) measurement because of its low fat solubility compared with, for example,  $Xe^{133}$ .

0.02 ml (0.1 MBq) IAP in isotonic saline was injected at two sites 100mm distal from and 30mm medial and lateral to the tibial tuberosity, along the flap line of a proposed TTA. A sodium iodide scintillation detector and photomultiplier (PM) tube was placed over the

injection site. The output from the PM tube was fed to a charge amplifier, energy analyser and ratemeter (Nuclear Enterprises, Edinburgh) before being plotted semi-logarithmically using a pen-recorder (Servoscribe). The time taken for the count rate to reach one half of its initial value ( $T_{1/2}$  method) was used for calculating skin blood flow values in ml/100g/min.

#### Patients

41 consenting patients undergoing regular amputation level assessment were investigated. Spectrophotometric measurements were carried out at both sites of IAP injection (see above) immediately prior to the injection. Nine spectra were recorded at 10mm intervals in a 30 x 30mm matrix centering on each site of injection. Further measurements were made at 10mm intervals along the critically ischaemic limb from approximately 30mm medial to the tibial tuberosity to as far as the big toe. The protocol was approved by the local ethical committee.

#### Normal subjects

Although the use of a radioactive substance for the measurement of skin blood flow in CLI is associated with minimal risk and is thus entirely justified on the grounds of its diagnostic efficacy, it was not considered appropriate for use in young, healthy volunteers for comparisons with the photometric measurements. For investigations in normal subjects, therefore, only photometric measurements were carried out, as described above, along the legs of twelve healthy adult volunteers (age range 21-42).

#### Treatment of results

The spectrophotometric data at each location were analysed as described above to give  $SO_2$  values. The 9 values measured around the injection site were averaged (mean) whereas the values measured along the limb were analysed in terms of frequency histograms representing the real distribution of tissue  $SO_2$  (cf Fig. 2).

Statistical analysis of the results, carried out with the aid of Lotus 123 and SPSS for Windows software, involved the Mann Whitney U and Wilcoxon tests for non-parametric data. Values of  $p < 0.05$  were considered to be statistically significant.



## Results

Histograms including all of the measurements along the lower limbs of normal volunteers, patients who went on to have successful TTAs and patients who had TFAs are shown in Figure 2. Of the 41 patients studied, 32 patients were predicted from SBF measurements to be TTA and 9 to be TFA cases. One successful TTA was carried out on a recommended TFA case as the result of clinical judgement: this was re-classified here as TTA. Four of the recommended TTAs were clinically judged to be more suitable for, or subsequently revised to, TFAs. These patients were thus excluded from the sample as the eventual TFA was not primarily due to critically low skin blood flow, the clinical reasons being existing infection at site of proposed TTA or subsequent wound infection. The final sample thus

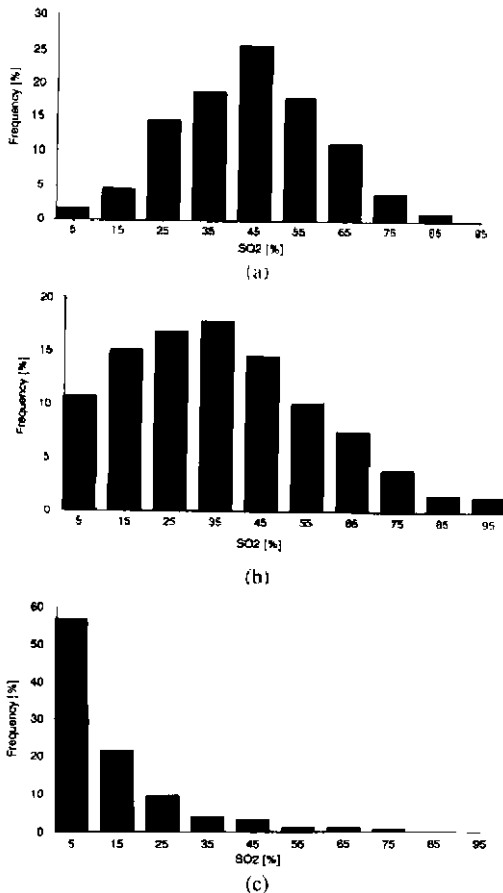


Fig. 2. SO<sub>2</sub> histograms recorded along the legs of (a) normals (N=12, n=547); (b) TTAs (N=29, n=1103) and (c) TFAs (N=8, n=286).

consisted of 29 TTA and 8 TFA cases.

It can be seen (Fig. 2a) that the distribution of SO<sub>2</sub> values along the legs of normal volunteers was normal with a mean value of 44.7% (SD 15.4%) and with very few values (less than 2%) in the lowest class (0-10% SO<sub>2</sub>). Figure 2b shows that the summary histogram for the TTA patients is shifted somewhat to the left with a mean of 37.6% SO<sub>2</sub> (SD 21.9%, median 33.5%, interquartile range (IQR) 21%). The summary histogram from the trans-femoral amputees is shifted even more to the left with 57.6% of SO<sub>2</sub> values less than 10%. The mean SO<sub>2</sub> of this group was 14.3 (SD 16.5%, median 10.2%, IQR 11.5%).

On the basis of these summary histograms, the recorded data was interrogated for the degree of skin ischaemia in terms of the percentage of SO<sub>2</sub> values in the classes 0-5%, 0-10%, and 0-20% SO<sub>2</sub>. These are given in Table 1. The table also gives the statistical summary of SBF and SO<sub>2</sub> values at the medial and lateral sites in the patients together with the values along the leg for both patient groups and normal volunteers. The significance levels (p values) of the differences between the groups are also given in Table 1. It can be seen that significant differences were observed between all parameters (except lateral site SO<sub>2</sub>) between TTA and TFA patients. Highly significant differences were also observed between normal subjects and TFAs in all SO<sub>2</sub> parameters, but the difference between the mean leg SO<sub>2</sub> measured in normals and TTA patients just failed to reach significance.

Differences between SBF and SO<sub>2</sub> values at the medial (SBF<sub>m</sub>, SO<sub>2m</sub>) and lateral sites (SBF<sub>l</sub>, SO<sub>2l</sub>) were also investigated in the TTA and TFA groups and the whole (TTA + TFA) patient sample. Significant differences between SBF at the two sites were observed in the whole patient group (SBF<sub>m</sub>-SBF<sub>l</sub>=2.8ml/100g/min;  $z=3.04$ , 34 df,  $p < 0.01$ ) and the TTA group (SBF<sub>m</sub>-SBF<sub>l</sub>=3.9ml/100g/min;  $z=3.21$ , 26 df,  $p < 0.01$ ). There was no difference in SBF between the sites in the TFA group ( $z=0.05$ , 14 df,  $p=0.96$ ). On the other hand, the only difference between SO<sub>2</sub> values at the two injection sites (SO<sub>2l</sub>-SO<sub>2m</sub>=9.8%;  $z=1.94$ , 14 df,  $p=0.04$ ) was found in the TFA group: note that SO<sub>2l</sub>>SO<sub>2m</sub>. The mean difference between SO<sub>2l</sub> and SO<sub>2m</sub> of 1.5% was not significant ( $z=0.24$ , 51 df,  $p=0.81$ ) in the TTA group. Table



Table 1. Statistical summary of results

		Mean	Median	IQR	95% CI for Mean Upper, Lower	Difference	Significance p <
Mean	Normal					TTA v Normal	
(ml/100g/min)	TTA	7.9	7.7	6.3	6.1, 9.7	TFA v Normal	
	TFA	1.5	1.5	0.8	1.2, 1.9	TFA v TTA	.0000
Lat SBF	Normal					TTA v Normal	
(ml/100g/min)	TTA	6.0	6.0	4.0	4.8, 7.2	TFA v Normal	
	TFA	1.6	1.5	0.8	1.2, 1.9	TFA v TTA	.0000
Med SBF	Normal					TTA v Normal	
(ml/100g/min)	TTA	9.8	8.3	8.4	6.9, 12.7	TFA v Normal	
	TFA	1.5	1.6	0.8	0.9, 2.1	TFA v TTA	.0000
Mean SO <sub>2</sub>	Normal	44.7	44.3	17.6	38.2, 51.3	TTA v Normal	.06
(%)	TTA	37.6	33.5	21.0	32.8, 42.4	TFA v Normal	.0002
	TFA	14.3	10.2	11.5	5.3, 23.2	TFA v TTA	.0001
Lat SO <sub>2</sub>	Normal					TTA v Normal	
(%)	TTA	33.9	32.7	24.4	27.0, 40.8	TFA v Normal	
	TFA	20.6	23.8	23.3	9.8, 31.6	TFA v TTA	.07
Med SO <sub>2</sub>	Normal					TTA v Normal	
(%)	TTA	35.4	36.9	27.1	29.0, 41.7	TFA v Normal	
	TFA	10.4	9.5	16.3	3.5, 17.3	TFA v TTA	.0001
SO <sub>2</sub> <20%	Normal	3.5	0.0	5.3	-0.7, 7.7	TTA v Normal	.0001
(%)	TTA	23.7	24.0	26.6	17.4, 29.9	TFA v Normal	.0000
	TFA	79.7	89.3	27.7	60.2, 99.2	TFA v TTA	.0000
SO <sub>2</sub> <10%	Normal	0.9	0.0	2.3	-0.2, 2.0	TTA v Normal	.0001
(%)	TTA	9.8	8.5	12.0	6.9, 12.7	TFA v Normal	.0000
	TFA	57.6	68.5	51.7	32.9, 82.4	TFA v TTA	.0000
SO <sub>2</sub> <5%	Normal	0.1	0.0	0.0	-0.2, 0.4	TTA v Normal	.0004
(%)	TTA	5.4	3.0	8.0	3.0, 7.6	TFA v Normal	.0000
	TFA	30.3	28.0	34.4	15.1, 45.4	TFA v TTA	.0000

1 shows that there was a significant difference ( $p < .01$ ) in  $SO_{2m}$  between the TFA and TTA groups but that the difference in  $SO_{2l}$  just failed to reach significance ( $p=0.07$ ). Taking the mean  $SO_2$  value from the two sites for each patient, it was found that there was a significant difference ( $z=3.15, 32$  df,  $p < 0.01$ ) between the TFA group (15.5%) and TTA (34.3%).

Figure 3 shows the scatter of the mean  $SO_2$  values along the legs in the three groups. It can be seen that there is a considerable overlap between the three groups. However, taking arbitrary mean  $SO_2$  values of 22% and 17%, in 7/8 TFAs but only 1/29 TTAs, the mean  $SO_2$  was less than or equal to 22%. On the other hand in 29/29 TTAs, but only 2/8 TFAs, the mean  $SO_2$  values were greater than or equal to 17%.

The scatters of  $SO_2$  values in the classes 0-5%, 0-10% and 0-20%  $SO_2$  are shown in Figure

4. The three measures of degree of ischaemia (i.e. the percentage of  $SO_2$  values in the classes) were tested for various upper and lower limits in order to determine the most reliable parameters for discriminating between the TFA and TTA groups. The 0-10% class was found to be the most sensitive and specific for determining degree of ischaemia between the

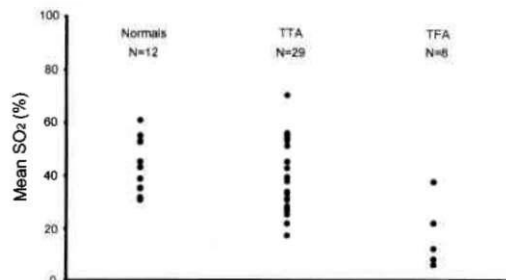


Fig. 3. Mean  $SO_2$  values measured along the legs of the three groups.

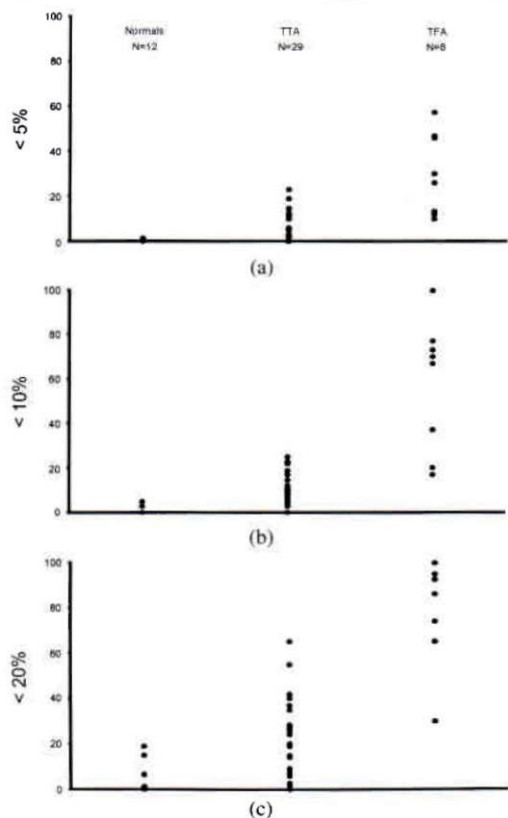


Fig. 4. Percentage of  $\text{SO}_2$  values in the range (a) 0-5%; (b) 0-10% and (c) 0-20% along the legs in the three groups.

groups and the value 10%  $\text{SO}_2$  was designated as being the value,  $\text{SO}_{2\text{crit}}$  below which tissue could be considered as being critically hypoxic. Furthermore, the degree of tissue hypoxia (DTH) is defined in this paper as the percentage of  $\text{SO}_2$  values less than 10%  $\text{SO}_2$  measured along the limb. A test for predicting a TFA versus TTA using the criterion  $\text{DTH} > 30\%$  gave a sensitivity of 0.75 and specificity of 1.0. A test for predicting a TTA versus TFA using the criterion  $\text{DTH} < 15\%$  gave a sensitivity of 0.76 and specificity of 1.0.

The results therefore demonstrate a DTH less than 15% is indicative of a successful TTA and a DTH greater than 30% indicates the necessity for TFA, but that there is a "grey area" between these limits.

## Discussion

It is clear from Figure 2 that there are considerable differences between the normal, TTA and TFA groups in terms of the degree of

hypoxia detected in the skin along the limb. This can also be seen in Figure 4 where the percentage of values <5%, 10% and 20%  $\text{SO}_2$  are shown for each individual within the three groups. The value of 10%  $\text{SO}_2$  was chosen as the critical level of hypoxia ( $\text{SO}_{2\text{crit}}$ ) on the basis of the fact that it was the value which, although not perfect on its own, could be used with the greatest sensitivity and specificity to discriminate between the groups. This level of hypoxia has also been used to delineate ischaemic regions of the human myocardium during coronary bypass operations (Frank *et al.*, 1989<sup>2</sup>).

A value of 10%  $\text{SO}_2$  in arterial blood is equivalent a  $\text{pO}_2$  of 1.3 kPa (10 mmHg). In ischaemic tissue, however, decreased affinity of haemoglobin for  $\text{O}_2$  due to increased  $\text{H}^+$  ion activity (Harrison and Walker, 1979), may give rise to a higher tissue  $\text{pO}_2$  of up to 2.0 kPa (15 mmHg) which is well above the level indicative of critical intracellular hypoxia (>1% of  $\text{pO}_2$  values below 0.7 kPa (5 mmHg)) when cytochrome  $\text{a+a}_3$  becomes reduced (Starlinger and Lübbers, 1973; Chance, 1988). However, Figure 4a shows that all of those patients in the TFA group and a substantial number in the TTA group had more than 10% of values along the leg of less than 5%  $\text{SO}_2$  (equivalent to about 1 kPa (8 mmHg)  $\text{pO}_2$  at pH 7.2) which would tend to indicate substantial critical intracellular hypoxia. Surprisingly, the value of 10%, and not 5%,  $\text{SO}_2$  turned out in practice to be the most sensitive value for  $\text{SO}_{2\text{crit}}$  in discriminating between the TTA and TFA groups. This is probably due largely to the fact that with the catchment volume of the lightguides used (Egglshaw, 1994), the majority of the haemoglobin signal detected is emanating from parts of the vascular network other than capillaries. Hence the degree of anoxia represented in Figures 2-5 is probably substantially underestimated.

It can be seen both by comparing Figures 3 and 4b, which show the individual values for the three groups, and from Table 1, which compares the mean values from the groups, that DTH (i.e. the percentage of values below 10%  $\text{SO}_2$ ) gives clearer differences, with greater degrees of significance, between the groups than the mean  $\text{SO}_2$  value along the leg. However, it could be argued that the local oxygen supply along the line of the proposed

skin flap may be of more relevance than that along the part of the limb which is to be amputated.

Figure 5 shows a scatter diagram of the mean  $SO_2$  values from the measurements at the two injection sites plotted against DTH in the TFA and TTA groups. As already seen in Figure 4b, all patients with a DTH of more than 30% along the leg had a TFA and all with a DTH less than 15% had a successful TTA. However, Figure 5 shows that although the primary criterion may be the DTH along the limb, an important secondary criterion may well be the local oxygen supply along the proposed line of amputation. It can be seen from the figure that all patients with mean site  $SO_2$  values of greater than 30% had successful TTAs.

The present results thus permit the conclusion that a DTH of 15% measured along the limb prior to amputation is indicative of wound healing whereas a DTH of 30% or greater indicates an insufficient rate of oxygen supply to support tissue healing. More local measurements along the site of the proposed wound show that a minimum pre-amputation level of 30%  $SO_2$  is required in the skin if tissue healing is to be assured. However, a successful TTA was carried out even at a mean site  $SO_2$  value of 5% given a primary criterion of a DTH less than 15% in the lower leg.

If these combined criteria of lower leg hypoxia and mean site  $SO_2$  are applied as the selection criteria for trans-tibial versus trans-femoral amputation, then both a selectivity and specificity of 1.0 can be applied to this experimental data. This compares with the skin blood flow "gold standard" with a predicted

TFA for  $SBF < 2.5\text{ml}/100\text{g}/\text{min}$  and TTA for  $SBF > 2.5\text{ml}/100\text{g}/\text{min}$  which has a sensitivity of 1.0 and specificity of 0.93 in the present study.

The apparent reliability of the spectrophotometric technique, demonstrated in Figure 5 and described above, must be judged against the calibration of the system which was carried out partly *in vitro* and partly *in vivo* in human skin. The accuracy of the method is reported as  $\pm 5\%$  (Harrison *et al.*, 1992), which, with close examination of Figure 5, could give rise to the mis-classification of some patients. In practice, in the present study, the pathological differences were apparently much greater than the errors involved in the technique.

This study raises one particularly interesting question. Table 1 shows that the SBF at the medial site was significantly higher than at the lateral site for the TTA group of patients. On the other hand, the  $SO_2$  values were significantly higher at the lateral than at the medial site in the TFA group of patients. Otherwise, no significant differences were found in either parameter.

The higher blood flow at the TTA medial site is likely to be related to the presence of the saphenous and crural arteries on the medial and posterior aspects of the leg providing collateral flow (there is no equivalent on the lateral side). There may also be local regulatory mechanisms operating in order to maintain tissue oxygen supply (Harrison *et al.*, 1990). This, however, may only be possible in the presence of a continuing, albeit low, arterial supply. In the case of the TFA patients, arterial blood flow may be inadequate for the global tissue needs with the result that metabolic changes (e.g. drop in pH (Harrison and Walker, 1979)) may interfere with local regulation and produce pathological patterns of capillary blood flow (Harrison *et al.*, 1989).

Whilst the use of such a large lightguide system in the present study did not allow the detection of highly localised hypoxia (within one or two capillaries) such pathological, localised anomalies in flow patterns have been observed, using a scanning laser Doppler technique, in experimentally induced inflammation in the skin (Harrison *et al.*, 1993). The way forward for tissue spectrophotometry in the detection of local tissue ischaemia in small volumes may thus well be the application

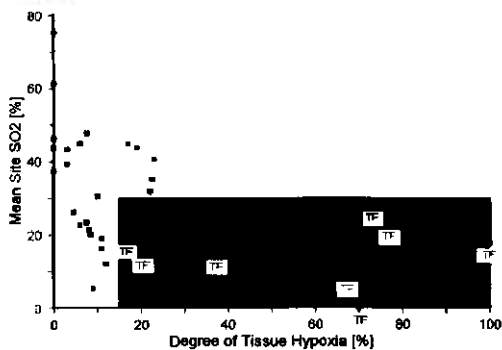


Fig. 5. Scatter plot of the mean site  $SO_2$  (mean of the medial and lateral  $SO_2$  values measured at the sites of SBF injection) against the degree of hypoxia along the limb. All TFA patients fall within the shaded area.

of micro-lightguide techniques (Frank *et al.*, 1989; Harrison *et al.*, 1994).

It is clear that this is a relatively small study and further evaluation of the technique is continuing. However, lightguide tissue spectrophotometry is fast: the sequence of measurements and evaluation taking less than 10 minutes to complete. This compares favourably with the IAP technique which takes at least as long to carry out a single measurement, dependent on the skin blood flow. Unlike the IAP clearance technique, tissue spectrophotometry is completely non-invasive and patients do not have to keep their legs completely still during the period of measurement. It therefore offers considerable advantages for this area of application.

In summary, the current study demonstrates that lightguide tissue spectrophotometry can be used for predicting the outcome of amputation in critical limb ischaemia using the criteria described.

#### Acknowledgements

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## **Congenital limb anomalies and amputees Tayside, Scotland 1965-1994**

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### **Abstract**

The purpose of this study was to review the 68 patients who had been referred to Dundee Limb Fitting Centre during the period 1965-1994, with a congenital anomaly of a major limb requiring prosthetic replacement. A profile of the incidence of congenital anomalies, amputation levels and prosthetic fitting was obtained. During the period only 68 cases with 80 congenital anomalies were referred. During these 29 years, 20 cases required surgical amputation and overall 35 surgical procedures were performed in these cases, only 3 were in the upper limb. The incidence of upper and lower limb deficiency was similar. The patients represented a small proportion (1.6%) of the patients who were reported to have congenital anomalies. Figures indicated that about 8% of all live/still births have some form of anomaly. Prosthetic fitting and use was successful in all 68 cases but long term life follow-up is necessary to ensure continued prosthetic use.

### **Introduction**

The incidence of limb deficiency, that is absence of part of a limb, present at birth has been variously reported as being 1:4264 in Canada (McDonnell, 1988) and 5:10,000 in Australia (Jones and Lipson, 1991).

Scottish figures from the Common Services Agency (CSA) of the Scottish Office report rates of 130:10,000 for 1988 (Scottish Health Service Common Services Agency, 1991) and 120:10,000 for 1989 (Scottish Health Service Common Services Agency, 1992) for all congenital anomalies. Tayside, Scotland, with a

population of 400,000 inhabitants, had a high incidence of limb anomaly (limb absence or deficiency) at 3.1% (310:10,000) of all live/still births in 1988 and 3.3% (330:10,000) in 1989. This incidence of limb anomalies is much higher than previously reported. Care must be taken when comparing results of one centre with another since incidence of reporting may vary significantly from place to place.

However, in a 29 year period only 68 patients were referred for prosthetic fitting to Dundee Limb Fitting Centre (DLFC) which caters for the entire Tayside and North Fife area. All 68 patients were successfully fitted with a prosthesis (Tables 1, 2, 3, 4, 5 and 6).

### **Method**

The records of all patients who attended DLFC since its establishment in 1965 were reviewed. The centre maintains its own medical records and stores the files on site, retaining the records of those patients who have died or left the area. No files are destroyed.

Those patients who had undergone an amputation or had a limb deficiency at birth were identified and their records scrutinised. The level of amputation was defined using the nomenclature described by Day (1988) and now published as an international standard (International Standards Organisation, 1989).

### **Results**

Table 1 lists the levels of amputation in those with limb deficiency present at birth. There were 80 congenital limb anomalies in the 68 patients referred for prosthetic fitting and these were equally divided between upper (40 anomalies) and lower limb (40 anomalies). Fourteen patients had mixed limb anomalies and 10 patients had other anomalies.

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Table 1. Congenital cases requiring a prosthesis presented at DLFC 1965-94 (N = 68)

Level of prosthetic fitting	Limb anomalies	Surgical amputation	Deceased
Wrist disarticulation (WD)	1		
Partial hand (PH)	13		1
Trans-radial (TR)	21	1	2
Elbow disarticulation (ED)	1		
Trans-humeral (TH)	4		
Ankle disarticulation (AD)	20	20	1
Trans-tibial (TT)	8	7	
Knee disarticulation (KD)	5	1	
Trans-femoral (TF)	5	4	
Hip disarticulation (HD)	2	2	
Total	80	35	1 case with clubhand 5

Table 2 lists levels of prosthetic fitting on patients with a one sided anomaly of the lower limb. All 8 patients required a surgical amputation and 7 were subsequently fitted with an ankle disarticulation prosthesis and 1 with a trans-femoral prosthesis.

Table 3 lists the level of prosthetic fitting on patients with a one sided anomaly of the upper limb. Of the 32 patients 8 were fitted with a partial hand (PH) prosthesis, 1 with a wrist disarticulation (WD) prosthesis, 20 with a trans-radial (TR) prosthesis, 1 with an elbow disarticulation (ED) prosthesis and 2 with a trans-humeral (TH) prosthesis.

There were 14 cases of patients with multiple deficiency of which 1 was bilateral upper limb, 6 upper and lower limb, 4 bilateral lower limb and 3 mixed anomaly. Fifteen surgical amputations were carried out on 11 of the patients. Twenty-six prostheses were fitted (Table 4). The details of the individual patients are outlined in Table 5.

Table 6 lists the 14 patients with limb anomalies with a specific diagnosis. Twelve surgical amputations were carried out on 10 of the patients.

In Tayside in 1988 there were a total of 4850

Table 2. Level of prosthetic fitting on patients\* with a one sided anomaly of the lower limb (N=8)

Prosthetic fitting	Anomaly	Number of cases
AD	Clubfoot	2
	Metatarsal dysplasia	1
	Longitudinal femur partial	1
	Longitudinal fibula total	2
	Longitudinal fibula total, tarsus partial, rays 4 + 5 total	1
TF	Hypoplastic limb	1

\*All patients had surgical amputation.

live and still births. Table 7 gives information with regards congenital anomalies reported in Tayside in that year (Scottish Health Service Common Services Agency, 1991).

In Scotland as a whole there were 66,569 live and still births in 1988 with 5,974 having some form of anomaly. Table 8 lists information about congenital anomalies reported in Scotland in that year (Scottish Health Service Common

Table 3. Level of prosthetic fitting on patients with a one sided anomaly of the upper limb (N=32)

Prosthetic fitting	Anomaly	Number of cases
PH	<i>Longitudinal</i> Ulna total, carpus partial, metacarpals/phalanges partial.	1
	Radius partial, carpus total, 1-3 metacarpals/phalanges partial.	1
	Metacarpals 4+5 total, 3rd phalanges total.	1
	<i>Transverse</i> Carpus partial.	3
	Phalanges partial.	2
	WD	Longitudinal carpus partial 2-4 rays total.
TR	<i>Longitudinal</i> Radius total, ulna partial, carpus total, phalanges partial.	2
	Radius total, ulna partial, carpus total.	1
	Radius partial, carpus total, metacarpal total, 1-4 phalanges total, 5 phalanges partial.	1
	<i>Transverse</i> Radius partial, ulna partial carpus total.	2
	Forearm upper.	3
	Forearm middle (with bud).	1
	Forearm middle.*	1
	Forearm middle.	4
	Forearm lower.	3
	Carpus partial.	2
ED	Forearm total.	1
TH	Arm middle.	2

\*Case requiring surgical amputations

Table 4. Level of prosthetic fitting on patients (N=14) with multiple deficiency

Prosthetic fitting	Number of fittings
Ankle disarticulation (AD)	9
Trans-tibial (TT)	4
Knee-disarticulation (KD)	4
Trans-femoral (TF)	1
Partial hand (PH)	5
Trans-humeral (TH)	2
Trans-radial (TR)	1
Total	26

Services Agency, 1991).

Table 9 lists surgical events other than the amputation performed on 15 patients. All told there were a further 30 surgical events.

Table 10 lists incidence of the surgical events in the 68 cases.

### Discussion

The incidence of limb anomaly has previously been reported as 7:10,000 in Japan (Kakurai and Kida, 1991), 5:10,000 in Australia (Jones and Lipson, 1991) and a similar rate in the United Kingdom (Evans *et al.*, 1991)

The overall incidence of anomaly as reported by the Scottish Health Service Common Services Agency (1991 and 1992) for 1988 and 1989 were much higher with rates of 8.97% and 7.47% respectively (900:10,000 and 750:10,000) but these figures included all anomalies not only limbs. The limb anomalies rate was 1.3% of all live/still births (130:10,000) but only a relatively small number are referred for prosthetic management.

Tayside figures are the highest reported by the CSA with 3.1% (310:10,000) of all live/still births in 1988 having some limb anomaly (Table 7). There were 4,850 live/still births in 1988 resulting in 150 babies having a limb anomaly. It was surprising that only one of these 150 cases was referred for prosthetic supply.

These figures are much higher than previously reported and thus of considerable interest (Table 8). Previously in the UK the reported figure was 1:4400 for both congenital and acquired amputations (2.3:10,000) (Evans *et al.*, 1991). Similarly a review by Rogola *et al* (1974) of 52,000 live births in Scotland reported a rate of 1:3000 for congenital upper and lower limb amputation (3.3:10,000).

The CSA in their reports of 1988 and 1989

Table 5. Multiple cases (no specific diagnosis) indicating level of prosthetic fitting.

1.	(a) KD Transverse leg total. (b) KD Transverse leg total.
2.	(a) KD Transverse leg total. (b) KD Transverse leg total, polydactaly
3.	(a) PH Longitudinal 3-4 metacarpal partial, phalanges partial. (b)* AD Longitudinal fibula total, tarsal/metatarsal/phalanges total. (c) Longitudinal fibula total, tarsus partial, 4 and 5 rays total. (d) Breast different sizes.
4.	(a) TH Transverse arm middle 1/3rd. (b) TH Transverse arm middle 1/3rd.
5.	(a)* AD Longitudinal tarsus partial, 3, 4 and 5 metatarsal/phalanges partial. (b)* AD Longitudinal tarsus partial, 3, 4 and 5 metatarsal/phalanges partial.
6.	(a)* AD Longitudinal fibula total, tarsus partial, 4 and 5 rays total. (b) Ilium abnormal, longitudinal ulna total, carpus partial 4 and 5 rays total.
7.	(a)* TT Longitudinal metatarsal partial, 3, 4 and 5 rays total. (b)* TT Longitudinal metatarsal partial 3, 4 and 5 rays total. (c) Pelvic dysplasia.
8.	(a) PH Transverse carpus total. (b) PH Longitudinal carpus partial. (c)* AD Longitudinal 2, 3 and 4 metatarsals partial.
9.	(a) Deformed foot - dysplasia of fibula, longitudinal tibial partial, metatarsal fusion 2 and 3. (b)* AD Longitudinal tibia total. (c) PH Transverse carpus total. (d) PH Longitudinal metacarpal 4 and 5 rays total.
10.	(a) Clubhand, Longitudinal radius total, carpus partial, 4 and 5 rays total. (b)* AD Clubfoot. Converted to AD.
11.	(a)* AD Clubfoot. (b)* TF Clubfoot initial amputation AD.
12.	(a)* AD Dysplastic femur, fibula total. (b)* TR Transverse forearm middle.
13.	(a)* TT Clubfoot. (b) Dysplastic radius.
14.	(a)* TT Dysplastic limb. (b) Hand polydactaly.

\* Surgical amputations performed.

drew attention to the difficulties in comparing one statistic to another from different studies. This is mainly due to different degrees of anomaly being reported. The CSA results are drawn from information obtained from a form



Table 6. Cases with a specific diagnosis (N = 14).

Diagnosis	Deformity	Level of prosthetic fitting
1. Coxa vara	Short limb	Inclusion prosthesis
2. Spina bifida	Neuropathy and foot deformity	TT*
3. Spina bifida	Scoliosis and foot deformity	AD*
4. Spina bifida	Shortened limb and foot deformity	TF*
5. Spina bifida	Paralytic dislocated hip	HD*
6. Spina bifida	Avascular necrosis of talus	TT*
7. Chromosomal 16	Abnormality	KD
8. Neurofibromatous		TT
9. Femoral atresia	Umbilical cellulitis	TF*
10. Scleroderma	Abnormal ilium	HD*
11. VATER association	Transverse leg total Abnormal pelvis, spinal and renal abnormalities, anal atresia	TF
12. Arthrogryphosis multiplex	Bilateral clubfeet	AD* TT*
13. Drug induced (Thalidamide)	Limb dysplasia	KD*
14. Drug induced (debendox)	Transverse tarsus total Transverse tarsus total (webbed hand, carpus metacarpal fusion)	AD* AD*

\* Surgical amputations performed.

(SMR 11) which is completed by clinicians when an anomaly is noted, using the Manual of the International Classification of Diseases,

Table 7. Congenital anomalies in Tayside in 1988 from a total of 4850 live and still births as reported by Scottish Health Service Common Services Agency (1991).

Anomaly	Number
Some anomaly	801
Spina bifida	4
Polydactyly	4
Syndactyly	6
Anomalies of feet	135
Reduction of limb	1

Injuries and Cause of Death (1977). Minor and major anomalies are recorded on this form, therefore, widely differing report rates have emerged. "The overall congenital anomaly rate varies greatly between Boards (Health Boards) almost entirely because of the differences in noting minor anomalies such as skin tags etc" (Scottish Health Service Common Services Agency, 1991).

Historically no universally acceptable nomenclature of limb deficiency has been produced. Franz and O'Rahilly (1961) devised a widely accepted classification, but other also appear in various publications which creates difficulty when comparing results.

Table 8. Congenital anomalies in Scotland in 1988 as reported by Scottish Health Service Common Service Agency (1991).

Anamolies	Live births	Still births	Rates of anomalies/10,000 live/still births			
			Live births		Still births	
			Male	Female	Male	Female
Anomalies of feet	664	2	91.6	109.5	—	117
Polydactyly	55	1	7.9	8.7	53.8	—
Syndactyly	66	4	11.7	8.1	107.5	117
Reduction of deformity	19	3	4.1	1.6	161.3	—
Other anomalies of limbs	206	6	33.5	28.6	215.1	117
Multiple congenital anomalies	41	2	6.8	5.6	53.8	58.5

Table 9. Surgical events other than single amputation (N = 30).

Problem	Level of amputation or absence	Number of operations
1. Clubfoot, radial Clubhand	AD	1 Rotation osteotomy of femur.
2. Spina bifida	KD	2 Osteotomy femur, bone graft.
3. Bilateral "deformed" feet, left hand has finger buds	AD	3 Fusions of 2, 3, 4, metatarsals. Revision of stump.
4. Spina bifida	TT	2 Multiple foot operations.
5. Radial clubhand, partial radius, scoliosis subluxation of patella	TT	1 Hip replacement
6. Spina bifida, aplastic hip, neuropathy and foot deformity	TT	3 Multiple revisions to stump
7. Spina bifida, deformed feet	AD	1 Valgus tibia osteotomy.
8. Arthrogryphosis multiplex, clubfeet	AD x 2 (later AD/TT)	3 Stump level revision. Bilateral hip replacements.
9. Scoliosis, limb shortening	TF	1 Stump level revision.
10. Fibula absence	AD	5 Tibial osteotomy x 3. Femoral osteotomy x 1. Lateral muscle release.
11. Clubfoot	AD	3 Arthrodesis, tendon transplant. Amputation Neuroma excised.
12. Syndactyly (hands and feet)	Bilateral AD	2 Ankle disarticulations x 2. Hand operations.
13. Fibula loss "flipper" arm	AD	1 Ankle disarticulation and pollicization.
14. "Abnormal" leg (Triphalyngeal thumbs)	TT	1 Tibia ankylosed to femur
15. Absent fibula	AD	1 Ankle disarticulation, Tibial osteotomy.

\* Surgical amputations performed.

O'Rahilly (1971) emphasized the need for an internationally accepted nomenclature. Day's (1988) nomenclature dispensed with ambiguous terms such as "amelia" and "macromelia".

Table 10. Summary of surgical events.

Of 68 patients, 35 (50%) had surgical amputations and 3 had bilateral surgical amputations (in total there were 64 surgical events, 1.1 per patient).
Of 20 patients who had an AD prosthesis fitted, 4 (20%) had an osteotomy, 5 of the tibia, 2 of the femur. There were 21 surgical events (one had 5 performed).
Of these patients only 3 had upper limb surgery, 3 of 35 (8.6%).
Of these 5 cases with spina bifida, 5 (100%) had a surgical amputation performed on the lower limb, in addition 2 had a varus osteotomy of the tibia and 2 had surgical amputation at trans-tibial level, one at trans-femoral.

Clear terms such as "transverse", "longitudinal", "partial" and "complete", have been described and accepted as the Standard by the International Standards Organization (1989).

During the 29 year period reported here only 68 patients were referred although one can presume that there were an approximate 4,350 cases of limb anomalies (i.e. 150 x 29). Only 14 of these have multiple anomalies with 15 amputations in 11 patients (Table 4 and Fig. 1).

Thirty-five (50%) required surgical amputation. The 35 who had surgery had 64 surgical procedures (1.1 per patient) but only 3 had upper limb surgery and this was pollicization of the fingers in two cases. These are listed in Table 9. It was found that the commonest operation was an osteotomy to correct limb alignment.



Fig. 1. Child with transverse deficiency at right radius and ulna level.

Kakurai and Kida (1991) reported that in a children's hospital where 101 upper limb amputees were presented. 6 (5.9%) required a surgical procedure whereas 44 of the 87 lower limb amputees (50.6%) had some surgical procedure. These figures are different to those of DLFC where 34 of the 38 lower limb amputees (85%) and 2 of the 40 (2.5%) with upper limb anomalies required surgery (Table 9).

The incidence of abnormal limbs has been reported by O'Rahilly (1971) as being commonest in fibula, femur, tibia, ulna and lastly humerus. Coventry and Johnson (1952) reported that the fibula was the most common deficit.

In keeping with other studies, Williams (1962) reported that fibular loss was the most common, the classical picture being one of a short limb with tibial bowing (skin dimple over the point of greatest angulation) with later delays in ossification at both ends of the tibia (Fig 2). Osteotomy may be required to correct

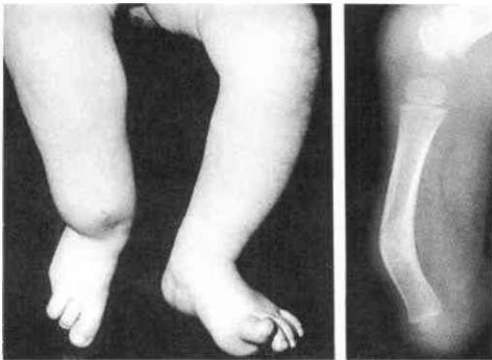


Fig. 2. (a) Child with fibular loss and tibial bowing (skin dimple over the point of greatest angulation).

(b) Delayed ossification at both ends of the tibia.

this bowing and in the period of the present study 7 such surgical procedures were performed in 4 patients, 5 in the tibia and 2 in the femur. Prosthetic fitting was achieved in all these cases.

It is important to remember that the group in the present study was highly selected as the patients were only those referred to the Centre for prosthetic fitting. In this group carpal anomaly was the most common congenital deficiency with ulna and radius being the next most common. Fibular deficiency was specifically present in 7 of the 68 cases (10%).

Humeral deficiency was less common in this group (2 cases: 3%) which was lower than O'Rahilly's (1971) figures. It is interesting that the CSA figures for Tayside, however, show that in live/still births the most common anomaly is in the foot (Table 7).

A specific diagnosis relating to the congenital problem could only be made in a small number of cases (4) (Table 6). Spina bifida was recorded in 5 cases the levels being 1 ankle disarticulation, 2 trans-tibial, 1 trans-femoral and 1 hip disarticulation. All of the five cases (100%) required surgery (Table 10). All were fitted satisfactorily with a prosthesis.

One other case was described as having an "abnormal limb" and was unable to recall why a trans-tibial amputation had been performed.

The current policy is to fit children with upper and lower limb deficiency at approximately one year old. This allows limb use to coincide with the normal developmental milestone. On average a three-monthly review is carried out with prosthetic replacement as and when necessary. Once skeletal maturity has been reached, attendance for review is as required, usually at the patient's own request. From the records it would appear that at this stage the prosthesis lasts approximately 2 years before a replacement is required.

Four children had been fitted with myoelectric prostheses for upper limb deficiency and the prosthetic follow-up has been carried out at the Scottish National Centre in Edinburgh. All have continued to use their prostheses but one patient is only an occasional user.

In 2 cases an inclusion prosthesis was used, one with coxa vara who had a short limb and one who had muscle fibrosis. Neither had an amputation but were considered as prosthetic

cases as they had anomalies requiring a prosthesis for ambulation.

Five of the 68 patients have died, 20 have left the area and 43 currently attend on a regular basis.

This review shows that only a small number of patients with limb deficiency at birth are referred for prosthetic fitting. Those patients who are referred may require surgery during their life time although functional prosthetic fitting has been successful in all those referred.

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## Lower limb amputees in Southern Finland

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### Abstract

The purpose of this study was to look at the current epidemiological trends of lower limb amputees in 1992 in the area of Southern Finland with 1.2 million inhabitants.

Similar data was collected earlier in 1984-85 and 1989. The amputation incidence was found to be 27.4 per 100,000 inhabitants. The trans-tibial/trans-femoral ratio was 0.78. The percentage of prosthetic fitting among patients undergoing unilateral trans-tibial amputation was 68% and the corresponding figure among the trans-femoral patients was 35%. The epidemiological data showed an improvement on that found 8 years earlier although the overall age structure is shifting upwards.

### Introduction

Amputations resulting from end-stage peripheral vascular diseases are a common health problem. Amputations of the lower limb are increasing in number not only because of the increasing number of elderly people in the population, but also because of factors such as diabetes, smoking, nutrition and lowered physical activity (Persson, 1980).

There is a specific requirement for lower-limb amputee statistics for the planning of preventive, operative and rehabilitative activities and the evaluation of future needs in personnel, facilities and funds. To assess the epidemiological situation concerning amputees in Southern Finland the data on all limb amputations made in all the operative units in the catchment area of the Helsinki University Central Hospital (HUCH) were collected for the

period 1984-1985 and 1989 (Pohjolainen and Alaranta, 1988; Pohjolainen *et al.*, 1989; Lääperi *et al.*, 1993). The amputation rate was 32.5 per 100,000 inhabitants in 1984, 28.1 in 1985 and 22.0 in 1989. The trans-tibial (TT)/trans-femoral (TF) ratio was 0.54 in 1984-1985 and 0.57 in 1989. The mortality rate during the first postoperative year was 39% in 1984-1985 and 36% in 1989. After the basic epidemiological survey in 1984-1985 there were various activities to inform medical and rehabilitation staff of results using local medical journals, seminars, multicentre video counselling of hospitals etc.

The purpose of the current study was to look at the epidemiological trends of lower limb amputees in the same catchment area as in 1984-1985 and 1989: incidence, diagnosis, levels of amputations, TT/TF ratio, postoperative mortality and survival. The aim was to establish whether there had been any improvement in the situation compared with the earlier studies.

### Methods

In 1992 HUCH had a catchment population of 1,258,496 representing about 25% of the total Finnish population. In this area there were 15 surgical hospitals where amputations were performed. To study the situation with regard to lower limb amputations in this area during 1992, all data concerning amputees in these hospitals were collected. The patients' hospital records were examined and data concerning diagnosis and levels of amputation were recorded. Mortality during the one-year follow-up was investigated in collaboration with the national Social Insurance Institution. The data on prosthetic fitting were collected from both prosthetic workshops in the area for the one-year period following the amputation.

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**Table 1. Distribution of amputees**

Level	Number	Percentage
Unilateral amputees		
Hemipelvectomy or hip disarticulation	5	1.4%
Trans-femoral	113	32.8%
Trans-tibial	88	25.5%
Syme, Chopart, Pirogoff	13	3.8%
Transmetatarsal or toe	60	17.4%
Bilateral amputees	66	19.1%
Total	345	100.0%

## Results

During 1992 lower limb amputations were performed on a total of 345 patients (mean age 70.5 yrs): 184 males (53%, mean age 65.3 yrs) and 161 females (47%, mean age 76.4 yrs). The amputation incidence was 27.4 per 100,000 inhabitants.

Unilateral amputation was performed in 279 cases and bilateral in 66 cases (Table 1). Unilateral TF amputation was performed on 113 patients and unilateral TT amputation on 88 cases. The TT/TF ratio was 0.78.

Vascular dysfunction was the cause of amputation in more than 90% of all patients (Table 2). The most frequent disease was diabetes.

Percentage of prosthetic fitting for those unilateral TF amputees who survived more than two months postoperatively was 35%. Corresponding percentage for unilateral amputees was 68%.

A total of 63% of all the amputees survived more than a year postoperatively.

## Discussion

According to the predictions of the Central Statistical Office of Finland the overall age structure of the population will continue to shift upwards causing a twofold increase in the proportion of over 60 year olds in the next 30-40 years. Because the increase in amputation rate seems to increase in the older age groups (Liedberg and Persson, 1983; Coddington, 1988; Pohjolainen and Alaranta, 1988), there is a need for amputee statistics with a view to planning the rehabilitation of lower limb amputees in Finland.

Lower limb amputation has not increased in the catchment area of HUCH compared with 1984-1985 but it has increased compared with 1989 (Table 3). In 1984, the amputee rate per 100,000 inhabitants was 32.5, in 1985 28.1 (Pohjolainen and Alaranta, 1988), in 1989 22.0 (Lääperi *et al.*, 1993) whereas in 1992 it was 27.4. There are few epidemiological studies concerning the incidence of limb amputations. The incidences reported above are very close to the figures reported by Liedberg and Persson (1983) 32.0 per 100,000 inhabitants in Sweden and Sethia *et al.* (1986) 27.5 per 100,000 in 1976 and 31 in 1982 in United Kingdom. Jones (1989) found lower incidences in three Australian States, 22.6 in 1981, 22.5 in 1983 and 23.6 in 1984. In Finland, the age structure of the population has caused an increase in the amputee rates. Vascular surgery can delay amputation. However, in a large North American survey Tunis *et al.* (1991) found that the rate of lower limb amputation remained stable despite the increase in vascular and reconstructive surgery. The rehabilitation and

**Table 2. Diagnosis and mean ages in three surveys of the same area**

Diagnosis	1984-85		1989		1992	
	%	Age	%	Age	%	Age
Diabetes	40.7	75	42.5	71	47.7	72
Arteriosclerosis	43.1	72	36.2	77	30.6	78
Embolism	3.8	71	2.6	69	4.9	74
Frostbite	4.4	51	1.5	55	2.3	44
Tumour	2.4	39	2.2	53	1.2	46
Trauma	2.0	41	6.4	46	6.6	43
Others	3.5	52	8.6	63	6.6	63
Total	100	71	100	70	100	70

Table 3. The trends of the amputees in three surveys of the same area

	1984-85	1989	1992
Incidence per 100,000	32.5-28.1	22.0	27.4
TT/TF ratio	0.54	0.57	0.78
Mortality during two postoperative months among unilateral amputees			
TF	32%	28%	32%
TT	17%	16%	19%
Mortality of all the amputees during the first postoperative year	39%	36%	37%
Prosthetic fitting among unilateral amputees who survived more than two months			
TF	27%	26%	35%
TT	62%	63%	68%

prosthetic services of the lower limb amputees can be planned on the basis of the incidence of 30 per 100,000 inhabitants.

Theoretically, diabetes mellitus has become the most common cause of amputation. This may be partly due to the variations in reporting comorbidity diagnosis. The total percentage of diabetes and arteriosclerosis has not significantly changed. There is no clear difference between diabetic and so-called arteriosclerotic non-diabetic gangrene. In clinical surgical practice, it can be difficult to distinguish the diagnosis of "non-diabetic" gangrene in cases of diabetic patients. Patient education regarding foot care plays an important role in prevention and management of disease. Early recognition of foot lesions, local care of lesions and aggressive treatment of infection prevent extension of the disease to adjacent areas.

Survival figures showed that 61% of the patients in 1984-1985 and 64% of patients in 1989 were alive after one year (Pohjolainen and Alaranta, 1988). The mortality rate during the first postoperative year has not changed. The mortality rate among amputees with vascular diseases indicates the advanced state of the disease. It must be emphasized that early rehabilitation and ambulation of patients especially in the case of geriatric patients is of importance.

The low TT/TF ratio of 0.54 in 1984-1985 and 0.57 in 1989 has improved but is not yet satisfactory. Among surgical hospitals in Finland there are geriatric units where TT

amputations are impossible or useless because of the final stage of the patients. Despite this more emphasis must be put on the concept of preserving the knee joint and the importance of preoperative assessment of vascular patients. In 1984-85 among the patients undergoing unilateral amputation 27% of the TF amputees and 62% of the TT amputees received a prosthesis. The corresponding figures in 1989 were 26% and 63%. In this series the prosthetic fitting in both groups was better (35% and 68%). The efforts of the medical and rehabilitation team, the use of training prostheses and more effective rehabilitation has increased the proportion of patients discharged with a prosthesis and increased the effectiveness of long term rehabilitation.

#### Acknowledgements

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## Pre and post-amputation mobility of trans-tibial amputees: correlation to medical problems, age and mortality

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### Abstract

This retrospective study compares pre and post-amputation mobility and the influence of age and associated medical problems. Data from the charts of 120 male patients who underwent unilateral trans-tibial (below-knee) amputation at the Dallas Veteran's Administration Hospital between June, 1983 and October, 1991, were collected and analyzed. Mobility was assessed with a six level scale developed by Volpicelli *et al.* (1983). The presence of cardiac disease, pulmonary disease (COPD), peripheral vascular disease (PVD), diabetes mellitus, degenerative joint disease, blindness, cerebral vascular accident (CVA), and age are correlated with changes in mobility after amputation. Older patients had more medical problems and lower post-amputation scores. Individual medical problems did not influence mobility scores, but the presence of COPD and PVD lowered pre-amputation mobility scores. Cardiac disease and diabetes mellitus influenced post-amputation mobility scores by lowering them, either together or individually. Regardless of age, however, patients with more medical problems were poor ambulators. The cause of amputation *per se* did not influence mobility scores.

### Introduction

The majority of amputations performed in the United States are a result of vascular disease or diabetes mellitus or both. Amputation surgery not only removes the pathology but also reconstructs the limb, to allow for prosthetic fitting and subsequent ambulation. Often the

medical problem that leads to amputation may influence post-amputation mobility, morbidity and patient mortality. Medical problems unrelated to the cause of amputation may influence mobility and ambulation. Awareness of those conditions which affect patient mobility, are important in the evaluation of the patient's potential ability to use a prosthesis.

The purpose of this retrospective study is to evaluate pre-amputation and post-amputation mobility in unilateral trans-tibial amputees and its association with age and the presence of medical problems. Although several studies have evaluated amputee mobility, most of these have looked at the level of amputation, with no correlation to the concomitant medical problems (Brodzka *et al.*, 1990; Durance *et al.*, 1989; Pohjolainen *et al.*, 1990; Siriwardena and Bertrand, 1991; Steinberg *et al.*, 1985). It has been shown that the more distal the amputation, the greater the likelihood of the patient being a prosthetic user.

### Methods

Data was collected retrospectively from the charts of 120 male patients who underwent unilateral trans-tibial amputation at the Dallas Veteran's Administration Hospital between June, 1983 and October, 1991. The information included age at amputation, number and types of medical problems, cause of amputation, pre-amputation mobility scores and post-amputation mobility scores. Mobility scores were assigned according to the six level scale developed by Volpicelli *et al.* (1983) (Table 1). This tool for evaluating mobility was selected because it contained specific objective criteria for the assignment of scores. The categories used were unlimited or limited community and household ambulator, wheelchair ambulator or bedridden. Other activity scores such as developed by Day

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Table 1. Volpicelli ambulatory status scale

Grade I	Wheelchair Ambulator	1. uses wheelchair at all times; 2. able to transfer with prostheses and to propel wheelchair.
Grade II	Supervised household ambulator	1. blind; 2. needs supervision during limited household ambulation.
Grade III	Limited household ambulator	1. walks less than 100 feet (30.5 metres) with prostheses in the house; 2. uses wheelchair for longer distances outside the house, may use cane, crutches, or walker; and 3. able to negotiate independently on stairs with rails, carpets, and chairs.
Grade IV	Unlimited household ambulator	1. walks at least 100 feet (30.5 metres) with prostheses in the house; 2. uses wheelchair for longer distances outside the house, may use cane, crutches, or walker; and 3. able to negotiate independently on stairs with rails, carpets, and chairs.
Grade V	Limited community ambulator	1. walks one to five blocks with prostheses; 2. uses wheelchair for longer distances, may use cane or crutches; and 3. able to negotiate independently on stairs without rails, kerbs, rough terrain, and public transportation.
Grade VI	Unlimited community ambulator	1. walks at least five blocks with prostheses; 2. uses wheelchair for longer distances, may use cane or crutches; and 3. able to negotiate independently on stairs without rails, kerbs, rough terrain, and public transportation.

(Volpicelli *et al.*, 1983)

(1981) were not used because of the availability of the simplified system by Volpicelli *et al.* (1983). Scores were recorded by either the therapist who was following the patient during in-patient or out-patient therapy or by the physician director of the amputation programme.

Data were analyzed using the Statistical Analysis System (SAS) software by an independent statistician who was not involved with data collection. The Spearman rank

correlation was used to determine linear relationships between pre and post-amputation mobility scores, number of medical problems, and age at amputation.

Paired multiple conclusions using the Kruskal-Wallis analysis were performed and significance was determined by the Student-Newmans-Keuls procedure. Association between mobility scores and presence or absence of each medical problem was determined with Mantel-Haenszel test.

Table 2. Age, mean number of medical problems, mean mobility scores

Age	Number	Number of medical problems	Mean mobility score	
			pre-amputation	post-amputation
20-29	1	0	*	5
30-39	8	0.75	6	5.87
40-49	17	1.17	4.9	5.47
50-59	28	1.78	4.71	4.96
60-69	46	1.95	4.63	4.34
70-79	17	2.17	4.63	4.0
80-89	3	1.66	*	2.0
20-49	26	1.04	5.2	5.6
70-89	20	2.1	4.6	3.7

\*none recorded

Table 3. Individual correlation of medical problems, age and mobility scores

Variable	Spearman correlation coefficient (p value)
Age, number of medical problems	+0.28 (p=0.002)
Age, post-amputation score	-0.46 (p=0.0001)
Number medical problems, post-amputation score	-0.23 (p=0.01)*
Pre, post-amputation score	+0.52 (p=0.0001)
Age, pre-amputation	-0.25 (not significant)**
Number medical problems, pre-amputation score	-0.18 (not significant)

\*N=54, \*\*N=120

## Results

Ages of the veterans in this study ranged between 25 and 89 years, with a mean age of 58 years and a standard deviation of 11 years (Table 2). Age at amputation and number of medical problems were correlated with an  $r$  value of +0.28 ( $p=0.002$ ) indicating that as patients grow older, they have greater numbers of medical problems.

A negative correlation was found between age at amputation and post-amputation mobility score ( $r$ -value of -0.46,  $p=0.0001$ ), indicating that elderly patients did poorer in terms of ambulation than younger patients (Table 3). In the multiple correlation analysis, age at amputation influenced post-amputation mobility scores, decreasing scores by 0.04 for each year over age 25. Age and pre-amputation scores were correlated with an  $r$ -value of -0.25, but this did not reach statistical significance. Reasons for the lack of significant correlation between age and pre-amputation mobility score may be that older patients may have greater morbidity or mortality due to the amputation surgery itself.

### Number of medical problems

The number of medical problems ranged from zero to four, with a mean of 1.75. Table 4 lists the number of patients with zero, one, two, three, or four medical problems, their mean age, and mean post-amputation scores. The average

Table 4. Frequency of number of medical problems and means of age, post-amputation mobility scores

Number medical problems	Number of patients	Age	Post-amputation mobility scores
Zero	16	42.1	5.6
One	30	60.3	4.5
Two	44	60.1	4.5
Three	27	60.4	4.5
Four	3	66	3.0

age of patients without medical problems is 42 years, compared to 60 years in patients with one, two, or three medical problems, and 66 years in patients with four medical problems. Pre and post-amputation scores in the zero medical problem group were at least one grade higher than all other groups. Mean post-amputation score for patients with one, two or three medical problems was the same, 4.5. The Spearman correlation coefficient of number of medical problems and post-amputation mobility scores had an  $r$  value of -0.23,  $p=0.01$ . A negative but not significant correlation ( $r=0.18$ ) was found for the number of medical problems and pre-amputation mobility score. In multiple regression analysis, the number of medical problems was not an independent predictor of post-amputation mobility score (Table 5).

### Mobility score

The mean pre-amputation mobility score was 4.61, tabulated for the 54 patients who had this score listed. The mean post-amputation mobility score was 4.65, but this was calculated using all 120 patients. Pre and post-amputation mobility scores were highly correlated, with an  $r$  value of +0.52,  $p=0.0001$ . This demonstrated that most patients were able to maintain or improve their mobility after amputation. (Table 2).

### Selected medical problems

Individual medical problems were analyzed

Table 5. Medical problem and influence on post-amputation mobility score

Medical problem	Influence on post-amputation score
CAD	-1.26
DM	-1.76
PVD + DM	-0.96
PVD + CVA	-4

Table 6. Numbers of patients with each medical problem

Medical problem	Number of patients
Cardiac disease	31
COPD	9
PVD	80
Diabetes mellitus	76
Degenerative joint disease	2
Blindness	2
CVA	10

for influence on either pre or post-amputation mobility scores, and on differences between the pre and post scores. Only two of the seven medical problems showed individual influence on mobility, and three other medical problems showed influence on mobility, and three other medical problems showed influence when in combination with other medical problems (Table 5). The number of patients with each medical problem is listed in Table 6. The total exceeds 120 because some patients had multiple medical problems.

#### Cardiac disease

Cardiac disease was listed as a medical problem for 31 patients. In multiple correlation analysis, this condition negatively influenced mobility, lowering the post-amputation scores by 1.26.

#### Diabetes mellitus

The presence of diabetes lowered post-amputation mobility scores by 1.76. Not surprisingly, the combination of cardiac disease and diabetes had a negative influence on mobility. An additive effect might have been expected, but the combination had nearly the same effect as having either diabetes alone or cardiac disease alone ( $r=-1.23$ ,  $p=0.004$ ).

#### Cardiovascular accident (CVA)

There was a significant difference in post-amputation mobility scores between patients with and without CVA. The mean score of patients with stroke was 3.3, while the mean score of patients without stroke was 4.8. The Mantel-Haenszel calculation shows a clear trend to higher scores among patients without CVA. In this calculation, 63% of patients with a score of one did not have CVA, whereas 96% of patients with score six were patients without CVA. No associations were noted between side of CVA and side of amputation or CVA prior to

or following amputation among this patient population.

#### Peripheral vascular disease (PVD)

PVD alone had no demonstrable impact on ambulation. However, peripheral vascular disease and diabetes together reduced post-amputation scores by 0.96 ( $p=0.031$ ). When analyzed in combination with CVA, PVD was shown to decrease mobility scores by more than four.

#### Chronic obstructive pulmonary disease (COPD)

Separately, COPD did not influence mobility scores. In the Mantel-Haenszel calculation a trend to lower scores was found in patients with COPD when compared to those without this disease. Among patients with pre-amputation mobility score 1, 67% were without COPD. Among patients with pre-amputation mobility score 6, 95% were without COPD. The validity and utility of this finding is unclear, since only 5 patients had the disease. An unexpected finding was that the combination of both peripheral vascular disease and COPD lowered the pre-amputation mobility scores.

#### Cause of amputation

Chart data was summarized into four possible causes of amputation: trauma, peripheral vascular disease associated with or without diabetes, and infection. Table 7 tabulates the number of patients with each cause. The total is greater than 120 because some patients had multiple causes listed. All diabetic patients for whom infection was a stated cause of amputation were assumed to also have peripheral vascular disease and this cause was added to their data. Cause factors dropped out of the multiple correlation, showing no influence on mobility.

#### Discussion

It is well known that younger amputee

Table 7. Number of patients with each cause of amputation

Cause of amputation	Number of patients
Trauma	15
Infection	62
PVD associated with diabetes	79
PVD not associated with diabetes	14

patients ambulate better than the elderly. It is generally assumed that walking difficulties are due to increasing medical complications with ageing. Several reports have noted that increasing age adversely affects mobility (Durance *et al.*, 1989; Steinberg *et al.*, 1985; Siriwardena and Bertrand, 1991). Brodzka *et al.* (1990) noted that non-ambulatory patients had a large number of medical problems and that these were more likely than ageing, to have an effect on ambulation. This was true at all levels of amputation. In this study as age increased, so did the number of medical problems. Even though pre and post-amputation scores were positively correlated, it is the post-amputation score alone that showed statistical significance in relation to age and number of medical problems. It is probable that patients with lower pre-amputation scores have a higher mortality associated with their surgery.

Ottman and Stahlgren (1965) noted serious medical problems in 71% of the patients studied, and that they had a peri-amputation mortality double that of healthier subjects. Ambulation scores in the present study may be higher than expected since selection criteria for patients included referral for prostheses, thus omitting patients in poor health who were thought not to be potential ambulators.

There was no relationship between cause of amputation and the variables of age, number of medical problems, or mobility. The reason for this may be that the categories selected may be too broad, thus making the variability within each category too great for mobility trends to be found.

Several reports have found that coronary artery disease affects walking of amputees (Couch *et al.* 1977; Moore *et al.*, 1989; Reyes *et al.*, 1977; Steinberg *et al.*, 1985). A study of trans-femoral, trans-tibial and bilateral trans-tibial amputees, noted that the incidence of cardiac disease was nearly equal in ambulatory and non-ambulatory patients (Steinberg *et al.*, 1985). There was also an influence on ambulation among trans-femoral and bilateral trans-tibial amputees, but not on unilateral trans-tibial amputees. Weiss *et al.* (1990) reported that amputees with multiple diseases and extensive atherosclerosis were less likely to walk.

There has been no uniformity regarding functional success for prosthetic rehabilitation.

Thornhill *et al.* (1986) defined success as prosthetic use three times per week whereas Couch *et al.* (1977) regarded successful prosthetic rehabilitation as prosthetic wear greater than 25% of waking hours.

One of the difficulties of evaluating groups of patients from the literature is that most studies have included bilateral amputees as well as trans-femoral and trans-tibial amputee patients. However, in most of the studies where concomitant medical problems are documented, the reasons for failure to use a prosthesis include debility, dementia, stroke and cardiac problems.

It is well accepted that traumatic amputees in general have better functional outcome. Moore *et al.* (1989) noted that non-ambulatory patients were 15 years older than prosthetic users who were 57.1 years on average. Patients who were regarded as prosthetic failures were much older. In their study only two-thirds of the trans-tibial amputees were prosthetic users.

Some reports have not found an association between diabetes mellitus and the ability of the amputee to walk. The patients in the study reported here showed significantly decreased ambulation in the presence of diabetes mellitus. The combination of diabetes with cerebrovascular accident or peripheral vascular disease was associated with lower mobility scores. One explanation would be that these diabetic patients were poorly controlled or had complications from the diabetes that negatively influenced their ability to walk.

There are conflicting reports about the influence of stroke on patient mobility. However, it would appear that if the amputee patient was ambulatory before the stroke, then continued mobility was possible (Siriwardena and Bertrand, 1991; Varghese *et al.*, 1978; O'Connell and Gnatz, 1989). No trends were noted in patients in this study, most probably due to the low number of strokes in this group.

Chronic obstructive airway disease alone did not appear as an influence on mobility, in the multiple correlation analysis. Patients with pulmonary disease did show a trend toward lower ambulation scores when compared to patients without COPD.

Although this study did not address the presence or absence of psychological problems, it has been shown that patients with cognitive deficits or covert psychiatric illness are severely

hampered in the rehabilitation potential (Pinzur *et al.*, 1988).

### Conclusion

A retrospective chart review of the ambulation of 120 trans-tibial amputees, revealed several factors to be related to post-amputation walking. Older patients had more medical problems and poorer ambulation. Regardless of age, patients with more medical problems had poor ambulation. Coronary artery disease, diabetes mellitus, cerebrovascular disease, and the combinations of diabetes and cerebrovascular accident, diabetes and peripheral vascular disease, and cerebrovascular accident and peripheral vascular disease negatively influenced ambulation. The cause of amputation did not influence mobility scores.

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## Is body powered operation of upper limb prostheses feasible for young limb deficient children?

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### Abstract

The investigators measured efficiencies of body powered prehensors and cable control components of prostheses available for young children. Results indicated that the cable control systems and hook type prehensors have moderate to high efficiencies, but children's body powered hands have very low efficiencies. Measures of arm and shoulder strength of 3-5 year-old limb deficient children, both on the limb deficient and sound sides, were less than that reported in the literature for normal children. The findings were examined in relation to children's strength available compared with prosthesis strength requirements. The comparison demonstrates a way to establish measurable efficiency targets for new prehensor designs.

The article includes detailed findings on children's strength, and findings on efficiencies of the prehensors and cable control systems of children's upper limb prostheses. Sample calculations may be useful to future designers of body powered prehensors for young children. A more efficient body powered hand is especially needed. Preliminary calculations indicate that the use of currently available children's voluntary opening (VO), body powered hands is not feasible, given the low strength of young limb deficient children and the low efficiencies of the VO hands. The use of

voluntary closing (VC) hands may be feasible but remains to be tested.

### Introduction

Young children wearing body powered prehensors have difficulty gaining a firm grip on objects. Voluntary opening (VO) prehensors would give good grip if children were strong enough to overcome the resistance of strong rubber bands or springs. Voluntary closing (VC) prehensors would give good grip if children could sustain a strong muscle contraction over an extended period of time to hold the terminal device closed. Clinicians find that children can operate body powered prehensors well enough to get a good grip by the time they start kindergarten, but the infant and pre-school period is often frustrating for children because objects slip from the prehensors.

Researchers and clinicians have addressed the problem of poor grip force in several ways. One solution has been to develop body powered prehensors with features to enhance grip or reduce effort, such as CAPP I and Adept F.<sup>1</sup> These prehensors lessen the problem of poor grip, but many parents object to the appearance of a prehensor that does not resemble a hand. Another solution has been to fit young children with externally powered, myoelectrically controlled hands. These prehensors give very firm grip, but they are costlier, heavier and require more care and access to repair than body powered prehensors. Also, the geometry of the

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<sup>1</sup>CAPP Terminal Device, Size 1 is available from Hosmer-Dorrance Corp., 561 Division Street, PO Box 37, Campbell, CA 95008, USA. Adept prehensors are available from TRS, 1280-28th Street, Suite 3, Boulder, CO 80303-1797, USA.



small myoelectric hand does not accommodate many objects used in children's activities.

In the past, researchers have had little objective data on children's strength or prehensor grip strength requirements to use as targets in establishing design criteria for the development of body powered prehensors. Objective design criteria are highly desirable in solving the problem of poor grip.

The authors have conducted studies to identify and measure sources of inefficiency in the operation of children's body powered prehensors and measure children's body strength available for operation. The investigators compared available strength input with prehensor grip strength required (Fig. 1). The findings of the analysis are summarized in this article and will provide design criteria for development of a child's prehensor.

### Review of the literature

The authors conducted an extensive review of the literature. Two studies reported objective measures of young children's arm and shoulder strength (Backman *et al.*, 1989; Sykanda and Armstrong, 1989). One study reported efficiencies of body powered prehensors (Corin *et al.*, 1987), and previously unpublished research reports provided data on control system efficiencies (Carlson and Long, 1988; LeBlanc, 1985). The reports contributed important information, but additional information was required to complete the analysis identified above. For example, the

STRENGTH AVAILABLE COMPARED WITH STRENGTH REQUIRED

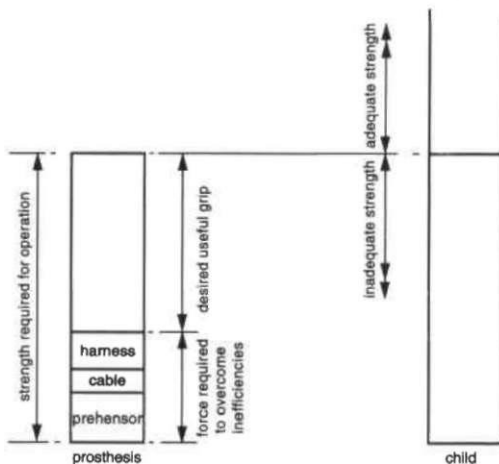


Fig. 1. Strength available compared with strength required.

studies of children's strength were conducted on normal children and no studies were found which demonstrated that normative strength standards could appropriately be applied to children with limb deficiencies. The report of Corin *et al.* (1987) on prehensors did not include all currently available children's prehensors. No studies were found that reported on efficiency of the harness or suggested any way of measuring it. Also, no studies were found that analyzed the relationship between body strength available and prosthesis operating requirements. Finally, studies on children's needs for performing activities at various ages require considerable interpretation because grip strength in children's hands varies by type of grip. Also, factors besides measured grip strength make important contributions to secure gripping of objects.

### Children's strength available for operation

Since young limb deficient children can perform gross play activities relatively well, and seem able to function almost normally, the investigators decided to compare strength of a small group of limb deficient children with published strength standards for children. Then, if limb deficient children's strength was within normal limits, published data could be used. A pilot study on arm and shoulder strength of 14 unilateral, congenital, trans-radial limb deficient children showed that limb deficient children were weaker than normal children on both the limb deficient and sound sides (Shaperman *et al.*, 1992).

### Method

Since published strength standards could not be used for limb deficient children, the investigators gathered new data by measuring arm and shoulder strength of 37 limb deficient children. All children (20 boys and 17 girls) had unilateral, congenital, trans-radial limb deficiencies and were 3, 4 or 5 years old. They came from three clinics in California. For consistency, the study was limited to children with unilateral, congenital, trans-radial limb deficiencies although the investigators recognize that children with higher level limb deficiencies also greatly need improved prehensors. Children younger than age three also greatly need firmer grip with body powered



Table 1. Description of the sample (N=37)

Age and sex						
Age (years)	3.0	3.5	4.0	4.5	5.0	5.5
Girls (N=17)	0	3	3	6	2	3
Boys (N=20)	5	3	3	2	2	5
Trans-radial limb deficiency						
Left	23	Short trans-radial			26	
Right	14	Long trans-radial			11	
Ethnic origin						
Present prehensor						
Anglo	21	CAPP terminal device			19	
Asian	0	Dorrance hook			1	
Black	2	Mechanical hand			7	
Hispanic	14	Myoelectric hand			5	
		No prosthesis			5	

prehensors, but it was not feasible to get valid measures of strength on children younger than age three. Table 1 lists the characteristics of the children in the study.

The investigators measured the strength in four motions that were considered potential sources of strength for prehensor operation: shoulder flexion (anterior motion of the humerus in the sagittal plane which represents both glenohumeral and axioscapular motion), shoulder abduction (lateral motion of the humerus in the frontal plane which represents both glenohumeral and axioscapular motion), shoulder (girdle) elevation (elevation of the scapula and clavicle measured in the frontal plane), and shoulder protraction (scapular motion away from the spinal column measured in the sagittal plane at the tip of the acromion). Procedures for measurements followed muscle strength measurement protocols reported in the literature by investigators who measured and reported strength of normal children (Sykanda and Armstrong, 1989). Measurements were made using a portable electronic dynamometer with force transducer. This myometer shows peak force attained at each trial on a digital readout in kilograms. The manufacturer reports

accuracy of the myometer to  $\pm 0.3$  kg of force\*. The same therapist made all measurements, and restrained the children from using substitute motions during measurements. Children were seated in a straight chair and held in position with a posture vest. The therapist positioned the myometer on the distal end of the humerus for measures of shoulder flexion and abduction on the tip of the acromion for shoulder (girdle) elevation and on the anterior surface of the head of the humerus for shoulder protraction. Each measure was repeated three times with rest periods between trials.

Data analysis included consideration of the following variables: age, height, sex, current prehensor, strength of sound side compared with limb deficient side and single peak force measurement compared with mean of three maximum efforts.

### Results

*Age and sex:* Findings on children's strength are reported by age, sex and side of limb deficiency in Tables 2 and 3. With few exceptions, boys were stronger than girls of the same age. On average, boys were between 0.7 and 1.9 kg of force stronger than girls of the same age, except in shoulder (girdle) elevation where differences reached 2.8 kg. Also, strength differences between boys and girls were more pronounced among five-year-olds than among three or four-year-olds.

*Sound side versus limb deficient side:* Examination of Tables 2 and 3 shows children were not consistently stronger on either the sound or limb deficient side. For each motion, each child's strength on the sound side and limb deficient side were compared. For the majority

\*Kilogram force is used throughout this paper as the measuring instrument displayed in these units. 1 kilogram force is equivalent to 9.81 newtons.

Table 2. Mean strength of 4 motions of 37 unilateral trans-radial limb deficient children

Age	Shoulder flexion				Shoulder abduction				Shoulder girdle elevation				Shoulder girdle protraction			
	LD side		Sound side		LD side		Sound side		LD side		Sound side		LD side		Sound side	
3.0	Boys	Girls	Boys	Girls	Boys	Girls	Boys	Girls	Boys	Girls	Boys	Girls	Boys	Girls	Boys	Girls
3.0	3.0	-	3.2	-	2.7	-	2.9	-	4.4	-	4.0	-	3.9	-	3.7	-
3.5	3.8	2.6	4.0	2.9	2.8	2.4	3.1	3.1	4.2	4.6	4.0	3.9	4.0	2.7	3.7	3.5
4.0	3.9	3.4	4.5	4.1	3.6	2.7	3.9	2.8	5.3	4.4	5.1	4.6	4.3	3.3	4.7	3.1
4.5	4.1	2.9	6.0	3.3	3.9	2.9	4.6	3.2	5.7	6.2	6.7	4.9	4.4	4.2	4.8	4.2
5.0	5.4	3.4	5.8	3.6	5.4	3.3	4.8	3.2	6.3	6.1	7.1	5.0	5.1	4.3	5.8	3.0
5.5	4.9	3.9	5.0	4.0	4.8	4.7	5.2	4.4	8.5	6.8	8.7	5.9	6.9	4.7	6.1	5.0

Strength is in kilograms of force, Age is in years.

Table 3. Peak strength of 4 motions of 37 unilateral trans-radial limb deficient children

Age	Shoulder flexion				Shoulder abduction				Shoulder girdle elevation				Shoulder girdle protraction			
	LD side		Sound side		LD side		Sound side		LD side		Sound side		LD side		Sound side	
	Boys	Girls	Boys	Girls	Boys	Girls	Boys	Girls	Boys	Girls	Boys	Girls	Boys	Girls	Boys	Girls
3.0	3.1	—	3.5	—	2.8	—	3.0	—	4.9	—	4.8	—	4.1	—	3.9	—
3.5	4.1	2.9	4.4	3.9	3.1	3.1	3.5	3.4	5.0	5.0	4.6	4.1	4.5	3.2	4.2	3.8
4.0	4.6	4.1	5.1	4.2	3.9	2.8	4.2	3.1	5.9	4.8	5.6	5.4	4.8	3.5	5.3	3.3
4.5	4.2	3.3	6.3	3.8	4.3	3.2	5.2	3.6	6.4	6.6	8.0	5.5	4.6	4.5	5.2	4.6
5.0	6.2	3.6	6.8	4.0	5.8	3.5	5.2	3.6	7.5	6.7	7.9	6.5	5.7	4.7	6.8	3.4
5.5	5.3	4.0	5.6	4.6	5.2	5.1	5.8	5.2	9.9	7.4	10.5	6.8	7.8	5.7	6.5	5.6

Strength is in kilograms of force. Age is in years.

of children, shoulder flexion and abduction were 10% stronger on the sound side, and shoulder (girdle) elevation and protraction were 10% stronger on the limb deficient side. Table 4 shows average differences in mean strength between sound and limb deficient side for each motion.

*Single peak force measure versus mean of three maximum efforts:* Table 2 shows mean strength of three maximum trials and Table 3 shows peak strength achieved in all trials for each motion. The average of three maximum trials consistently was slightly lower than the single peak measure. Peak strength varied by approximately 0.5 kg from the mean of three maximum trials for most motions measured, but reached differences up to 1.8 kg for shoulder (girdle) elevation, especially among five year

olds. Table 5 is a summary of the differences between the single measure of peak strength and the average of three maximum trials. Although theoreticians usually consider the single peak measure more accurate than the mean of three maximum contractions, there is growing acceptance of the use of the mean of three maximum contractions because of its greater reliability (Magnussen *et al.*, 1990; Magnussen, 1993; Sapega, 1990; Hood and Forward, 1965).

Examination of Tables 2 and 3 shows that mean strength does not increase with age in an even, linear pattern. This may be due to the small sample size. Review of the findings on individual children in this study suggests that most children had a spurt in strength at the age of 4½ to 5 years, but the effect is masked when data are aggregated to calculate means. The clearest picture of children's strength progression is shown in Table 6, where data are presented in percentiles. Findings for boys and girls are combined in Table 6 since boys and girls wear the same prehensors, and strength was measured to provide data for prehensor design criteria. The table is formatted in the same manner as pediatric growth charts. Finally, data from Table 6 are summarized in Table 7 so they are more convenient to use in the analysis.

Table 4. Differences in mean strength: sound/limb deficient side\*

Motion	Stronger side	Average difference
Shoulder flexion	Sound side	0.5 kg or 10%
Shoulder abduction	Sound side	0.4 kg or 10%
Shoulder (girdle) elevation	Limb deficient side**	0.6 kg or 12%
Shoulder protraction	Limb deficient side	0.5 kg or 12%

\* Findings reflect trends among a majority of the 37 limb deficient children in the study.

\*\*Four and five year old boys were stronger on the sound side in shoulder (girdle) elevation.

Table 5. Peak strength versus average of three maximum trials: a summary of differences. Amount that peak strength exceeded the mean of three maximum trials.

Motion	Side	Average kg	Difference Percent	Range of Differences	
				kg	Percent
Shoulder flexion	Limb deficient side	0.3	10.3	0.1 to 0.8	3 to 21
Shoulder flexion	Sound side	0.6	13.6	0.2 to 1.0	2 to 34
Shoulder abduction	Limb deficient side	0.3	9.7	0.1 to 0.7	4 to 29
Shoulder abduction	Sound side	0.4	11.0	0.1 to 0.8	3 to 18
Shoulder girdle elevation	Limb deficient side	0.8	12.0	0.4 to 1.4	9 to 19
Shoulder girdle elevation	Sound side	0.9	16.0	0.2 to 1.8	5 to 30
Shoulder protraction	Limb deficient side	0.5	11.0	0.2 to 1.0	5 to 21
Shoulder protraction	Sound side	0.5	10.3	0.2 to 1.0	5 to 17

Table 6. Strength percentiles (kg) by children's age (years)

Sound side					Limb deficient side					
5th	25th	50th	75th	95th		5th	25th	50th	75th	95th
Shoulder flexion Strength in kg										
2.2	3.1	3.5	4.0	5.2	3 years	2.2	2.6	2.6	2.9	4.4
2.6	3.5	4.4	5.0	6.8	4 years	2.3	2.7	3.1	3.5	4.7
2.6	3.5	4.6	5.6	6.2	5 years	2.5	4.0	4.4	5.5	6.7
Shoulder abduction Strength in kg										
2.1	2.5	3.0	3.2	3.6	3 years	1.7	2.0	2.2	2.9	4.4
2.5	2.8	3.2	4.1	5.0	4 years	1.8	2.5	3.1	3.5	4.7
2.2	4.1	4.5	5.0	6.4	5 years	2.6	4.0	4.4	5.5	6.7
Shoulder girdle elevation Strength in kg										
2.6	2.6	4.1	4.5	5.2	3 years	2.8	3.5	4.8	5.1	6.3
3.5	4.6	5.2	5.4	6.9	4 years	3.6	4.1	5.3	6.6	10.5
3.7	6.3	6.6	8.0	11.4	5 years	3.8	6.3	6.8	8.3	12.0
Shoulder protraction Strength in kg										
2.0	2.6	3.9	4.2	5.6	3 years	2.1	3.1	3.8	4.0	4.4
2.4	3.3	4.1	5.1	5.7	4 years	2.5	3.3	4.1	4.7	6.2
2.4	5.0	5.5	6.0	7.1	5 years	3.7	4.7	5.1	6.1	8.4

On average, boys' strength was 16% greater and girls' strength was 11% less than percentiles shown.

Arm and shoulder strength of the 37 children in this study was similar to that reported for the 14 children in the pilot study, and was less than the strength standards reported for normal Canadian and Scandinavian children (Backman *et al.*, 1989; Sykanda and Armstrong, 1989). Comparison of strength of the 37 limb deficient children with 33 normal five year olds from the same locale showed that the strength of the limb deficient children compared more favourably with a normal population (Baldwin and Priete, 1992). The strength data gathered directly from limb deficient children was used in calculations related to operation of children's body powered

prehensors in the following sections of this report.

**Related studies**

A pilot study was conducted in the pathokinesiology laboratory at Rancho Los Amigos Medical Center to learn how much of a child's strength is actually used to operate a body powered prehensor. A seven year old child with body powered trans-radial prosthesis exerted 50% to 100% of her maximum available muscle force when operating her CAPP I prehensor (exact percentage of maximum force varied for specific muscles).

Table 7. A summary of children's strength for operation of body powered prehensors

Age	Force available (kg)			
	Shoulder flexion	Shoulder abduction	Shoulder girdle elevation	Shoulder girdle protraction
3 years	3	2½ to 3	4 to 4½	2½ to 4
4 years	3 to 4	3 to 4	4½ to 6½	3½ to 4½
5 years	3½ to 5½	3 to 5½	6 to 8½	4 to 7

Ranges based on mean of three maximum contractions on limb deficient side.



An adult male with body powered trans-radial prosthesis used 50% or less of his maximum available muscle force for hook operation in a similar assessment (McNeal and DeRuyter, 1994). This finding supports observations that young children require more strength than they have available for prehensor operation, and it suggests that a new child's prehensor should require less than the maximum strength levels shown in Tables 6 and 7 if it is to give better grip than current prehensors.

Another study investigated whether a home-supervised activity programme would be effective for increasing arm and shoulder strength among 3-5 year old limb deficient children. The study followed 23 children over a one year period. Results showed that good compliance with the home activity programme was associated with increased strength beyond that which would be expected from growth and maturation alone, but the effect was still not large enough to compensate for the high operating force requirements associated with good grip in today's body powered prehensors.

### Upper-limb prosthesis strength requirements Sources of prosthesis inefficiencies

The prosthesis can be viewed as a system with three parts which contribute to the overall inefficiency: (1) the prehensor mechanism itself, (2) the cable and housing which transmit force from the harness to the prehensor, and (3) the harness which transmits force from the body to the cable.

*Prehensors:* The efficiencies of five body powered prehensors commercially available for

young children are summarized in Table 8 (LeBlanc *et al.*, 1992). The efficiency of the prehensor (EP) equals the ratio of work out at the prehensor finger tips divided by the work into the prehensor at the distal cable. The efficiency of the prehensor (EP) factor accounts for the losses due to friction in the prehensor mechanism.

*Cable and housing control system:* The efficiencies of various types and combinations of cable and housing control systems have been calculated by Carlson and Long (1988) (Fig. 2) and by LeBlanc (1985). The efficiency for trans-radial prostheses using plastic cable and steel housing with plastic liner is about 90%. For this analysis, the measured efficiency of the cable (EC) is the ratio of force out of the cable housing divided by force into the cable housing.

*Harnesses:* The authors reviewed the literature, consulted with other researchers in the field, contemplated how to measure the efficiencies of harnesses, and decided that it is not feasible to do so within the constraints of this project. It is feasible to measure relative differences between harnesses, but it is difficult to measure the absolute efficiencies of harnesses. For this analysis, the estimated efficiency of the harness (EH) is assumed as the ratio of the force out of the harness to the proximal cable divided by the force into the harness provided by body motion.

### Ratio of prehension force to cable force

The geometry of prehensors determines this ratio. For instance, the length of "thumb" lever arm to which the cable is attached on the Hosmer 10X hook is 4.45 cm. The length of the fingers is 6.35 cm to the point of prehension. The ratio of these two lengths (4.45/6.35) or 0.70 is the ratio of the prehension force to the prehensor cable pull force. If one pulls on the thumb with 4.55 kg of cable force, the prehension force at the finger tips will be 3.18 kg. This prehensor ratio (PR) will vary with the geometry of each prehensor and assumes 100% efficiency.

### Estimation of prehension force available

The ultimate prehension force which a child will be able to achieve with a prosthesis is determined by the maximum measurable force which the child can generate minus the effects of the efficiencies and ratios listed above

Table 8. Efficiencies of five body powered prehensors

Prehensor	Work efficiency*
CAPP I (regular spring)	68%
Hosmer 10X hook (two bands)	82%
Steeper 2.0 hand	25%
NYU child size hand	27%
Adept F III	**

\* Work efficiency is the ratio in percentage of the work output of each prehensor divided by the work input required for operation. In this case, work efficiency rather than force efficiency is used to take into account the effect of varying cable excursion.

\*\* The Adept is a voluntary closing (VC) prehensor and work efficiency cannot be compared in the same way as for the four voluntary opening (VO) prehensors.

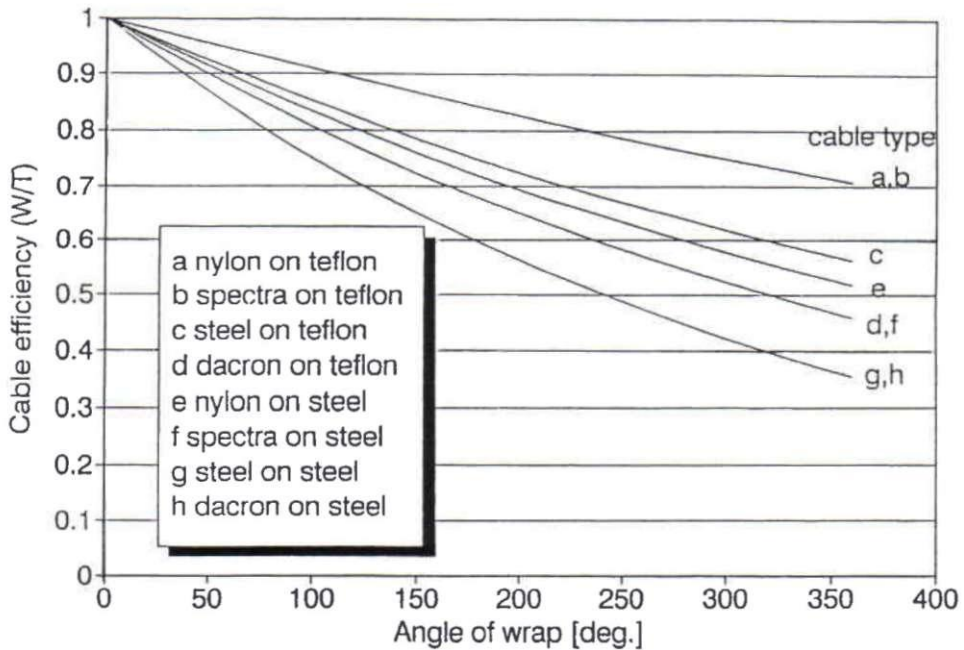


Fig. 2. Cable efficiency.

according to the formula below.

Prehension force (PF) = maximum force (MF) x efficiency of the harness (EH) x efficiency of cable/housing (EC) x efficiency of prehensor (EP) x prehensor ratio (PR).

Using a Hosmer 10X with two rubber bands on a trans-radial prosthesis as an example:

EH = 90% unknown but liberally assumed to be high

EC = 90% measured

EP = 82% measured

PR = 70% measured

Then PF = MF X 90% x 82% x 70% = MF x 46%

Table 9. Prehension force/maximum force for children's prehensors

Prehensor	Prehension force/maximum force
CAPP I (regular spring)	26%
Hosmer 10X (two bands)	46%
Steeper 2.0 hand	12%
NYU child's hand	10%
Adept	50%*

\* Assuming a 95% VC prehensor efficiency.

If the MF a child can generate is 4.55 kg, then the PF to be expected is 2.1 kg.

Taking the measured values for all five prehensors and making the assumptions as in the above example, the relationships are shown in Table 9.

**Strength requirements**

Using the previous calculations for VO

Table 10. Calculated prehension force available given children's strength available for operation

Prehensor	Preh Force/Max Force Percentage*	Max Force Available (kg)			Calculated Preh Force (kg)			Desired Preh Force (kg)			Calculated PF/Desired PF Percentage		
		3 yrs	4 yrs	5 yrs	3 yrs	4 yrs	5 yrs	3 yrs	4 yrs	5 yrs	3 yrs	4 yrs	5 yrs
CAPP 1 (reg spring)	26%	3.1	3.8	5.0	0.8	1.0	1.3	1.8	1.8	1.8	45%	54%	72%
Hosmer 10X (2 bds)	46%	3.1	3.8	5.0	1.4	1.7	2.3	1.8	1.8	1.8	80%	95%	128%
Steeper 2.0 VO	12%	3.1	3.8	5.0	0.4	0.5	0.6	1.8	1.8	1.8	21%	25%	33%
NYU Hand	10%	3.1	3.8	5.0	0.3	0.4	0.5	1.8	1.8	1.8	17%	21%	28%
Adept	50%	3.1	3.8	5.0	1.6	1.9	2.5	1.8	1.8	1.8	87%	104%	139%

\*Maximum force available is calculated from the average of shoulder flexion and shoulder protraction as listed in Table 7. These are the two body control motions usually used for figure of 8 harnesses.

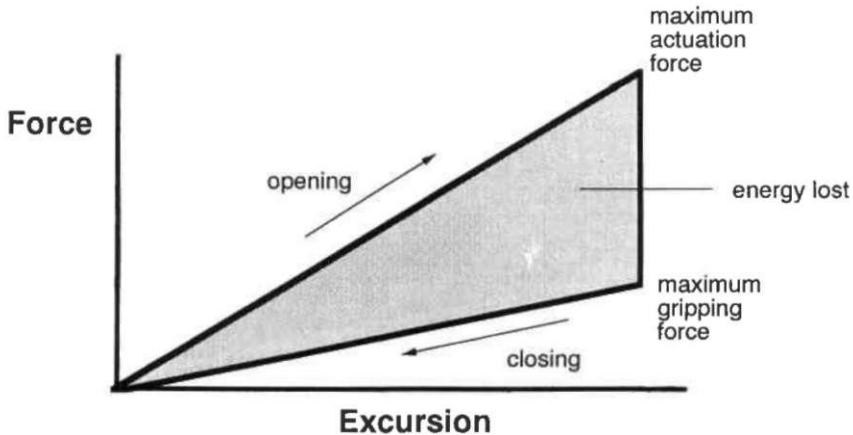


Fig. 3. Efficiency of voluntary opening prehensor.

prehensors and a prehensor force/cable force ratio of 70%, the prehensors compare as shown in Table 10. Calculations indicate the prehension grip which can be expected given the measured strength the children have for operation.

Table 10 refers to desired prehension force as 1.8 kg based on an early study of grip force needed to hold objects used by children at ages 2-5 years (Gottlieb, 1954). Desired grip force factors are explored further in another publication (Shaperman and LeBlanc, 1995).

Inherent differences in efficiency between VC and VO prehensors are shown in Figures 3 and 4. With VO prehensors, the amputee must exert force to overcome the grip force of the closing spring. As noted in Table 8, currently

available hands have lower efficiencies than "non-hands". VO prehensors have low efficiencies and resultant low grip force, and hands are especially inefficient. This leads one to consider the option of the development of VC hands which are inherently more efficient.

#### Discussion

##### *Practical assessment of the formula to estimate prehension force*

In the text above, the authors propose a formula for estimation of prehension force ( $PF = MF \times EH \times EC \times EP \times PR$ ). A practical test was performed to find out how close this formula would come to actual measures in a clinical setting. In this test, PF and MF were measured on 3 five year old limb deficient

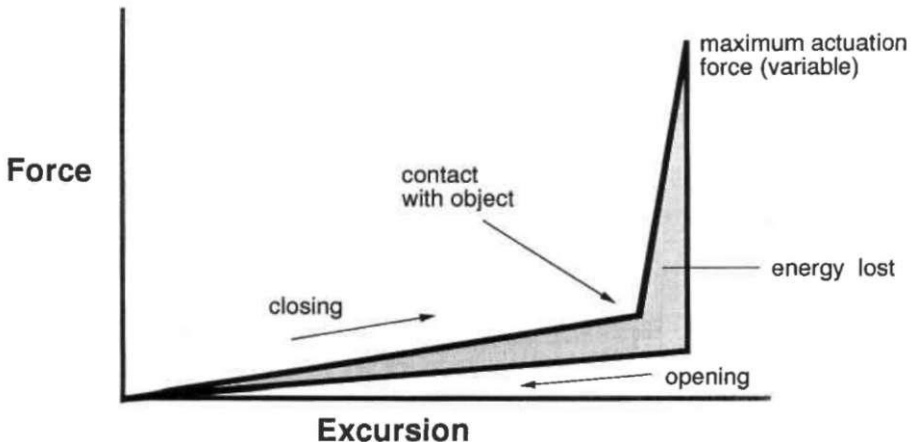


Fig. 4. Efficiency of voluntary closing prehensor.



children using the CAPP I voluntary opening prehensor. Then, PF divided by MF should produce an efficiency factor as shown in Table 9. In the test which was carried out, PF was measured using a specially designed bracket to hold the thumb tip of the CAPP I next to a pinch meter which recorded how strongly the child could push the prehensor open against the meter. MF was measured by averaging strength of shoulder flexion and shoulder protraction. Strength was measured with a myometer. According to Table 9, the expected PF/MF ratio for CAPP I with a regular spring is 26%. However, the children were using the soft spring with a PF/MF ratio of 28%. The three subjects achieved PF/MF ratios of 37%, 49% and 54%. All of these are higher than the expected ratio. The reason for the difference may be that averaging strength of two anatomically measured motions is not an exact representation of the amount of force children exert when performing the control motion. The result might also differ if the actual efficiencies of components of the children's prostheses differ from the estimates in the model. Considering these possibilities, it is likely that the formula gives a conservative estimate of prehension force. Possibly, the outlook is more positive than might be anticipated for developing a body powered prehensor that children can comfortably operate and that gives good grip.

#### ***Children's strength available for operation***

A child cannot exert maximum muscle force and sustain it for a long time. The usable or sustainable force is less than maximum. Studies have shown that muscle action at 15% to 20% of maximum is the level at which repeated muscle action can occur over time without fatigue (Monod, 1985). For intermittent activity, such as using prostheses, the ratio observed by preliminary testing in the pathokinesiology laboratory study was 50%. The usable to maximum force ratio for prosthesis use is not scientifically documented and was not used for calculations of estimated grip force in the formula above. If muscle strength data is used to establish design criteria for a child's prehensor, it is important to consider this limitation. This is especially relevant to VC prehensors where sustained force must be applied for holding objects.

#### ***Upper limb prosthesis strength requirements***

At the meeting of the Advisory Group to this project on 19 May, 1993, it was the strong opinion of parents and professionals that hooks and non-hand prehensors are no longer acceptable to patients and that hands are the prehensors of choice for future development. The Steeper and NYU hands are the only body powered hands presently commercially available for young children and the prehension force/maximum force ratio is very low for each. The efficiencies of the VO hands and the resultant grip forces achievable are very low. For children to achieve a usable grip force requires a very high body force input which they simply do not have. Current VO hands do not appear to be feasible for practical use by young children.

The efficiencies of VC hands are theoretically much higher and give correspondingly higher grip forces than for VO hands for the same force input. However, the authors still are not sure if children have sufficient body power to effectively operate VC hands. The calculations made in this paper must be confirmed by testing. Also, VC prehensors in general have two problems which must be solved for them to be clinically useful: (1) VC prehensors stay open when not in use – which looks awkward and the thumb and fingers can catch on objects; (2) Cable force must be maintained on VC prehensors when grasping and holding an object to keep from dropping it.

#### ***Questions for further study***

Some of the findings of this study raise questions that may be relevant to future research:

(1) How is upper limb strength influenced by wearing a body powered prehensor compared with using a myoelectric limb or no prosthesis at all? In this study, only five children have myoelectric limbs and five others have no prosthesis. Strength of children in each of these groups was compared with strength of the 27 children with body powered prehensors. All of the children with no prostheses were weaker than wearers of body powered prehensors in shoulder flexion and abduction on the limb deficient side, but the differences were less clear when comparing strength of shoulder (girdle) elevation and protraction. Conclusions cannot be drawn from the small number of

children in the two sub-groups, but finding suggest that further investigation may be worthwhile.

(2) Can young children be helped to increase arm and shoulder strength so they will be better able to get good grip with available body powered prehensors? The home supervised activity programme conducted in conjunction with this study did not increase strength enough to improve grip with prehensors, but studies using other interventions might be more successful for increasing strength.

(3) Shoulder flexion is a relatively weak motion. Can ways be found to harness alternate motions with greater strength to serve as power sources for prosthesis operation? Possibly, an entirely different harness design could be created to use shoulder (girdle) elevation alone or in combination with other motion(s). Although the standard system works well for adults, the difficulty children have in using shoulder flexion for prosthesis operation suggests that a new look at ways to harness sources of greater strength may be highly beneficial for them.

(4) Can the data from this study be used to estimate strength of two year old limb deficient children even though their strength cannot directly be measured? Since two year olds have the greatest difficulty getting good grip with body powered prehensors, knowing their strength is important in establishing design criteria for a new body powered prehensor. If it is assumed that strength at age two is related to strength at a later age, and that no strength spurt occurs between age two and age three, the strength of two year olds may be roughly estimated through graphic analysis of percentile data presented in Table 6. This analysis suggests that two year olds at the 50th percentile for strength have the following forces available: shoulder flexion = 2.8 kg, shoulder abduction = 1.4 kg, shoulder (girdle) elevation = 4.0 kg, and shoulder protraction = 3.8 kg. Another important factor to consider besides strength of two year olds is that lower development levels of coordination and cognition may lessen their ability to use their available strength effectively.

(5) How much grip strength does a child's prehensor have to provide? Studies on strength of normal children's hands show that three year olds have full hand grip strength up to 4.3 kg

(Brown, 1973), but studies of grip force needs in children's prehensors suggest that as little as 1.8 kg is sufficient for most activities performed by three year olds (Gottlieb, 1954). Grip may be enhanced by factors such as friction, resilience of grasping surfaces (such as soft tissues of human hands), and prehensor geometry (such as opening span, depth and shape of grasping area). These factors are important to the extent that they may offset the need for high grip force. Children with myoelectric hands appear to use the strong grip force these prehensors provide to compensate for deficiencies in prehensor geometry. The myoelectric experience has not demonstrated that prehensors must provide very high amounts of grip force in order to hold objects securely.

### Conclusions

Although clinicians treating young children with unilateral trans-radial and trans-humeral deficiencies have accepted the concept of early prosthetic fitting, the body powered prehensors available for these children provide poor grip force. New voluntary opening (VO) prehensors, such as CAPP I and voluntary closing (VC) prehensors such as the Adept have been developed to try to improve prehensile function. Today's body powered hands require high operating forces and do not give young children good grip. The enthusiasm for myoelectric hands has been partly due to the need for improved grip, and also for cosmesis.

The purpose of this study was to analyze inefficiencies in body powered prostheses for young children, and to learn whether problems in children's strength prevent them from obtaining maximum function from prostheses. Studies have shown:

(1) Conventional trans-radial systems are relatively efficient, except for body powered hands. The use of today's voluntary opening hands does not appear feasible for young, limb deficient children. VC hands are inherently more efficient, and may be feasible, but this supposition needs to be tested. The major target for improvement is mechanical hands and their cosmetic gloves.

(2) Child amputees have less strength in the limb deficient arm than normal children, thereby making operation a bigger challenge. These findings indicate a need to consider ways of designing a body powered prehensor with

better prehensile function. Studies of effects of friction, resilience and geometry of prehensors might provide better contact with objects to reduce the need for high grip forces. Design criteria have been established for developing a hand with sufficient grip to meet the needs of young limb deficient children.

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## Different approaches and cultural considerations in third world prosthetics

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### Abstract

The major objective of prosthetics the world over is the same, i.e. to restore the amputee to as functional a capacity as possible in his cultural environment, whilst attaining as good a cosmetic result as can be achieved. At first glance it would seem that this would mean there would be very little difference in approach to the subject in western and in third world countries. Availability of materials, resources and skilled personnel, together with a variety of cultural differences, however, make third world prosthetics a subject in itself.

This paper reviews the literature available on the subject, examines some different approaches to prosthetics in the third world, gives an overview of some materials and designs used and considers adaptations for cultural differences. It concludes that, whilst direct transfer of western prosthetics technology is useful in the short term, for long term benefit to the poorer amputees in the third world, culture-specific designs and materials are more appropriate.

### Introduction

A survey has shown that the number of amputees in India in 1981 was approximately half a million (Mohan, 1986); by 1986, although the number of prostheses provided was rising rapidly, the supply was falling further and further behind demand. It was estimated that the number of amputees with no prosthesis was increasing by 17,000 annually. Given that resources for medical care are severely limited in poorer countries and are therefore often used mainly for primary health care affecting a

greater proportion of the population, it would not be surprising if similar figures were reflected elsewhere. The need for prosthetics in the third world is clearly immense.

Golding (1967) has observed that better peripheral circulation, tolerance of minor discomfort and almost total lack of phantom pain actually make the prosthetist's job in the tropics easier in some respects.

The overriding problem in most third world countries is lack of funds, although even where funds are available, direct transfer of western prosthetics techniques is not always successful (Sethi *et al.*, 1978) as cultural differences are not taken into account. There are many aspects to the development of prosthetic services appropriate in these countries, and this paper only touches on some of them.

### Approaches to third world prosthetics

In response to the need for prosthetics, many projects have been initiated both from the industrial world and indigenously. Peizer (1977) divides the former into three categories: research aid, where equipment, finance and facilities are provided for research in the developing country; the introduction of a technique or component into an established prosthetics facility; and long range plans with repeated visits to the host country by a team who teach and set up a facility there. He is a strong advocate of co-operation and co-ordination between different projects seeking to bring "modern rehabilitation treatment" to the third world.

The range of activities initiated from the industrial world is, however, much wider than this. For example, relief teams sent regularly for a few days to Haiti (Frederick, 1992) by the "Adopt a Village" campaign in Florida take impressions and measurements which are used

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to fabricate prostheses in the USA. Mobile army units are used in rural Thailand and computer aided techniques in Vietnam (Keokarn *et al.*, 1992). Representatives from some western agencies stay in the host country for a number of years in order to train local people to a sufficiently high standard to continue to run the facility after their departure (Staros, 1992; Heim, 1979; Fago, 1992). Indeed, Staros (1992) attributes the success of their project to a high level of training and a well planned handover. Heim (1992), on the other hand, found that a five year follow-up of a project in Tunisia revealed little advance in procedures after the expatriates left, and recommends periodic follow-up.

Indigenous projects are also wide-ranging in their scope and clientele, with the urban wealthy receiving a very different service from the rural poor e.g. in Latin America (Jensen, 1979). This has resulted in a debate as to the most appropriate way forward for third world prosthetics. The decision makers in many countries are the influential urban professionals, whereas the majority of the amputees are poorer people, often with little education, and with very little voice in the debate (Sethi, 1989). High technology prostheses made in urban centres have, according to Sethi (1989), "symbolic value as opposed to use value"; they are inevitably too expensive for the majority of amputees.

It has also been reported (Sharma, 1991) that many prostheses and orthoses are not worn because they are considered inappropriate, and that the amputees themselves are the best judges of what is functional. Even when attempts are made to ensure that they allow for cultural requirements, like squatting and sitting cross-legged (Chaudhry *et al.*, 1982), they can be so complex that they require considerable maintenance and are totally unsuited to patients who live any distance from a prosthetics facility. Even within one country there can be those struggling to use indigenous materials such as bamboo (Banerji and Banerji, 1984), while others are making highly technical above knee prostheses with linkages, springs and automatic turntables (Chaudhry *et al.*, 1982).

It is widely held, however, that a high priority should be to develop prostheses which are cost effective, durable, easily maintained by local craftsmen, simple in design and suitable for

local climatic, cultural and occupational needs (e.g. Doshi *et al.*, 1992; Girling and Commings, 1972; Gleeson and Marosszeky, 1992; Sethi, 1989; and Vossberg, 1988).

### Materials used

According to Wilson (1957), the design of simplified limbs for the third world has been a consideration since at least 1954. Little has been found in the literature on materials and designs used for the upper limb, with the exception of a report on body powered limbs in India. Here different task-specific terminal devices are offered, but hooks are preferred by the few patients for whom cosmesis is not the primary function (Narang *et al.*, 1986). This section, therefore, deals with lower limb components and materials used.

### Sockets and shin sections

Polyester resin is commonly used with some success in third world countries e.g. in India (Girling and Commings, 1972; Wollstein, 1972) and Nepal (Meanley, 1991; Vossberg, 1985), but its availability (Meanley, 1991) and reduced shelf life at tropical temperatures (Golding, 1967) make it far from ideal. Traditional material for sockets and exoskeletal limbs or endoskeletal pylons are aluminium (Sethi, 1989), wood (Angarami and Samaria, 1989; Mustapha, 1966), leather and metal (Angarami and Samaria, 1989), and bamboo/cane (Banerji and Banerji, 1984; Kijkusol, 1986). As an alternative for laminating, old X-ray film, acetone gum and stockinette have been used in Indonesia (Golding, 1967), and waste plastic bottles dissolved in thinner are used with gauze bandage in Thailand (Girling, 1972). Thermoplastics have also increased in popularity in recent years, as they have become more widely available (Fago, 1992; Meanley, 1991; Oberg, 1991) and have been used for both sockets and shin sections. Indeed, Oberg (1991) reports that in centres known to him in 14 countries in Africa, South America and Asia, thermoplastics are used alongside more conventional polyester resin and wood. In Argentina, permanently adjustable endoskeletal prostheses have been made using steel and weighing less than 1.8 kg (Angarami and Samaria, 1989).

For soft liners, microcellular rubber and soft leather have been used in Jamaica, India and

Nepal (Golding, 1967; Meanley, 1991), although in Jamaica, there have been problems with the rubber perishing.

#### *Knee joints*

Again, India uses a wide range of materials and displays different degrees of complexity. Banerji and Banerji (1984) report the use of coconut shell, but little detail is given. The opposite end of the spectrum is the knee described by Chaudhry (1982), which incorporates a lever system linking knee flexion with ankle dorsiflexion to enable the amputee to squat smoothly. It also uses an automatically operated turntable to allow for descent into a cross-legged sitting position. The middle path is taken by Girling (1972), who describes a constant friction knee joint made from the axis of the front wheel of a bicycle. Although a factory in the north of the country makes components for prostheses, including knees, it is reported that there are no stance phase control joints available in India (Mensch, 1986).

In Argentina, a low cost single axis knee has been developed using a bronze band bearing on the bolt which forms the axis of the knee. It also has an adjustable extension assist on the anterior of the prosthesis (Angarami and Samaria, 1989).

#### *Feet*

SACH feet appear to be by far the most popular model in the third world. They are easy to make from local materials (Girling and Commings, 1972; Golding, 1967; Kijkusol, 1986; Meanley, 1991; Mensch, 1986; Pe, 1988) and require little maintenance. Different styles of SACH foot have been made from wood and microcellular rubber. These include designs with a toe-break; a wooden keel and a rubber heel and toe (Kijkusol, 1986); and a solid wooden forefoot with a hyperextended toe section. In some cases the foot is fabricated as part of the laminated shin piece (Wollstein, 1972). This technique is also used for a design modelled on a foot with a forefoot amputation. The advantage of the technique is that the limb can withstand immersion in water or mud repeatedly without damage. Fifteen years of follow-up have suggested that the "forefoot amputation foot" is satisfactory (Kijkusol, 1986). Peg-legs are sometimes used, but are often found to be unacceptable to the patients

(Golding, 1967; Heim, 1979). A wooden rocker foot with a tyre sole has been used in rural Thailand (Jivacate and Tippaya, 1992).

The most remarkable innovation in this area, however, is the Jaipur foot (Sethi *et al.*, 1978; Sethi, 1989 and 1991). This is a cosmetically acceptable foot for use by barefoot walkers. It is made using laminated microcellular rubber forefoot, hindfoot and toe pieces, and a laminated wooden ankle piece with an attachment bolt. The structure is reinforced with cord rubber, encased in coloured rubber cushion compound on the dorsum and rubber tread compound on the sole, and can be vulcanised in a hospital autoclave or large pressure cooker. The use of this style of foot is spreading throughout the Indian subcontinent and further afield. Work is, however, still continuing to produce a lighter weight foot (Moll, 1991).

A more bizarre version of this has been described by other workers in Jaipur (Kabra and Narayanan, 1991). In this foot, the laminated microcellular rubber sections are replaced by preserved, defatted and dried cadaveric bone, complete with ligaments and capsules. The authors claim that in laboratory tests the foot compares with western designs of dynamic feet, but is considerably cheaper. The lack of available cadaveric feet was noted, and the authors suggested that it might be possible to use the amputated foot of the patient, this having a psychological advantage "since a part of the anatomy hitherto considered irretrievable is being used for his or her rehabilitation".

#### **Cultural considerations and adaptations**

Many centres are seeking to ensure that the limbs they produce are appropriate to the culture in which they are situated. Although some of these specific requirements are being met, a lot more work is required to fulfil the goal of prosthetics to return the amputee to as functional a capacity as possible within his environment. A few of these considerations and adaptations are described below.

In many countries in tropical areas, barefoot walking or the use of open toed sandals is common, or footwear is removed when entering a home or place of worship. Cosmesis is clearly important in these areas, as well as durability of the foot. The Jaipur foot seeks to address this problem. Indeed, in response to requests by amputees wishing to wear and remove shoes, a



Jaipur foot with removable heel has been made to allow for the heel height of the foot (Sethi, 1989). This foot also allows for squatting, absorbs torque sufficiently for cross-legged sitting of trans-tibial amputees and has enough inversion/eversion to facilitate walking on uneven ground, all of these activities being carried out several times daily in many countries. Due to its dorsiflexion capacities and the grip of the sole rubber, climbing trees is also possible for some amputees (Sethi *et al.*, 1978; Sethi 1989 and 1991); this can be an essential activity for picking fruit or collecting leaves or branches for animal fodder. The spring loaded turntable is an attempt to allow for cross-legged sitting and squatting of trans-femoral amputees (Chaudhry *et al.*, 1982).

In some countries it is customary for younger people to genuflect or kneel before elders, and failure to do this is considered an insult; the use of a knee joint in these circumstances become very important (Oshin, 1981). In addition, simple, durable designs are vital for rural communities where patients may live several days' walk from the prosthetics centre, and cannot afford time away from the fields to attend a clinic for repairs; local craftsmen may find themselves doubling as prosthetics technicians in this situation. The use of plastics or other materials which will neither rust nor rot is important in communities where much walking is done through mud, and wading through rivers is a daily occurrence (Sethi, 1989; Meanley, 1991).

Clearly, when cultural requirements such as squatting, sitting cross-legged, tree climbing and kneeling are considered, it can be seen that western prosthesis designs will not be appropriate in such settings. Local conditions such as uneven terrain, the need to walk through rivers and mud, the presence of pests which eat wood, rubber etc., will also have an effect on the design and materials used.

Finally, "preventative prosthetics" might be an appropriate line to take in some areas. For instance, in Cambodia, which has the highest number of amputees per capita in the world, the main cause is explosion of land mines left over from the war. The majority of casualties are civilians. Detection and destruction of the remaining mines, though a difficult task, would clearly reduce the number of amputees. In India, most upper limb amputations are the

result of a poor understanding of the use of new farming machines (Narang *et al.*, 1986; Vohdra, 1992); training in their use and the importance of safety procedures would be beneficial here.

### Conclusion

The need for prosthetic services in third world countries is immense and although much is being done to address this problem, there is sometimes a lack of co-ordination (Peizer, 1977). There is the "relief" approach, which seeks to meet some of the immediate need, and has its successes, and there is the "development" approach, which looks to work for more lasting solutions, these being both indigenous and western initiated. Although this paper emphasises the "development" angle, "relief" work is also acknowledged to be valuable in improving the functional capacity of many people.

The direct transfer of western prosthesis design to the third world is not considered to be appropriate in the long term for two main reasons. Materials and equipment can be prohibitively expensive in many countries, so that their use can lead to aid dependency rather than independence. Secondly, local conditions and customs are not taken into account in their designs so they can be inappropriate in many settings.

There is much to be done to improve both designs and availability of prostheses in the third world, but perhaps the words of Sethi (1991), the designer of the Jaipur Foot, will prove to be the key to his and others' success in this area: "Need for alterations kept on arising and were ready to respond to them".

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## A comparison of trial shoe and shell shoe fitting techniques

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### Abstract

In Europe, bespoke orthopaedic shoes are usually sent for a trial fitting in order to check the fit and indicate any modifications required before final finishing. The use of shell shoes at the fit assessment stage, rather than the traditional alternative of partially or fully finished shoes, can offer service advantages, and is widely used for example in the Netherlands. However the comparability of shell fit assessment with the traditional method of trial shoe fit assessment has not been evaluated, either to assess its sensitivity or to elucidate any difference in assessment technique required of the orthotist. In this work, the results of fit assessments by both methods are compared. The trial involved a group of normal subjects wearing high street shoes of styles similar to those used for orthopaedic footwear. The results indicate that the shell fit assessments were in the main comparable to those for trial shoe fit. The only consistent area of deviation noted, in the heel at the topline, is attributable to a construction factor in shoe making. Apart from this area, the orthotist need not adjust his technique to make use of the shell method.

### Introduction

In the orthopaedic shoe trade in the UK, bespoke shoes are made to measure or from casts, and usually sent for fitting at the stage of rough finishing, i.e. with the uppers tacked in place and a temporary sole attached. However in continental Europe fit assessment is often made on the basis of a shell shoe, made by

vacuum moulding PVC materials over the shoe last (the model over which the shoe is constructed) to form a temporary shoe (Fig.1). This has the advantage that the shoe need not be constructed before fit assessment is made, which reduces both the time to the first fit and materials wastage in achieving the final shoe. Because the shoe last shape is adjusted before patterns and shoe uppers are cut, more complex styles can be attempted with confidence.

Questions remain as to the comparability of fit assessments made using these two techniques. Does the person performing the fitting have to make allowances for the two methods? To what extent does an assessment made with a shell accurately indicate the fit of the final shoe? The process of fit assessment by either method is a skill rather than a science, which reflects the basic lack of quantification of what constitutes a good fit (Rossi, 1983). Fit assessment is an area of considerable impact on the volume shoe trade, and one where increased effort has been expended recently in view of the trend towards more quantitative descriptions needed for computer aided design systems (Browne, 1993; van der Zande *et al.*, 1995).

There are two groups of factors which affect

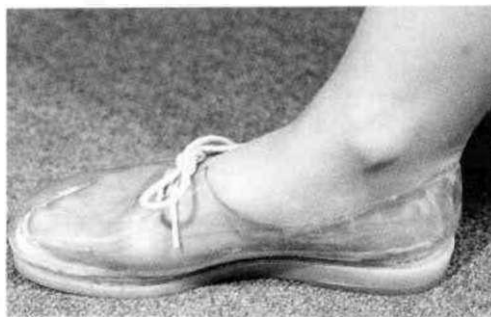


Fig. 1. One of the shell shoes made for this trial.

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the perceived fit of a pair of shoes on any individual. The most obvious group are attributable solely to the shoe, and relate to its dimensions and material properties — the shoe related factors. The last shape and shoe construction are two of the most important factors in this group. Shell fitting differs from trial shoe fitting in that the last shape factor may be identical but the construction factor is not. Also important to fit however are a group of factors relating to an individual's requirements — the subject related factors. These encompass the degree of flexibility of the foot, subjective preference for tightness, and pathology giving special problems such as hypersensitivity. The perceived fit of shell shoes may be affected by the different and unfamiliar feel compared to ordinary shoes.

In this investigations the authors set out to study the shoe related factors in making a fit assessment. This was done mainly by noting the fitter's assessment. The specific objectives of the study, part of a doctoral thesis (Chen, 1993) were:

- to document a procedure for assessing shell shoe fitting
- to compare assessment of fit by shell shoes and normal shoes for normal subjects
- to identify limitations of shell shoes fitting
- to separate fit factors due to last shaping from those due to shoe construction.

#### **Fit assessment procedure**

The assessment of fit is of its nature subjective. However, extensive fit assessments are routinely made in the volume trade before a new model of last and shoe is approved for production. Therefore considerable experience resides in the fitting departments at shoe manufacturing companies (as opposed to the limited skill in shoe shops). This expertise was tapped for the study. The protocol described below was derived from the fit assessment procedure used at C & J Clark International, Street, Somerset, UK. The method is also compatible with the British Standard 5943 (1980) Methods for Measurement and Recording for Orthopaedic Footwear.

The foot is first placed into the shoe and the shoe is then firmly fastened. Fit is assessed in the following areas during partial weight bearing as defined in British Standard 5943 i.e. with the subject seated, the shin vertical and the

weight of the limbs applied through the feet.

#### *Forepart*

**Length allowance** – Assess the effective length in front of the toes by pressing (shoes) or viewing (shells) and compare with the standard of around 8 mm for fashion shoes, or up to 15 mm for orthopaedic shoes. The effective length extends to where the shoe is still deep enough to accommodate the toes, which may exclude the end part in pointed or shallow toe boxes.

**Forepart width** – Check the width of the shoe across the joint. First locate the joint of the foot by palpation; this is the widest part of the forepart, running from the first to fifth metatarsal heads. The width in the forepart is correct if there is no excessive pressure across the joint or empty space to the sides of the foot.

**Alignment** – Check that the foot shape is aligned correctly in plan view in the forepart of the shoe and there is no centrally directed pressure on the big toe and the smallest toe.

**Forepart depth** – Squeeze the vamp area of the shoe across the joint inward from the medial and lateral side walls. If there are too many creases at the vamp of the upper, the forepart is too deep. Check that the forepart is not too tight (may also be the result of insufficient width). Check that there is sufficient clearance on the toes by palpation (shoes) or visually (shells).

**Heel-to-ball length** – Ensure that the ball of the foot is correctly positioned in the shoe. In this position, the joint of the foot should be aligned from the medial side to the widest part of the shoe. If the heel-to-ball measure of the shoe is too long, there will be a gap between the heel and the backseam of the shoe. If it is too short, the heel will be forced uncomfortably back in the shoe or the ball of the foot will be forced too far forward in the shoe.

#### *Midfoot*

**Waist fit** – Assess the fit of the waist especially checking the arch area. Check both the medial or lateral areas by pressing on the shoe/shell.

**Instep fit** – Record the facing gap or overlap and check it with the original design (shoes). For shells, the cut line at the facings should

just meet.

### Quarters

**Topline** – Observe the topline, i.e. the opening around the ankle. Feel with fingers along the front section of the topline to make sure it fits neatly against the foot.

**Under ankle height** – Observe any pressure on the medial and lateral malleoli. The malleoli must be clear of the topline, although this may not be necessary if the topline is padded.

### Backpart

**Seat width** – Assess whether the width of the heel seat is adequate. If the heel can be rocked in the shoe, the seat may be too wide – if the foot is too wide for the seat, it will tend to flatten the sides and cause gaping at the topline under the ankle.

**Heel curve** – Observe any excessive pressure or gaping at the top of the back seam (or in the case of a shell, the notional position of the back seam).

**Heel grip** – This final assessment is done initially during walking. First observe any heel slip which occurs during walking. Then ask the subject to sit down, lift the foot, and pull firmly down on the shoe heel which should not slip. Note if there appears to be excessive grip pressure from indentation of the skin.

It is not deemed possible to categorise fit more accurately than to a five-point scale. Each feature was put into one of these categories:

- UA– too tight/small
- AO– adequate: on the tight/small side
- OK good fit
- AO+ adequate: on the loose/large side
- UA+ to loose/large

### Shell shoe making

Shell shoes were made over the production shoe lasts for each of the models of shoe selected. These were made of a transparent material using a fairly stiff 500  $\mu$  PVC for the heel area and side walls, and more flexible 200  $\mu$  PVC over the top of the vamp. The shoe last was mounted bottom up in a vacuum moulding machine with the shoe insole already in place. The thinner stiff PVC was vacuum moulded over the bottom and sides, Figure 2.



Fig. 2. The first part of the PVC shell is formed by moulding over the bottom of the last: an example sole unit is also shown.

The sides were then trimmed *in situ* to leave the heel area and the side walls extending down the quarter and vamp right to the toe. Small 'v' notches were cut into the side wall to facilitate flexing at the metatarsal break during walking. Glue was applied around the edge of the walls, the last was turned upright, and the softer thicker PVC was vacuum moulded over the top of the last. This forms a closed shell. The same sole unit as used on the trial shoe was then attached, and the production insole inserted.

The top line of the uppers was trimmed consistently according to a set of geometric construction rules used by Dutch orthopaedic shoe-makers, which results in a standardised backseam height, under-ankle height and vamp point (the point corresponding to the base of the lace panel in standard Gibson style shoes). The vamp was split to allow for foot entry, and small holes were punched into the PVC to form a mock lace panel.

### Trial protocol

**Shoes:** Four styles of Clarks shoes were chosen, representative of typical styles which could be used for orthopaedic footwear, i.e. low heeled shoes (heel height lower than 4 mm) with lace fastening over the instep (Fig. 3). These styles were named 2nd Nature, Nocturne, Ohio and Pop-life. The lasts on which these styles were made were all different shapes: the 2nd Nature has a 'natureform' shape with a straight medial border and wide round toe box, the Ohio style is a moccasin, while the other two were more traditional designs. Standard last measures were taken to give evidence of the differences in designs and for further studies of



Fig. 3. This shows the natureform trial shoe and its matching shell.

the allowances between foot and last measures, although these are not discussed further here.

**Subjects:** Asymptomatic female subjects were used in the study. These were drawn from the usual staff volunteer panel used for assessment of new models of shoes at a large UK volume shoe manufacturers, C & J Clark International, Street, Somerset. The subjects are all deemed to represent average customers having no reported foot problems. There were ten subjects of nominal size UK 5D, continental 38, seven of whom tested each of three styles (2nd Nature, Nocturne and Ohio); the three remaining subjects tested only two styles because one style was not available at the time. A further eight subjects of nominal size UK 4E, continental 37, tested a single different style (Pop-life). Eight subjects were in the age group

16-25, five subjects in the age group 26-35, three subjects in the age group 36-45 and one subject each in the age groups 46-55 and 56-65. No further selection criteria were used.

**Procedure:** Each subject was brought to the fitting room, the trial procedure was explained and verbal consent gained. The feet were then measured by one of the authors, RCC, according to BS 5943. Assessments of shoes or shells, and different styles where applicable, were carried out in a random order. Although it would have been preferable to separate the two assessments in time, this was not feasible because of the time constraints on the subjects. All assessments were made by the senior fitter at Clarks (JT), with RCC recording the results. Additionally, spontaneous subjective comments regarding fit were noted.

## Results

### Foot measurements

The foot measurements are shown in Table 1. In the first section on foot length, the difference in foot length measures between the nominal UK sizes 4 and 5 subjects is as expected: an increase of approximately 10 mm (4.4%) in the average foot length corresponds to the standard shoe length increment of 8.5 mm ( $\frac{1}{3}$  inch) per size. The average joint girth differed by 3 mm (1.3%) compared to a full width size of 6.5 mm; in the UK sizing system, the 5D and 4E shoes nominally have the same girth. Other girth averages are comparable between the two sizes i.e. less than a full width difference, and no consistent differences in the heights taken were noted.

The wide ranges in the measures may appear large for subjects nominally the same size, representing for example  $\pm 3\%$  of stick length,

Table 1. Foot measurements in the groups 4E and 5D in millimetres.

Foot measurements	4E Subjects, n = 8		5E Subjects, n = 10	
	Average	Range	Average	Range
Foot (stick) length	228	220 - 233	238	232 - 242
Medial heel to ball	169	165 - 173	178	171 - 189
Joint girth	227	219 - 239	230	221 - 238
Waist girth	218	211 - 227	222	212 - 228
Instep girth	230	221 - 242	234	228 - 237
Short heel girth	302	285 - 317	309	300 - 320
Height over 1st metatarsal head	36	32 - 37	35	30 - 38
Instep height	57	50 - 62	58	49 - 64
Joint width	90	86 - 96	91	86 - 94



and  $\pm 4\%$  of joint girth. These ranges however are of the order of one full size or width fitting, and, due to the complex combination of measures and foot shape that produce a given nominal size, the ranges are not dissimilar to other (internal) survey data from Clarks.

#### Fit assessment

An example chart collating the results for all forefoot width fit assessment is shown in Table 2; this demonstrates the closeness of the assessments for shell and sample shoe fit. In all except two cases the fit is in the same category, and then these two cases are in adjacent categories.

Note that the majority of fit is in the central categories, which is expected since these subjects were fitted with shoes of their own

nominal size. Some of the shoes were deemed too tight for the subjects, but none too loose. It is also apparent that the Nature form design was looser in the forepart on average, which corresponds to its wider design.

A summary chart for all assessments of all features (Table 3) indicates that the majority of the fits were adequate, and for most features, a good fit was seen at both shell and trial shoe assessments. Again, the number of assessments in each category comparing shoe or shell fit are remarkably similar, differing by only one except for a trend in the heel area where the shells were assessed to be looser.

#### General observations

Most of the subjects reported that the shell shoes felt slightly bigger than the trial shoes.

Table 2. Example of chart for forefoot width fit, all assessments

Subject	Shoe Style	UA-	AO-	OK	AO+	UA+
4E01	Pop-life		*#			
4E02	Pop-life		*#			
4E03	Pop-life	*#				
4E04	Pop-life			*#		
4E05	Pop-life			*#		
4E06	Pop-life		*#			
4E07	Pop-life		*#			
4E08	Pop-life	*#				
5D01	Nocturne Ohio 2nd Nature		*#	*# *#		
5D02	Nocturne Ohio 2nd Nature		*	# *# *#		
5D03	Nocturne 2nd Nature		#	*	*#	
5D04	Nocturne 2nd Nature		*#		*#	
5D05	Nocturne 2nd Nature		*#	*#		
5D06	Nocturne Ohio 2nd Nature		*# *#	*#		
5D07	Nocturne Ohio 2nd Nature			*# *# *#		
5D08	Nocturne Ohio 2nd Nature			*# *#	*#	
5D09	Nocturne Ohio 2nd Nature		*# *#	*#		
5D10	Nocturne Ohio 2nd Nature			*#	*# *#	

\* indicates trial shoe fit, # indicates shell shoe fit.

Table 3. For all assessments of all features, the number occurring in a given fit category

Features		Number of assessments (n = 35)				
		UA-	AO-	OK	AO+	UA+
Length allowance	Shoes	2	2	26	4	1
	Shells	2	1	27	4	1
Forepart width	Shoes	2	12	16	5	-
	Shells	2	11	17	5	-
Alignment	Shoes	-	-	29*	-	-
	Shells	-	-	29*	-	-
Forepart depth	Shoes	3	8	16	8	-
	Shells	3	8	16	8	-
Heel-to-ball length	Shoes	1	6	28	-	-
	Shells	1	6	28	-	-
Waist fit	Shoes	1	6	25	3	-
	Shells	1	5	26	3	-
Instep fit	Shoes	3	8	21	3	-
	Shells	3	7	22	3	-
Top line	Shoes	1	2	30	2	-
	Shells	1	2	30	2	-
Under ankle height	Shoes	-	-	35	-	-
	Shells	-	-	35	-	-
Heel seat width	Shoes	-	5	30	-	-
	Shells	-	5	30	-	-
Heel curve	Shoes	-	1	32	2	-
	Shells	-	1	28	4	2
Heel grip	Shoes	-	3	30	2	-
	Shells	-	1	10	20	4

\* it was not possible to assess this feature in six cases where the foot was too wide for the shoe and caused distortion of the forepart.

With shell shoes, white patches on skin were seen on almost every subject, even where fit was satisfactory to the experienced shoe-fitters and subjects. At the topline point on mid-line of the forepart cone, the shells exhibited pressure to one side, at the medial (instep) dorsum of the foot. Pressure was also seen in the heel area and around the joint.

### Discussion

The main purpose of this trial was to compare the results of shell shoe fitting with those of trial shoe fitting. The results indicate that, for these subjects and shoes, the outcome of the two methods is very similar.

Only in the region of the heel were there any differences of note. The majority of the shell assessments were one category looser than the trial shoe assessments. This was not unexpected: it is normal practice to apply a 'heel clip', or removal of material, to the shoe upper patterns at the topline in the region of the heel backseam so that an adequate heel grip is obtained. A shell shoe obviously cannot incorporate this feature.

On the whole the subjects reported that the shoes were tighter than the shells. This

phenomenon might be attributed to any of the following possibilities:

- Where a shoe is formed by machine pulling the leather upper over the last, shrinkage occurs after the last is pulled out of the shoe; all the assessed shoes were made by machine lasting methods, and hence they would be slightly smaller than the last. In contrast, bespoke orthopaedic shoes are infrequently machine lasted.

- Different materials caused different sensations to subjects. Although shells are made from soft PVC material, it is not soft enough to mould to the foot closely. In addition the surface of PVC is too smooth to grasp the foot. This may cause some feelings of looseness for the subjects.

Many orthopaedic companies have experimented with, or use, alternative materials for making shells which more closely resemble both the feel and compliance of leather. These may be superior in respect of sensation although they are not transparent and do not allow visual inspection.

The shells allowed the fitter to observe the regions in which pressure is applied to the dorsum, thereby causing the skin to whiten by

occluding blood supply. It appears that such pressure on skin is tolerable to the normal foot. It would be instructive to define what level of pressure causes whitening vs. tolerable pressures on tissues, coupled to the limits of sustainable pressure and duration. Although this type of data is available for tissues involved in pressure sore formation, they are not known for the foot as yet. It is also noted that pressure levels tolerated on a normal foot might not be permissible in pathological conditions. This is an area in shell shoe fitting where only experience can at the moment be applied.

From the results, it would appear that of the shoe related fit factors, the construction factors are secondary to the last shape in determining initial fit. However, the shell cannot give any indication of problems which might arise due to poorly located seams, stitching or leather stressing. Normally, unlike fashion shoe styles, orthopaedic shoe styles are carefully controlled to avoid any possibility of these problems arising in any case.

The objective of this research is to provide information of use to the orthopaedic service. It is valid therefore to query whether a trial of normal shoes on normal subjects reflects the potential of shell shoe fitting for orthopaedic cases. It is already known, however, that the method is used successfully in European countries for fitting of bespoke orthopaedic shoes. The research primarily indicates that the fitting with shell shoes needs little modification to the orthotist's technique, since both shell and trial methods gave the same result. That is to say, the only compensation needed in interpretation of the fit is in the area of the heel grip.

### Conclusion

In the process of supply of bespoke orthopaedic shoes, assessment of fit by shell

shoes offers a method to improve service delivery. This research indicates that fit assessment by shell shoes provides very similar results to that by trial shoes, except in the area of heel grip where the fit of the shell shoes is one category looser. The orthotist need not otherwise adjust his fitting procedure to take advantage of this technique.

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## Effect of push handle height on net moments and forces on the musculoskeletal system during standardized wheelchair pushing tasks

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### Abstract

The aim of this investigation was to analyze the external forces and biomechanical loading on the musculoskeletal system during wheelchair pushing, in relation to different push handle heights. In addition, recommendations for wheelchair pushing in accordance with push handle height are made.

Eight young, female subjects carried out three different wheelchair transport tasks at five different push handle heights in a standardized laboratory setting. Five pushing heights were selected as a percentage of the subjects shoulder height (61, 69.5, 78, 86.5 and 95%). All three wheelchair transport tasks investigated required higher pushing handles in order to minimise net shoulder moments and external vertical forces on the hands. When pushing a wheelchair on to a pavement, net moments around wrists, elbows, shoulders compression and shear forces at L5-S1 and external vertical forces were lower using higher pushing heights. When low pushing handles are used, elderly female attendants are at risk of L5-S1 low back pain when lifting and pushing the wheelchair on to a pavement. A recommendation is made to reconsider height and position of the pushing handles of attendant propelled wheelchairs. For the investigated tasks, a pushing height of 86.5% (1.191 ± 0.034 m) was most favourable.

### Introduction

Pushing and pulling actions during manual materials handling are associated with low back

problems, as indicated by Pope (1989), who stated that 20% of the low back problems in the USA are related to pushing or pulling activities. Pushing and pulling tasks are regularly performed by a variety of professions, among which are: refuse collectors (Jager *et al.*, 1984), postmen (Haisman *et al.*, 1972), truck drivers, miners (Williams *et al.*, 1966) and nurses or orderlies (Harber<sup>1,2</sup> *et al.*, 1987; Winklemolen *et al.*, 1994).

Several physiological or psychophysiological studies of pushing and pulling tasks have been conducted (Haisman *et al.*, 1972; Strindberg and Petersson, 1972; Sanchez *et al.*, 1979; Ciriello and Snook, 1983; Snook and Ciriello, 1991). Also, biomechanical studies have addressed pushing and pulling tasks, predominantly in relation to manual materials handling (Lec *et al.*, 1992; Kerk *et al.*, 1994; Wolstad *et al.*, 1994).

Among nurses, musculoskeletal disorders are a common cause of sick leave. Estryn-Behar *et al.* (1988) examined the causes of sick leave among 1505 female hospital workers. The main causes of sick leave were musculoskeletal disorders which affected 16% of the population during the previous 12 months. Of the investigated population 47% described back pain in the previous year. Several aspects can be identified which contribute to the heavy load of this occupation. In recent research lifting patients is considered to be the main cause of low back pain (Winkelmolen *et al.*, 1994). However, other tasks such as transportation of patients in wheelchairs must be considered as well, since this task is performed during a great part of the working day (Harber<sup>1,2</sup> *et al.*, 1987). Harber<sup>1,2</sup> *et al.* (1987) investigated the relationship between nursing activities and the occurrence of back pain. Carrying and pushing

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were the only tasks significantly associated with occupational back pain. Nurses and orderlies are not the only persons who transport patients in wheelchairs. In many hospital settings porters transport patients to the various hospital departments for treatment. In domestic environments husbands, wives and other family members push wheelchairs. Persons depending on attendant propelled wheelchairs in their home environment are usually older than 65 years as will be their companions, who may thus experience difficulties manoeuvring attendant push wheelchairs, especially outdoors (Abel and Frank, 1991). Apart from personnel, also volunteers in institutions, mostly middle aged, consider pushing wheelchairs a heavy task (Stephan, 1990).

The majority of wheelchair studies are focused upon the optimization of the wheelchair-user combination of manually propelled wheelchairs (van der Woude *et al.*, 1988, 1989; Veger *et al.*, 1991, 1992; Rodgers *et al.*, 1994). In contrast, investigation of attendant propelled wheelchairs is scarce (Stephan, 1990, 1992; Abel and Frank, 1991). In pushing an attendant propelled wheelchair, forces are applied to the handles at the back of the wheelchair to overcome rolling resistance, internal friction and effects of gravity. Preferred positions for wheelchair push handles have been suggested to be in the region of 75% of shoulder height and 1.14 times shoulder width (Abel and Frank, 1991). However to date, experimental analysis has not revealed why some handle positions may cause more strain or be less comfortable for wheelchair pushing than others (Abel and Frank, 1991). Further biomechanical analysis is necessary to draw conclusions on the optimum push handle height in terms of preference and with respect to the level of forces and moments in the upper body joints and trunk (Abel and Frank, 1991). Thus risks for low back problems or musculoskeletal disorders in general may be more readily discerned.

In order to investigate the possible risks of wheelchair pushing tasks on the musculoskeletal system, the current study analyses the external forces and biomechanical loading of the musculoskeletal system during a limited number of standardized wheelchair pushing tasks at different push handle heights. For this purpose eight young, female subjects

carried out three different wheelchair transport tasks at five different handle pushing heights in a standardized laboratory setting. In order to calculate the biomechanical loading on upper body joints and the low back, a dynamic two-dimensional linked segment model was used (de Looze<sup>1,2</sup> *et al.*, 1992).

## Methods

### Subjects

Eight healthy female subjects participated in this study on a voluntary basis (age  $23.9 \pm 6.3$  years, weight  $58.6 \pm 4.0$  kg, height  $1.69 \pm 0.03$  m). The subjects were considered to be physically representative of the nursing population. None of the subjects had a previous history of musculoskeletal disorders of the upper limbs or back. Moreover, none of them had more than an incidental experience of transporting persons in wheelchairs. Informed consent was signed prior to the experiment.

### Experimental tasks

Three wheelchair pushing tasks were performed, each of them at 5 different push handle heights.

Thus, each subject performed fifteen trials, according to a standardized procedure. The three tasks were performed under laboratory conditions with an instrumented test wheelchair, loaded with an ISO-dummy, on a standardized test circuit. They were as follows:

1. increasing the velocity of the wheelchair from zero to walking speed starting on a flat circuit and finishing on the higher part of the circuit, using a slope of  $6.74^\circ$  (flat pushing);
2. increasing the velocity of the wheelchair from zero to walking speed starting in front of the slope up to the higher platform (slope pushing);
3. tilting the wheelchair backwards before pushing and lifting it on to the curb up to the higher platform (lifting).

The five pushing heights were chosen as a percentage of the subjects shoulder height. Pushing heights were 61, 69.5, 78, 86.5 and 95% of shoulder height. These percentages were based on realistic proportions between female body length and frequently used wheelchair pushing heights. The Joint Medical Services in the Netherlands suggest a push handle height of which equals the elbow height of the attendant and allows push handle heights

for wheelchairs between 0.9 m and 1.2 m (GMD, 1992). The highest push handle height of the experimental wheelchair was determined from shoulder height of a 5th percentile (of body height) Dutch woman and a 1.2 m handle height. The lowest push handle height was derived from the shoulder height of a 95th percentile Dutch woman and a 0.9 m handle height (Molenbroek and Dirken, 1986). The other percentages were chosen at regular intervals between the highest and the lowest percentage.

The sequence of pushing heights was randomised. The sequence of the tasks was as mentioned before. All subjects were trained to perform the tasks at a certain pace, thus to a certain extent standardising walking velocities. The first task was performed in about 5.8-6.3s, the second task in 5.2-5.8s, the third task in 5.8-6.9 seconds. The subjects were also trained to keep the arms in the sagittal plane as much as possible.

One subject performed an additional series of trials using the 78% push handle height, while the ISO-dummy was replaced by a young male person of 76 kg.

### **Anthropometry**

Before starting the experiments, age, body weight, body height, and shoulder height were measured. The application of the linked segment model (de Looze<sup>1,2</sup> *et al.*, 1992) requires individual anthropometric data of trunk, head, upper arm, forearm and hand. Therefore, segment length, volume, mass, centre of gravity and moment of inertia were established using the regression equations of Young (1983). Thus, for each subject 18 anthropometric measurements were obtained in order to apply these regression equations. The distal section plane of the trunk according to Young is positioned at the level of the L3-L4 intervertebral disc. One of the aims of the present study was to measure net forces and net moments at the L5-S1 level. Therefore it was necessary to establish the anthropometric data of a segment bounded by the L3-L4 and the L5-S1 level. This was done by means of the method described by Yeadon (1990).

### **Linked segment model**

A dynamic two dimensional Linked Segment Model (LSM) was used for calculation of net

reaction forces and net joint moments (de Looze<sup>1,2</sup> *et al.*, 1992). These calculations were made for the wrists, elbows, shoulders and L5-S1 intervertebral disc centre. For L5-S1 the compression forces and the shear forces were also calculated. The LSM is based on inverse dynamics. The segments of the body are represented by linear rigid links connected by joints. The two dimensional model was built of links representing the hands, lower arms and upper arms, the head, trunk and pelvis. Newtonian principles are applied in order to calculate net forces and moments working upon every joint:  $\sum F_y = m \cdot a_y$ ,  $\sum F_x = m \cdot a_x$ ,  $\sum M = I \cdot \alpha$

Using external hand forces from the instrumented push handles of the experimental wheelchair, position data and anthropometric data, net joint forces and net moments were calculated. The results of these calculations are net joint forces and net moments for two hands, forearms and upper arms. Compression forces of L5-S1 were calculated assuming that extensor muscles of the lower back exert their resultant force at a distance of 0.062 m from the centre of the L5-S1 disc (Nemeth and Ohlsen, 1985; Susnik and Gasvoda, 1986). During positive resultant forces of the abdominal muscles the compression force values were set at zero.

Reflective markers were placed on the right side of the head (just in front of the tragus), seventh cervical vertebra, shoulder (lateral part of the spina scapulae), elbow (lateral humeral epicondyle), lower arm halfway between the elbow and wrist (ulnar styloid process), L5-S1 as seen in sagittal plane, hip (upper margin of trochanter major) (de Looze<sup>1,2</sup> *et al.*, 1992) side of the pushing handle, and finally two on the frame of the wheelchair, placed in a vertical line. While the subjects performed their trials, the marker positions were recorded with a video-based 3-dimensional motion registration (and analysis) system (VICON@; 4 camera's; sample frequency: 60Hz).

### **Wheelchair and circuit**

The experimental wheelchair (Fig. 1) was a foldable attendant push wheelchair (Poirier 3 A 41; weight 20 kg; height push handles: 0.905 m; front wheel size 0.20 x 0.05 m; rear wheel size 0.30 x 0.06 m; front tyre pressure 300 KPa; rear wheel pressure 250 KPa; rolling resistance of ISO-dummy loaded wheelchair on a motor



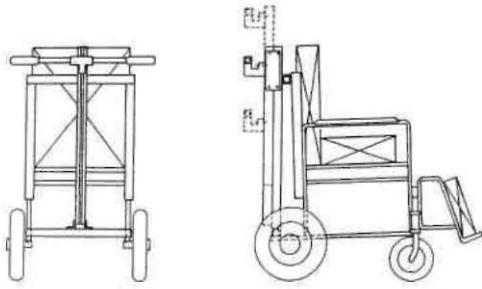


Fig. 1. Experimental wheelchair with pushing bar, which is adjustable in height.

driven treadmill:  $7.89 \text{ N}$ ,  $v=1.11 \text{ ms}^{-1}$ ). A special purpose push handle replaced the standard push handles and was mounted on the experimental wheelchair on a special frame which allowed adjustment of the push handle in a vertical direction over a sufficiently large range (total weight wheelchair was 30kg). The push handle was instrumented with a set of two dimensionally arranged strain gauges allowing the measurement of the horizontal and vertical force components in the sagittal plane ( $F_y$  and  $F_z$ ). Total width of the push bar was 0.58 m. Rubber hand grips were placed at the ends of the bar (diameter: 0.035 m). The experimental wheelchair was loaded with a 75 kg dummy according to ISO/DIS 1776-11, fixed to the wheelchair.

The test circuit (Fig. 2) was made up of two parts, each consisting of series of 4 x 8 paving stones, each measuring 0.30 x 0.30 m. The two parts differed 11.5 cm in height. The higher part was placed in series with the lower part. A removable board of 1 m length was designed to connect the lower to the higher part (slope  $6.74^\circ$ ).

### Data processing

The position data of the VICON-system were synchronized with the external forces from the calibrated force transducers of the push handle. Data of positions and forces were filtered with a digital low-pass second order recursive Butterworth filter with an effective cut-off

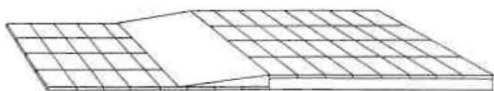


Fig. 2. Test circuit consisting of paving stones, the higher part connected to the lower by a removable board.

frequency of 5 Hz. In order to allow visual verification of the data during data processing, all trials were also recorded on ordinary video tape.  $F_y$  and  $F_z$  forces were calculated in fixed horizontal and vertical directions.

Mean and peak values of net external forces, net joint moments (wrists, elbows, shoulders, L5-S1) and L5-S1 compression and shear forces were determined with the two dimensional dynamic linked segment model for all trials, three tasks and eight subjects.

### Statistics

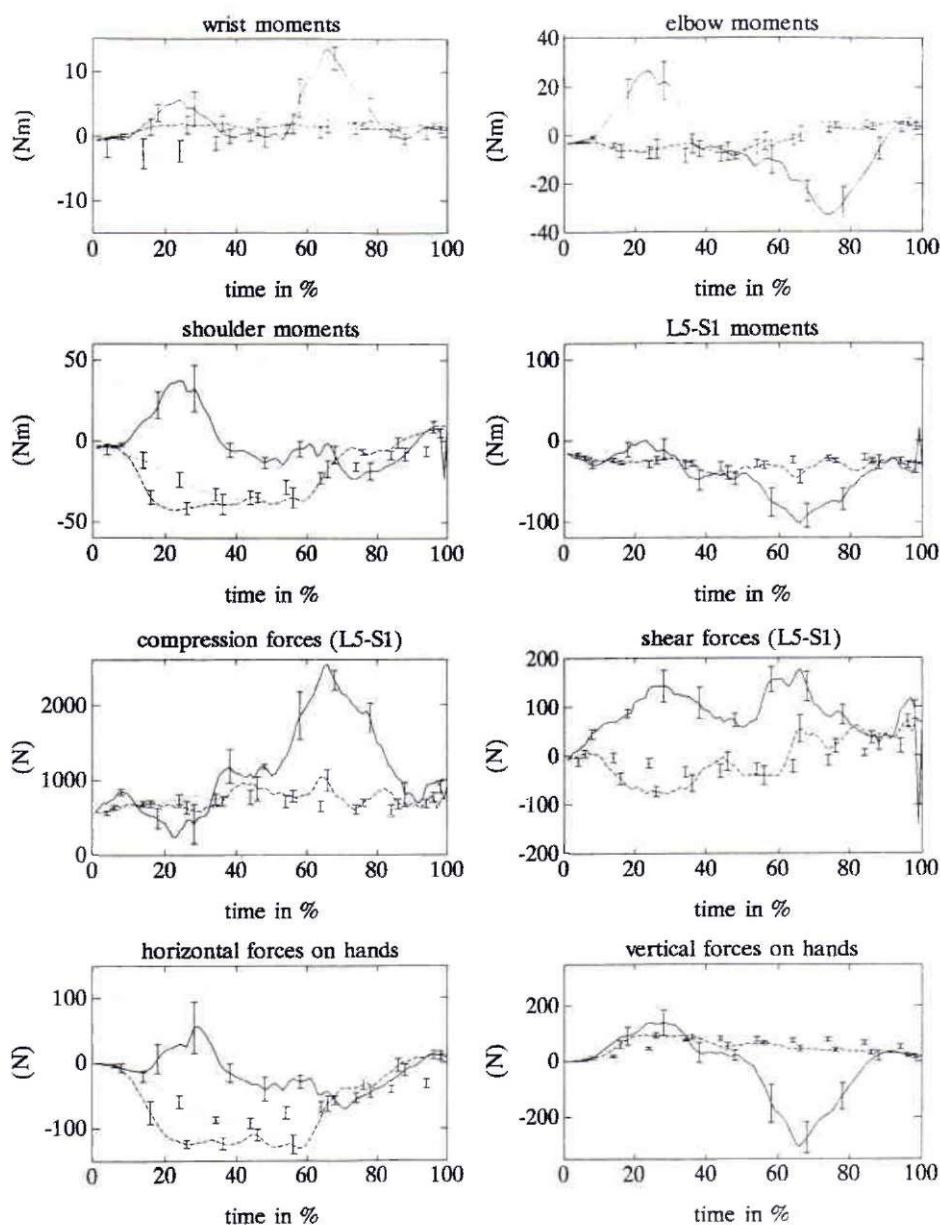
Effects of push handle height on mean and maximum net moments around wrists, elbows, shoulders and on L5-S1 compression and shear forces and the external hand forces were evaluated for the three tasks separately with an analysis of variance for repeated measures ( $P<0.05$ ). When push handle height appeared significant, a Tukey post-hoc test was used to determine which handle heights differed significantly from each other.

The difference between data obtained from trials using the living person loaded wheelchair, and from the trials using the dummy loaded wheelchair (78%) was tested with a paired T-test ( $P<0.05$ ). To study association between parameters a Pearson's product moment correlation was used ( $P<0.05$ ).

### Results

All 8 female subjects were able to perform the three tasks within the required experimental specification. Subjects did experience difficulties in performing the lifting task in a sagittal plane, especially when using the lowest push handle height. Although all trials were carried out, results for one subject performing wheelchair lifting at the lowest pushing height, could not be produced, due to loss of data.

Relative time histories are presented in Figure 3 to show mean curves over time ( $n=8$ ) of the net joint moments around wrists, elbows, shoulders and L5-S1, and of the compression and shear forces (L5-S1) and external horizontal and vertical forces on the hands for the duration of the tasks (100%) for the lowest pushing height only. Since a two dimensional biomechanical model was used, values of net joint moments in wrists, elbows and shoulders, present the sum of moments around both left and right joints.



Time histories of flat pushing (...), slope pushing (---) and lifting (—) at the lowest pushing height (61% shoulder height). Mean curves of net moments in wrists, elbows, shoulders and L5-S1, compression and shear forces (L5-S1), external horizontal and vertical forces (on hands). Vertical bars indicate the standard error of the means (S.E.M.) ( $n=8$  for flat and slope pushing,  $n=7$  for lifting).

Fig. 3. Time course (mean and standard deviation) of the net moments and compression and shear forces for the three tasks at the lowest push handle height.

Flat and slope pushing at the lowest pushing height showed little variation of moments at wrists and elbows; values were low and slightly

positive or negative, meaning small flexion or extension moments, respectively. Moments around shoulders and L5-S1 were negative,

indicating an anteflexion moment around the shoulders and an extension moment around L5-S1. During flat and slope pushing, negative values of horizontal forces on hands and positive values of vertical forces on hands were found due to forward and downward pushing, indicating a non-horizontal total force applied to the handle.

The lifting task at the lowest push handle height can be distinguished in two phases; a tilting and lifting phase. During the initial tilting phase the wheelchair is tilted around the axis of the rear wheels and moments of wrists, elbows and shoulders were positive (indicating a dorsal flexion moment in wrists, an extension moment in elbows and a retroflexion moment in shoulders). Negative extension moments of L5-S1 are reduced in this phase compared to the following phase. During the lifting phase, positive wrist moments and negative moments of elbows, shoulders and L5-S1 were found. Highest peak values of compression and shear forces (L5-S1), and vertical external forces on the hands were found during the lifting phase.

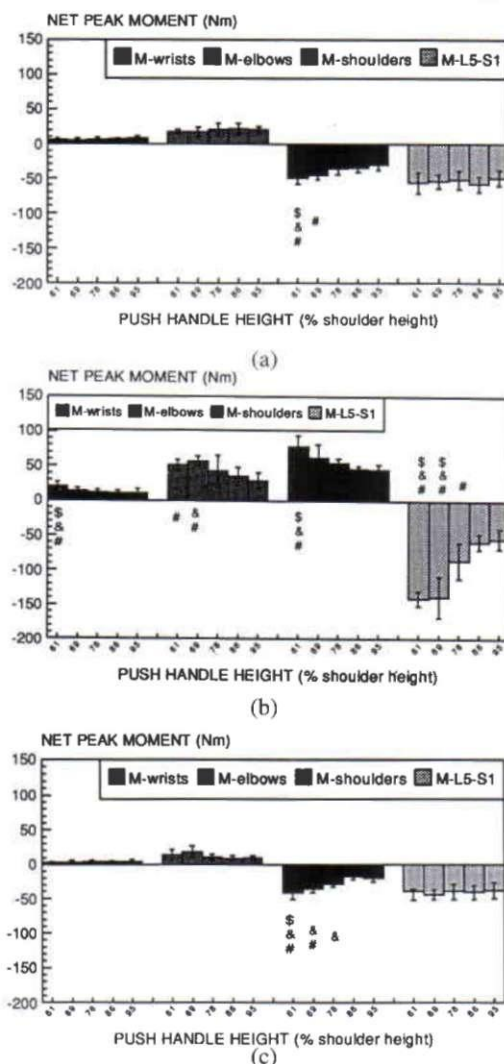
### Push handle height

Maximum values (mean and standard deviation) of net joint moments around the wrists, elbows, shoulders and L5-S1 are presented in Figure 4 for the different handle heights and the three tasks. Compression forces (L5-S1), shear forces (L5-S1) and external forces on the hands are presented in Table 1a to 1d.

#### Task 1: pushing the wheelchair on a flat surface

Only data of the first part of the trial were analyzed. During this part the wheelchair had not yet reached the slope. During flat pushing, peak moments were 4.5 Nm (at push handle height 78%) for the wrists, 18.1 Nm (69.5%) for the elbows, -41.9 Nm for the shoulders (61%) and -44.9 Nm for L5-S1 (69.5%) (Fig. 4a). The peak values for compression and shear forces (L5-S1) were 1051.6 N (69.5%) and 93.3 N (78%), respectively. Maximum horizontal and maximum vertical external forces on the hands were -114 N (78%) for  $F_x$  and 94 N (61%) for  $F_z$  (Table 1a).

Significant differences were found in relation to push handle height for the net moment around the shoulders, the mean moment around the elbows and the vertical forces on the hands.



N=8; MEAN & S.D.

\*, &, #, significantly different from 69.5, 78, 86.5 and 95%, respectively.

Fig. 4. Net peak moments around wrists, elbows, shoulders and L5-S1 (mean and standard deviation)

(a) flat pushing (Task 1)

(b) slope pushing (Task 2)

(c) wheelchair lifting (Task 3)

The shoulder moments and the external forces appeared to be lower at the higher pushing heights (Table 1a, Fig. 4a).

#### Task 2: pushing the wheelchair on an inclined surface

As was to be expected, during slope pushing the external forces, the peak net moments and

Table 1a. Flat pushing: maximal values of compression forces, shear forces and external forces in horizontal (F<sub>y</sub>) and vertical (F<sub>z</sub>) direction (N), depending on push handle height as a percentage of shoulder height (61 to 95%). Significant differences are indicated with \*, \$, & or #: significantly different from 69.5, 78, 86.5 or 95%, respectively.

Flat pushing (Task 1)	61%	69.5%	78%	86.5%	95%
CompF max					
MEAN	921.5	1051.6	988.8	982.1	971.5
SD	210.0	121.0	143.7	159.4	218.8
ShearF max					
MEAN	72.8	74.9	93.3	81.9	63.5
SD	22.7	10.0	40.5	25.5	24.0
F <sub>y</sub> max					
MEAN	-101.62	-110.1	-114.1	-94.5	-102.0
SD	14.9	5.1	18.0	16.4	17.9
F <sub>z</sub> max					
MEAN	93.7	62.2	31.3	25.0	12.5
SD	14.2	17.7	12.1	4.2	7.8
Sig. Tukey	*,\$,&,#	,\$,&,#	#		

the forces on L5-S1 were systematically higher in comparison with flat pushing (Task 1). The peak values of net moments were 7.9 Nm for the wrists (95%) and again relatively low. 21.8 Nm at 86.5% for the elbows, -50.8Nm for the shoulders at 61% shoulder height and -59.0 NM for L5-S1 at the highest push handle height (Fig. 4b). Highest values for the compression and shear forces were 1284.0 N and 130.1 N, respectively. Maximum F<sub>y</sub> values and F<sub>z</sub> values were -180.9 N and 109.1 N, respectively (Table 1b).

Significant differences in relation to pushing height were found for the net moment around the shoulders, the mean moment around the elbows and L5-S1, the mean horizontal force and maximum vertical forces on the hands. Net moments around shoulders and mean L5-S1 and the vertical forces appeared to be lower at higher pushing heights. Differences in mean net moment around the elbows were caused by a change in direction at the fourth pushing height (86.5%), but peak values showed no differences. Peak horizontal forces were higher

Table 1b. Slope pushing: maximal (or minimal) values of compression forces, shear forces and external forces (N), in horizontal (F<sub>y</sub>) and vertical (F<sub>z</sub>) direction, depending on pushing height as a percentage of shoulder height (61 to 95%). Significant differences are indicated with \*, \$, & or #: significantly different from 69.5, 78, 86.5 or 95%, respectively.

Slope pushing (Task 2)	61%	69.5%	78%	86.5%	95%
CompF max					
MEAN	1222.8	1196.9	1184.5	1284.0	1106.6
SD	260.2	179.9	221.4	195.6	175.7
ShearF max					
MEAN	130.1	115.8	126.6	116.6	123.7
SD	28.0	29.4	39.0	27.2	40.2
F <sub>y</sub> max					
MEAN	-161.7	-159.9	-167.0	-180.0	-180.9
SD	8.7	10.6	7.6	13.5	15.7
Sig. Tukey	&,#	&,#			
F <sub>z</sub> max					
MEAN	109.1	66.9	39.6	33.9	36.6
SD	23.7	11.2	19.7	17.7	11.7
Sig. Tukey	*,\$,&,#	,\$,&,#			



Table 1c Wheelchair lifting: maximal values of compression forces and shear forces (N) and external forces (N) – direction depending on pushing height as a percentage of shoulder height. Significant differences are indicated with \*, \$, & or #: significantly different from 69.5, 78, 86.5 or 95%, respectively.

Wheelchair lifting (Task 3)	61%	69.5%	78%	86.5%	95%
CompF max					
MEAN	2946.7	2839.8	1844.4	1397.0	1344.6
SD	206.3	443.3	493.0	261.3	308.0
Sig. Tukey	*, &, #	*, &, #			
ShearF max					
MEAN	266.0	274.0	200.7	170.7	138.8
SD	57.9	95.6	27.2	21.5	27.4
Sig. Tukey	&, #	&, #			

Table 1d. Wheelchair lifting: maximal values of external forces Fz and Fy (N) and of forces in horizontal pushing (Fy push) and pulling (Fy pull) direction, and forces in vertical (Fz) direction depending on pushing height as a percentage of shoulder height. Significant differences are indicated with \*, \$, & or #: significantly different from 69.5, 78, 86.5 or 95%, respectively.

Wheelchair lifting (Task 3)	61%	69.5	78%	86.5%	95%
Fy max push					
MEAN	-105.7	-105.7	-170.9	-153.3	-126.0
SD	32.0	23.8	33.8	24.9	20.1
Sig. Tukey	&, #	&, #	#		
Fy max pull					
MEAN	120.8	130.8	143.6	123.8	108.7
SD	47.8	66.4	17.9	24.9	28.4
Fz max					
MEAN	-425.4	-344.0	-203.9	-200.1	-235.4
SD	33.3	45.5	66.5	60.0	61.2
Sig. Tukey	*, &, #	*, &, #			

at the higher pushing heights (Table 1b).

### Task 3: lifting the wheelchair on to a curb

During the lifting task, peak net moments were 20.2 Nm around the wrists (at push height 61%), 57.1 Nm around the elbows (69.5%), 78.2 Nm around the shoulders (61%) and -140.9 Nm around L5-S1 (61%) (Fig. 4c). Peak compression and shear forces were 2946.7 N and 274.0 N respectively (61%) (Table 1c). Mean maximum horizontal push forces were -170.9 and mean maximum horizontal pulling forces were 143.6 N during this task (78%), while peak Fz was -425.4 N (61%) (Table 1d).

Peak net moments around wrists, elbows, shoulders and L5-S1, compression and shear forces (L5-S1) and vertical lifting forces were significantly lower for the higher push handle heights (Tables 1c and 1d; Fig. 4). Horizontal

push handle forces tended to be higher with the higher push handle heights. An overview of the significant trends in the data is shown in Table 2 for the different tasks.

No statistical differences were seen between

Table 2. Overview of significant decrease (↓) or increase (↑) of absolute values of maximum net joint moments, L5-S1 compression and shear forces and external forces in relation to increasing pushing heights.

	flat pushing	slope pushing	lifting
wrist			↓
elbow			↓
shoulder	↓	↓	↓
L5-S1			↓
compF			↓
shearF			↓
Fy		↑	↑
Fz	↓	↓	↓

the ISO-dummy test condition and the testing condition with the subject in the wheelchair.

## Discussion

### *Validity of experimental procedures*

Wheelchair handling obviously is a three dimensional (3-D) activity, however many problems in 3-D modelling still remain to be solved. Therefore 2-D modelling was used in this study, with tasks largely restricted to the sagittal plane. Obviously, axial rotation and lateroflexion of the trunk, both recognized as hazardous task components with respect to low back pain, cannot be studied.

Subjects in this study were young females without experience in wheelchair pushing. Clearly subjects had to be equally untrained on all push handle heights and tasks. Also the current results must be treated with caution when applied to older and probably more experienced women.

For reasons of standardisation an ISO-dummy was used in this study to prevent the possibly unpredictable role of a subject. Since no statistical differences were found between the ISO-dummy condition and the tests with the subject, it may be concluded that the dummy seems a valid replacement of a living person of the same weight under the given testing conditions. However, some additional remarks can be made. The position of the centre of gravity (COG) of a wheelchair containing a dummy or a living person is of great influence on the rolling resistance (Lemaire *et al.*, 1991) and therefore on the required effective push forces. There is also an important effect on the required pulling and lifting forces (Wawrzinek, 1981; Wawrzinek and Boenick, 1987). A COG positioned rearward with respect to the larger rear wheels generally causes a lower rolling resistance and thus lower push forces. During a tilting action (Task 3) this will cause lower pulling forces (due to a lower rolling resistance) but during lifting it causes higher lifting forces, due to a stronger torque effect of gravity with respect to the handles. Due to the extra weight of about 8 kg of the experimental frame and force transducers at the rear of the test wheelchair, the COG moved backwards in comparison with the standard wheelchair model.

Flat wheelchair pushing (Task 1) and slope pushing (Task 2) tasks as investigated in this

study were of a short duration. Therefore results of this study cannot be applied to comparable tasks of long duration. Whether wheelchair pushing tasks in all day practice are primarily of short or long duration is unknown. The duration of task performance was standardized and therefore tasks were performed in a sufficiently relaxed way. Maximum net forces and moments will indeed be influenced by (strong) variations in acceleration and deceleration and thus by the variation in performance time that can be expected in all day practice. It is obvious that the frequency of task components during wheelchair attending is greatly dependent on environmental aspects. It is unknown in what frequency the studied tasks do occur in different environments under daily life conditions. Furthermore, it is possible that different task components occur at the same time, such as flat or slope pushing while negotiating a side slope. These combinations might considerably increase biomechanical loading.

### *Tasks 1, 2 and 3*

Comparison of the three tasks shows differences in course and magnitude of net forces and net moments (Fig. 3). In general, slope pushing causes higher net forces and net moments than flat pushing, although the time course of the data is quite comparable. It is striking that net moments around wrists and elbows can be kept close to zero. Apparently subjects were able to choose a favourable position of wrists and elbows with respect to the resultant external forces at any pushing height. The orientation of the hand grips may have played an important role in this respect.

During slope pushing, higher handle heights lead to higher pushing forces, and at the same time caused lower net moments around the shoulders due to a more favourable direction of resultant external forces with respect to the shoulder joint centre of rotation.

Concerning net moments around L5-S1 one should realise that during moderate pushing the resultant external force causes an extending moment, which attributes to the extending moment exerted by the trunk extensor muscles. Therefore, pushing may have a decreasing effect on net L5-S1 moments. When pushing forces are high abdominal muscles, being trunk flexors, have to become active in order to compensate for the high extending moments



caused by external forces.

During wheelchair lifting a different time course, and much higher net forces and moments occurred than during flat and slope pushing. In fact three task components can be distinguished. Firstly when the wheelchair is pulled backwards, immediately followed by a downward pushing of the handles, thus tilting the wheelchair around the rear wheel axis. Then the conjunct forward pushing and lifting action started. This movement caused highest net forces and net moments. During performance of the lifting task the function of the curb is essential: when the rear wheels of the wheelchair are pushed against the curb, a lifting moment is created on the wheel axis by a combination of reaction forces of the curb and the external pushing forces (Wawrzinek, 1981). This lifting moment contributes to the upward movement of the wheelchair. The forward pushing forces cannot be exerted when the push handles are positioned too low (61%), as was derived from the video tapes. Due to the large distance between the hands and shoulders, the shoulder muscles seem unable to exert the required moments.

It is difficult to evaluate the possible consequence of the biomechanical loading during the investigated tasks. No standards have been established to relate net moments around the wrists, elbows and shoulders. With respect to compression forces of L5-S1 many studies have been conducted and standards are established. This will be discussed below.

### **Push handles**

Most attendant push wheelchairs in the Netherlands are equipped with backwards pointing hand grips, approximately 0.90 m high (GMD, 1992). This position is only favourable when used in combination with low push handle heights. On the contrary, when pushing handles are high this handle position is unfavourable with respect to the orientation of the wrists. During pushing, pulling and lifting the wrists

are forced into an extreme ulnar deviation when using the conventional grip handles. The test wheelchair had a horizontal pushing bar and hand grips allowing a symmetric and consistent pushing and pulling technique with a comparable orientation of the hands, wrists and lower arm at different pushing heights. The push bar did cause some trouble in lifting the wheelchair using lower handle heights. The wrists were then forced into an extreme dorsiflexed position. The net moments at the wrists during wheelchair lifting using the two lower pushing heights can be expected to be higher during this experiment than in all day practice using conventional grip handles. On the other hand, net moments around the wrists during pushing and pulling using higher push handle positions can be expected to be lower than during ordinary practice, when using backwards pointing handles.

### **Push handle height**

The average absolute push handle heights for the eight female subjects used during the experiments ranged from 0.84m ( $\pm 0.023$  at 61%) to 1.308 m ( $\pm 0.037$  at 95%). Since the average push handle height of an ordinary wheelchair is  $0.924 \pm 0.014$  m (GMD, 1992), this appeared comparable to the second push handle height (69.5%:  $0.957 \text{ m} \pm 0.026$ ) of shoulder height) in the current experiment. As is shown in Figure 4 and Tables 1 and 2, most significant differences in net moments and forces appeared to exist between the two lower and the three higher push handle heights. It can be concluded that the three higher push handle heights cause lower net moments than the two lower pushing heights with respect to the investigated tasks. Therefore, the common push handle height of about 0.920 m, being in the lower range of push handle heights of this study, might not be a too favourable pushing height from a biomechanical perspective and should be reconsidered. With respect to variable push handle heights, the pushing height that

Table 3. Compressive strength and damage load for women of 45 to 75 years according to Genaidy (1993) and NIOSH (1981)

	age (yr)	45	55	65	75
Genaidy	body weight (kg)	65	65	65	65
	compressive strength (N)	5958	5221	4484	3747
	damage load (N)	3637	3087	2538	1988
	NIOSH	compressive strength (N)	4150	3320	2075

equals elbow height, as advised by the GMD, matches 78% of shoulder height (Molenbroek and Dirken, 1986) being the third push handle height in the present study and lies just within the favourable pushing height range.

With respect to fixed push handle heights, the fourth push handle height ( $1.191 \text{ m} \pm 0.034$ ) appears to be acceptable for the 5th to 95th percentile Dutch women (Table 3), because this height remains within the favourable pushing range of 78% to 95% of shoulder height. Whether these higher pushing heights are also favourable with respect to other task components than the ones studied here, remains to be investigated.

Studies of Abel and Frank (1991) have shown that the preferred position for wheelchair handles probably lies in the region of  $75 \pm 3.8\%$  of shoulder height. This could be in accordance with results of this study, since middle and higher pushing heights (78 to 95%) cause less net moments around shoulders and L5-S1 than lower heights. Abel and Frank (1991) also stated that no difference was found between these moments at both high and low handle heights. This is not in accordance with the present study. Since Abel and Frank (1991) have not published their methods and procedures, it is not clear why different results have been found in their study, particularly because it is unknown which pushing heights have been investigated.

### Push forces

Recently measurements of the forces necessary to push wheelchairs were made (GMD-TNO 1991). This was done on low piled carpet. The highest force was measured during pushing the wheelchair, loaded with a 75 kg dummy (ISO/DIS 1776-11), quietly from standstill to walking speed (GMD-TNO, 1991). The average horizontal push force for eight different wheelchairs comparable to the wheelchair in the present study (GMD, 1992), was  $22.6 \pm 4.4 \text{ N}$ . The average push handle height of these wheelchairs is  $0.924 \text{ m} \pm 0.014$  (GMD, 1992), which is comparable to the second push handle height in the current study. However, in the present study, push forces in flat pushing using the second push handle height were as high as  $110.1 \pm 5.1 \text{ N}$ . Four possible explanations can be offered to interpret this difference:

- the initial position of the castor wheels. In this study, during flat pushing the front wheels were placed back to front, in order to create a realistic situation. GMD-TNO does not mention the position of the front wheels.
- the weight of the wheelchair, which in the current study was about 8 kg extra, due to the force transducer and its frame. GMD-TNO only mentions the weight of the wheelchair without equipment.
- the possibility of different accelerations of the wheelchairs which are unknown for the GMD-TNO study.
- differences in floor surfaces and tire pressures may attribute to the difference between push forces found by TNO and this study.

Glaser *et al.* (1980) have established horizontal external push forces of loaded wheelchairs at constant velocity on the level and on  $1^\circ$  to  $5^\circ$  inclined surfaces depending on total weight of the loaded wheelchair. On a flat tiled surface the authors measured a pushing force of about 11 N for a loaded wheelchair of 105 kg total weight. On a  $5^\circ$  inclined surface this wheelchair would require 108 N pushing force. If compared to pushing forces at the end of flat pushing and of inclined pushing tasks, forces measured by Glaser appear to be lower than pushing forces found in this study.

Glaser *et al.* (1980) came to the conclusion that handle height has no influence on the horizontal push force. This is in accordance with the results of the present study (Table 1). It should be mentioned though, that according to the results of this study, push handle height seems to have great influence on external forces in the vertical direction (Fig. 4, Table 1).

### Compressive strength

Table 3 shows compressive strength and damage load for elderly women (body weight 65 kg) according to Genaidy *et al.* (1993) and National Institute of Occupational Safety and Health (NIOSH, 1981). Genaidy *et al.* (1993) provide regression equations for compressive strength (compressive force at which tissue failure occurs) and damage load (force which causes first signs of damage) depending on sex and age. Obviously, increasing age results in considerably lower damage loads. NIOSH (1981) presents somewhat lower values for compressive strength. If compared to maximum compression forces during wheelchair lifting at

61 and 69.5% of shoulder height (2946.7 and 2839.8 N) these values appear to exceed the NIOSH compressive strength force and the damage loads according to Genaidy *et al.* (1993) for women of 65 years and older. Moreover, the NIOSH action limit sustains a maximum lifting force of 392 N under most favourable conditions for industrial working people. This is comparable to wheelchair lifting at the lowest pushing height. Average vertical peak lifting forces appear to be as high as 425.4 N. Therefore it can be concluded that lifting a 75 kg loaded wheelchair onto a curb equals or even exceeds the NIOSH action limit and should be avoided. In conclusion, women, especially those over 65 years are at risk when pushing-lifting wheelchairs on to a curb or steep slopes, when using low push handle heights and when the total weight of the wheelchair plus occupant exceeds 105 kg.

Finally, Snook and Ciriello (1991) have established maximum acceptable horizontal push forces for female industrial workers in relation to two push handle heights and two frequencies (every 2 and every 5 minutes) using psycho-physical methods. It is difficult to make a comparison between these pushing tasks and (outdoor) wheelchair pushing, but it does not seem to be unrealistic however, to presume the occurrence of an initial push every 2 or 5 minutes. The maximum acceptable forces as established by Snook and Ciriello were 180 and 200N for the respective frequencies and irrespective of push handle height. It can be concluded that during wheelchair pushing, no maximum acceptable limits are exceeded for industrial working females.

### Conclusions

Biomechanical loading when pushing a wheelchair is partly influenced by push handle height. In general higher push handles appear to offer some advantages with respect to the investigated tasks.

Pushing wheelchairs on a flat surface leads to higher net moments around shoulders and to higher vertical external forces on hands with a lower push handle height.

Pushing wheelchairs on an inclined surface leads to the highest maximal horizontal pushing forces when using high pushing handles. Nevertheless, higher net moments around shoulders and L5-S1 and vertical external forces

are seen in relation to lower push handle heights.

Low push handles used in moving wheelchairs on to a curb, cause higher net moments, compression forces and shear forces on L5-S1, and higher lifting forces in comparison to higher pushing handles. A high push handle allows the attendant to push the wheelchair upon the pavement rather than to first have to lift the wheelchair. This push technique leads to lower net moments around all joints involved, than seen in lifting.

86.5% of shoulder height in particular appears to offer some advantages with respect to the investigated tasks and parameters. For an average woman this percentage corresponds to a pushing height of about 1.182 m, which is within a favourable range of pushing heights for the 5th to 95th percentile of Dutch women. When pushed by taller or smaller people, an adjustable pushing height might be necessary.

Other adaptations to the design and orientation of push handles and the construction of the wheelchair should be reconsidered in future research. For instance backwards pointing pushing handles do not seem very appropriate at higher pushing heights. Especially elderly female wheelchair attendants may benefit from more appropriately designed wheelchairs, since they appear at risk of L5-S1 damage, when lifting wheelchairs onto a kerb.

Many other aspects of wheelchair pushing remain unclear. Specially biomechanical loading during asymmetric tasks such as turning and pushing on side slopes, and the physiological strain due to static muscular work need further investigation.

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## Book Review

### **The grasping hand.**

**L. MacKenzie and T. Iberall.**

**North-Holland, 1994.**

**ISBN 0-444-81746-8.**

**500pp. NLG 250, USD 142.75**

In the introduction of *The Grasping Hand* the authors state, "By critically evaluating the current knowledge about the hand, our goal is to make explicit a common language for understanding human prehension across robotics, cognitive science, kinesiology, computer science, psychology, neuroscience, medicine, and rehabilitation". This statement captures much of the essence of the book. The book is, primarily, a critical review related to the hand, to the arm, to artificial grasping, and to models of the hand/arm/brain system. As such, the name of the book does not really encompass its contents, although the contents are, in one way or another, related to the hand, to prehension, or to the provision for prehension. The reader will find the book contains major sections about arm movements; about experimental data, theories and models of motor control; about neural network modeling and about robotic principles; in fact, all the fields listed in the first sentence above – and more – are included in the book. It is doubtful if a common language is achieved in the book. The topic of the hand is too complex for that to happen as the result of one publication. Gait analysis investigators have been trying to achieve a common language between the clinical and engineering fields for many years – with only partial success – and it is unlikely that similar issues will be sorted out more easily with the hand/arm system.

*The Grasping Hand* has three main parts. In Part(1) the concept of "virtual fingers" is introduced. This concept, partially Iberall's own work, appears to be a valuable mental aid for thinking about prehension and may be a useful way to model prehension. For example, in lateral prehension the thumb is one virtual finger and the four other fingers constitute the

second virtual finger. The concept of pad opposition and palm opposition may also be useful for prosthesis designers. Part (1) also contains an interesting section on taxonomies of grasping patterns. Patterns of grasp have been classified in many ways through the years and Part (1), along with Appendix "B" (Taxonomies of Prehension), comprise a comprehensive review of classification schemes. One of the ideas presented in the book is the need for investigators to explore going "beyond classification toward quantification". Such exploration is applauded and should be encouraged. Nevertheless, little new material of a quantitative nature on the hand is presented in the book. The experimental results and the conceptual models described seem considerably more useful than the computational models presented (mostly neural networks). In the prosthetics and orthotics field, description and classification are important; however, as this book points out, description and classification should not be thought of as constituting understanding.

Part (2) consists of what the authors call "phases" of prehension. It deals with prehension task planning, movement before contact with the object to be grasped, movement during contact with the object, and general concepts of opposition space. The review of research work dealing with hand opening (aperture) as a function of arm movements and of specific arm/hand tasks may have importance in prosthetic arm design. The fact that the thumb appears to move less than the index finger in hand opening, while the hand is moving toward an object, suggests that humans may use the thumb as a directional guide to the object. This was a tenet of prosthetic prehensor design until advent of the electric powered prosthetic hands, most of which move the thumb and the fingers together to achieve appropriate grasp opening. Part (2) of the book is also concerned with mathematically-based models of human movements and with ideas from the robotics field. The most interesting aspect of Part (2), in



the reviewer's opinion, is the section of Chapter 6 concerned with the skin as an organ of grasp.

Part (3) considers constraints on human prehension, and future directions. Unfortunately, Part (3) adds little to the general themes already presented. Extensive appendices complete the book. Appendix "A" (Human Upper Limb Anatomy) and Appendix "C" (Computational Neural Modeling) seem extraneous because readers would probably do better by examining textbooks on these subjects. Appendix "B" on Taxonomies of Prehension is worthwhile, as already mentioned. Appendix "D" covers prosthetic and robotic hands. Although it contains a number of errors, it is a useful reference.

The book was written by two authors. One assumes that MacKenzie wrote the sections dealing with the more biological aspects of hands and arms, while Iberall wrote the technical sections, but this is not explicitly stated. The biological and the technical/mathematical aspects of reaching and grasping are often integrated within the same chapter. For example, one can read about eccrine glands in the same chapter with matrix manipulations. The authors state that one of their purposes in working on the book as a team was to promote an integrated view of prehension and to develop a common vocabulary. Although this may have happened between them, an integrated view is not so evident in their book. In many ways the topics, as presented, do not appear miscible.

MacKenzie and Iberall summarize the book as exploring two questions, the question of the nature of the human hand and what might be involved in the central nervous system (CNS) as it controls the hand. As was the case with the

authors' desire for common language and integration of viewpoint, the book is only partially successful at providing understanding of the nature of the human hand and of its interaction with the CNS. Perhaps the authors' goals are a bit too ambitious for such a complex system.

In summary, this book is a useful reference for researchers interested in human movement and motor control of the arm and hand; however, the review format makes it less useful for clinicians and educators. To the reviewer, the biological topics of the book are more compelling than the robotic and neural network sections. Robotics has had some influence on the prosthetics and orthotics field, but not much; probably because the design constraints associated with prostheses and orthoses are exceedingly different from those of robots.

A few words should be said about physical aspects of the book. While it is laudable that North-Holland publishes books of this nature, one could wish for higher paper and printing quality and for better formatting. Figures, tables, and graphs have a slightly amateurish appearance and that detracts from the book. Lines of drawings are often too thick and captions seem too dark. There is considerable "see through" from one page to the next. The use of extensive underlining instead of italics seems undesirable. Underlines often merge with the letters and appearance is degraded.

Dudley Childress  
Northwestern University Rehabilitation  
Engineering Research Laboratory  
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USA

## **Calendar of Events**

### **National Centre for Training and Education in Prosthetics and Orthotics Short Term Courses 1995-96**

#### **Courses for Physicians, Surgeons and Therapists**

- NC505 Lower Limb Prosthetics; 22-26 January, 1996  
NC514 Orthotic Management of the Diabetic Foot, 5-6 February, 1996  
NC510 Wheelchairs and Seating; 25-27 March, 1996  
NC513 Prosthetic Knee Mechanisms and Foot Design, 15-17 April, 1996  
NC511 Clinical Gait Analysis; 1-3 May, 1996  
NC506 Fracture Bracing; 7-10 May, 1996

#### **Course for Orthotists and Therapists**

- NC217 Ankle-Foot Orthoses for the Management of the Cerebral Palsied Child;  
24-26 April, 1996

Further information may be obtained by contacting Professor J. Hughes, National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, Curran Building, 131 St. James' Road, Glasgow G4 0LS, Scotland. Telephone: (+44) 141 552 4400 ext. 3298, Fax: (+44) 141 552 1283, E-mail [Annette.Hepburn@strath.ac.uk](mailto:Annette.Hepburn@strath.ac.uk)

#### **15-17 January, 1996**

2nd Congress of the International Council for Global Health Progress, "Cancer, Aids and Quality of Life", Paris, France.

Information: Airmec, 2 Boulevard Du Montparnasse, 75015 Paris, France.

#### **22-26 January, 1996**

7th Instructional Course of the International Society for the Study of the Lumbar Spine, Christchurch, New Zealand.

Information: Spine in Action, The Conference Office, University of Canterbury, Private Bag 4800, Christchurch, New Zealand.

#### **22-24 February, 1996**

Annual Scientific Meeting of ISPO UK National Member Society, Harrogate, England.

Information: Mr. D. Simpson, ISPO '96, National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, Curran Building, 131 St. James' Rd., Glasgow G4 0LS, Scotland.

#### **22-27 February, 1996**

Annual Convention of the American Academy of Orthopaedic Surgeons, Atlanta, USA.

Information: AAOS, 6630 North River Road, Rosemont, IL 60018-4226, USA.

#### **2-3 March, 1996**

Annual Scientific Meeting of British Association of Prosthetists and Orthotists, Glasgow, Scotland

Information: Mrs. H. Thomas, BAPO, Dunoon and District General Hospital, Dunoon, Argyll PA23 7RL, Scotland.

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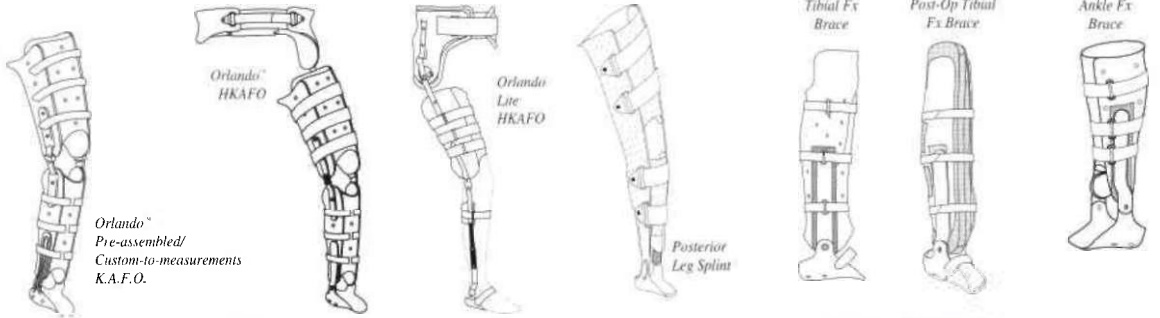
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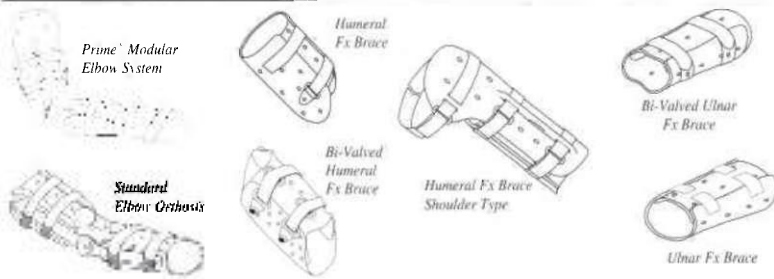
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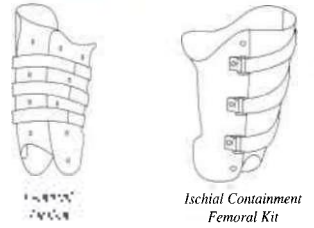
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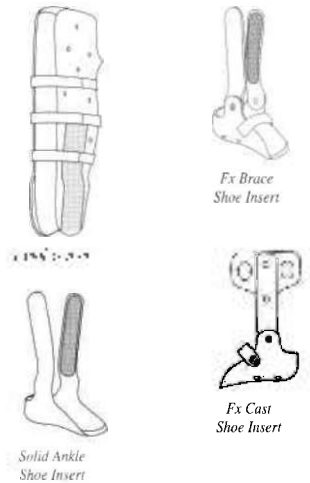
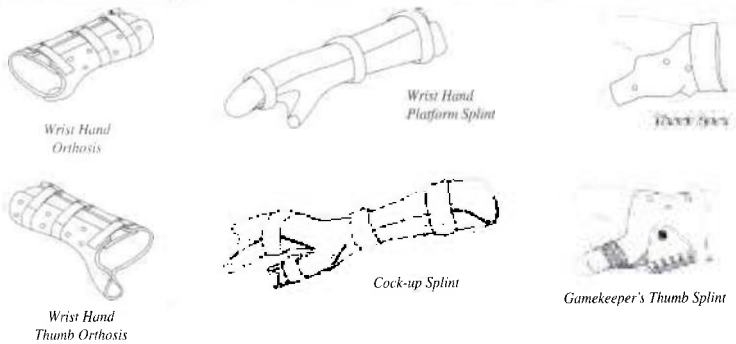
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## COMPONENTS



## HAND ORTHOSES



## FOOT ORTHOSES

