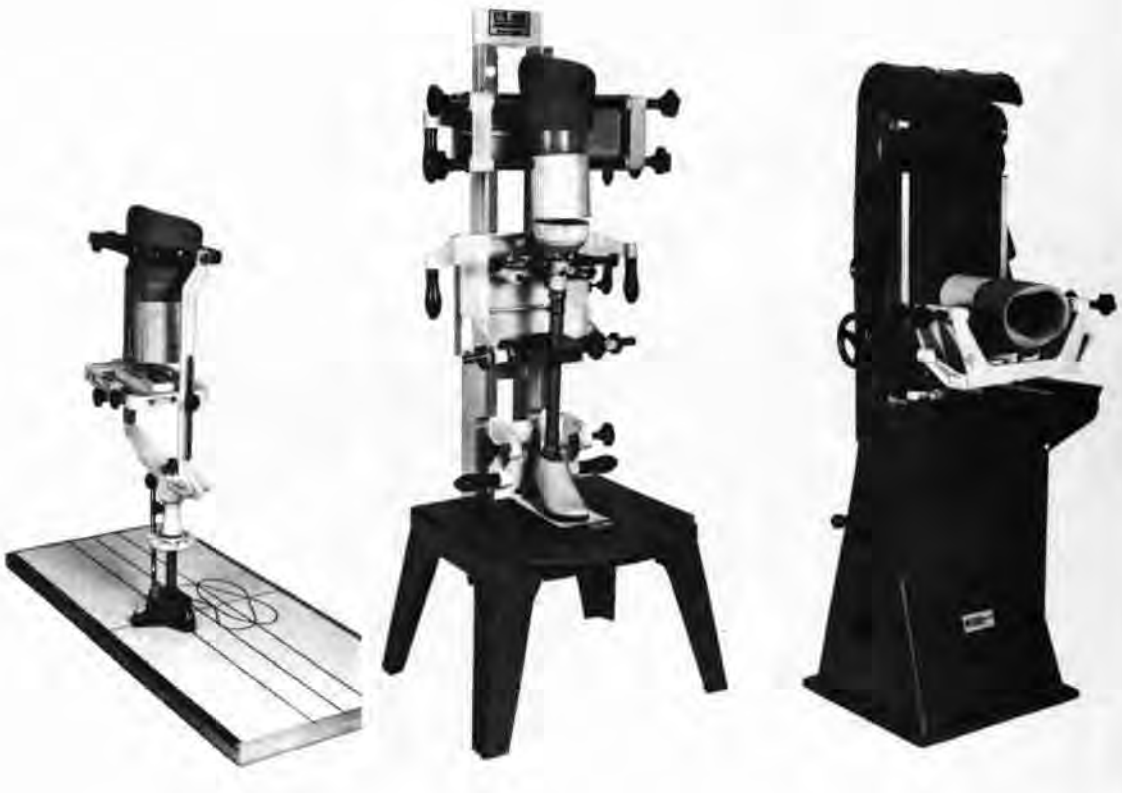
The Journal of the International Society for Prosthetics and Orthotics

Prosthetics and Orthotics International

December 1979, Vol. 3, No. 3



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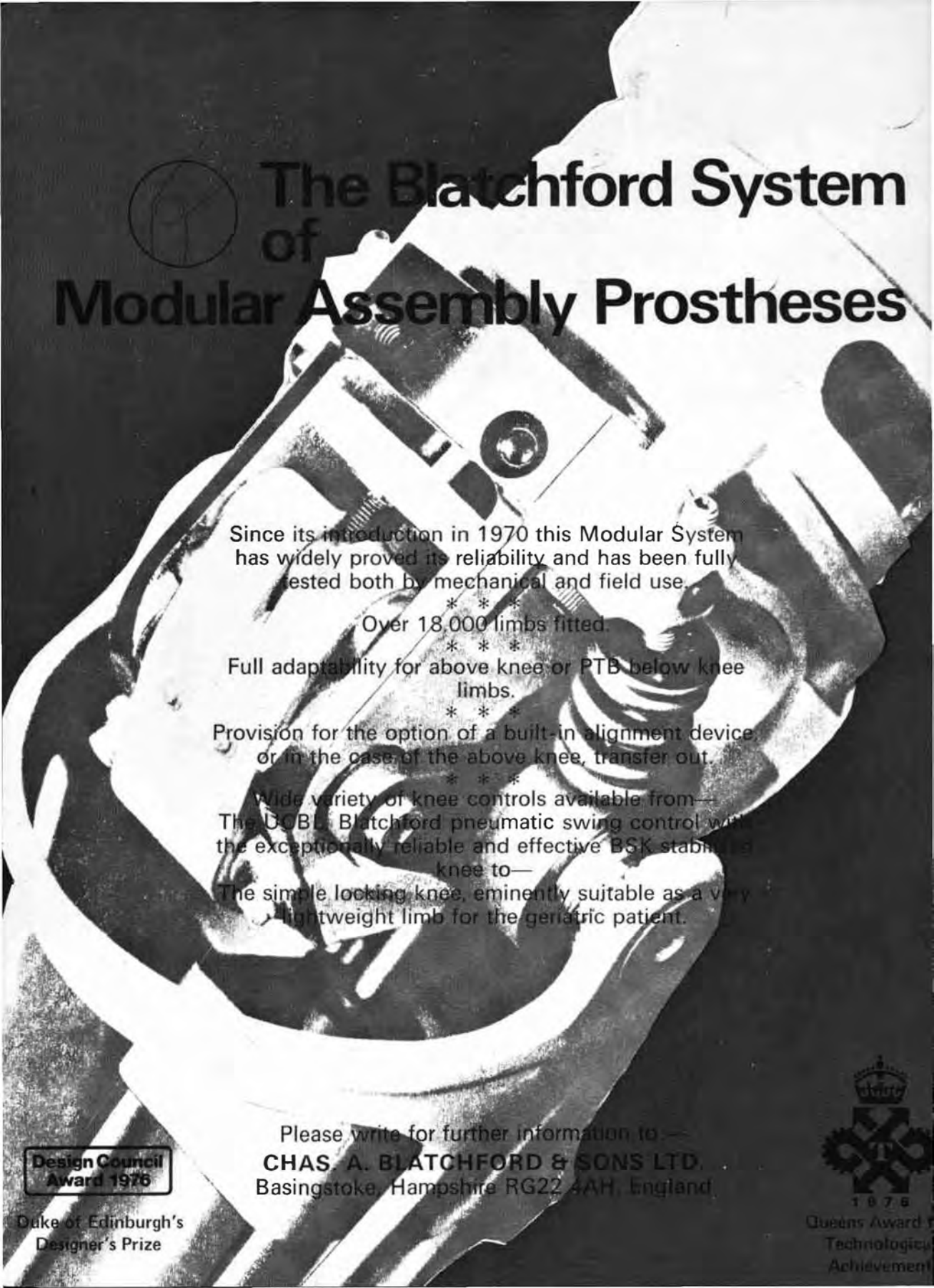
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The Journal of the International Society for Prosthetics and Orthotics

December 1979, Vol. 3, No. 3

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Prosthetics and Orthotics International, 1979, 3, 119–120

Editorial

THE Society during the past year has been involved in various activities but foremost in our thoughts has been the preparation for our Third World Assembly and Congress in Bologna, Italy, beginning 28th September, 1980. I hope that by the time the second announcement is in your hands you will have committed yourself to joining us in Bologna.

The first meeting of Technical Committee 168 of the International Standards Organization, which we as an international organization were in part responsible for establishing, was held in Pforzheim on 29th-30th May, 1979. A programme of work has been developed based on three working groups with ISPO members heavily represented in all three. While there is a great deal of work to be done now the membership should be proud of its efforts over the years in working towards the establishment of this new Technical Committee.

We still aspire to gaining funds which will put our Society on a firmer financial footing. From the beginning the Society has been sustained in financial terms by your subscriptions and, as a dynamic organization, by your work. There are now several National Member Societies which are very active in promoting conferences, seminars and instructional courses. These activities have, in some cases, been responsible for remarkable increases in our membership. Perhaps more important has been the influence on professionals outside our Society. Their eyes have been opened to a multitude of problems which, for the most part, governments and other agencies have been unable to solve. Indeed some governments are apparently unaware of these problems which may affect many thousands of their citizens. Our members in many countries have through their activities, begun to influence these governments and persuade them to face up to their responsibilities to the disabled. In some instances governments have been persuaded to legislate more favourably and in others to allocate funds for the training of professionals and for the provision of facilities. These activities have also brought increasing respect and status to our field of activity. The more traditional specialities, for example, internal medicine and surgery, now appreciate that in prosthetics and orthotics and related fields the body of knowledge is not just that embodied in manual crafts but has an academic base of considerable substance which can be intellectually very demanding.

To produce an impact such as this in the commercial world would demand the recruitment of many high powered salesmen and technical experts and would require that each be paid very high salaries to retain their services. Many of our members are poorly paid professionally but continue in their efforts in pursuit of the aims of ISPO out of dedication to their purpose and their enthusiasm. They remain unpaid for these services. I am sure that the majority of our activists are unknown to me personally but I can vouch for the spirited endeavours of their representatives who appear on the International Committee or who fill various consultative roles in the Executive Board or are members of the Board itself. Here again our representatives on the Standing Committees and on the Board have to spend many hours in their homes and their offices in correspondence with colleagues and others relating to our various objectives. Many spend weeks of travelling every year and have almost become inured to the soul-deadening environment of airports around the world, the disturbance to their biological rhythms, the disturbance to their families and to the ever increasing amount of work piling up on their own desks while they are away from home. They retain their enthusiasm despite the frustrations involved in international affairs. They remain dedicated to their purpose and they remain unpaid. Indeed many have had to dig into their own pockets to support the travelling essential to the Society's objectives.

In Copenhagen a small group of people whom I have styled as the Danish Resource Group, namely Knud Jansen, Jorgen Kjolbye, Erik Lyquist, Wilfred Kragstrup and Aase Larsson keep the wheels turning at the Secretariat. They continue to meet most Mondays to help and advise Aase Larsson in her task of keeping the Society together through her sustained and uniquely articulate correspondence. Aase ensures that every ISPO member contacting the Secretariat will get the appropriate service. She maintains a superb filing system not only in the filing boxes but also in her head and few of the nuances of

the dynamics of the Society escape her notice. She has a well organized appreciation of all the members in office in the Society and has devised her own elegant way of ensuring the necessary performance from each of them. The time that Aase Larsson spends working for the Society extends far beyond office hours and in another environment would command a far greater salary than she presently receives. Indeed, Aase Larsson is the only person in ISPO who is paid by the Society, but it is a tiny sum. Aase's devotion to her task and to our Society is such that a stranger might have thought she was married to the Society. I can now reveal that this suggestion has no foundation in fact although it is true that her relationship to the Society is a very intimate one. It is also true that she is married. In Rome, in October, she married Mr. Mugge Hansen a well-known Danish journalist. We congratulate this wise and fortunate Dane and we wish our Aase a long and happy married life.

George Murdoch

President.

Third World Congress

Bologna—28 September–4 October 1980—Prelude to 1981 Year of the Disabled

Bologna col suo centro storico quasi intatto, la bellezza delle sue piazze, il fascino dei suoi monumenti e delle sue raccolte d'arte, è una fra le città più interessanti d'Italia. Situata al centro dell'Emilia-Romagna, de cui è capoluogo, la città è famosa anche come centro culturale e ospita la più antica Università d'Europa, fondata nel secolo XI. La cordialità dei suoi abitanti e la fama universale della sua cucina fanno di Bologna un luogo di soggiorno piacevole e a dimensione umana.

Bologna with its beautifully preserved historical centre, its squares, its fascinating monuments and works of art, is one of the most interesting cities in Italy. Situated in the centre of the Emilia-Romagna Region, of which it is the capital, the city is also well-known as a cultural centre. It has the oldest University of Europe, founded in the XIth century. The sociability of its inhabitants and its gastronomic specialities have gained Bologna a well deserved reputation.

Bologna mit seinem nahezu vollkommen erhaltenen historischen Zentrum, der Schönheit seiner Plätze und der Anziehungskraft seiner Baudenkmäler und Kunstsammlungen ist eine der interessantesten Städt in Italien. Die Stadt liegt im Herzen der Emilia-Romagna deren Hauptstadt sie ist, und ist sowohl als Kulturzentrum berühmt, als auch als Sitz der ältesten Universität der Welt die im 12. Jahrhundert gegründet wurde. Die Herzlichkeit ihrer Einwohner und der Weltruf ihrer Küche machen den Aufenthalt in Bologna angenehm und geben ihm eine menschliche Dimension.

Bologna avec son centre historique presque intact, avec la beauté de ses places, le charme de ses monuments et de ses collections d'art, est une des plus intéressantes villes d'Italie. La ville, située au centre de l'Emilia-Romágna, dont elle est le chef-lieu, est également connue comme centre culturel possédant l'Université la plus ancienne d'Europe, fondée au XI siècle. L'amabilité de ses habitants ainsi que la renommée de sa cuisine font de Bologna un lieu de séjour agréable et à dimension humaine.

Bologna con su centro histórico casi intacto, la belleza de sus plazas, el atractivo de sus monumentos y de sus colecciones de arte, es una de las ciudades más interesantes de Italia. Situada en el centro del Emilia-Romagna, de la cual es capital, la ciudad es famosa aunque como centro cultural y es sede de la más antigua Universidad de Europa, fundada en el XI siglo. La cordialidad de sus habitantes y la fama universal de su cocina hacen de Bologna un lugar de agradable temporada y a dimensión humana.

La Società Internazionale per la Protetica e l'Ortotica (I.S.P.O.), in collaborazione con l'INTERBOR (Unione Internazionale dei tecnici ortopedici) ha scelto Bologna come sede del 3° Congresso mondiale dell'I.S.P.O. che si terrà dal 28 settembre al 4 ottobre 1980. Per il suo elevato livello scientifico che amministrativo, e per il vasto scambio di esperienze, il Congresso rappresenta un preludio ideale al 1981, anno dei Disabili proclamato dall'ONU.
Sono in programma: *corsi di istruzione* per offrire ai medici, ai chirurghi, ai tecnici ortopedici, ai terapisti e ai bioingegneri un complesso di informazioni di utilità immediata per l'ambiente clinico; *sedute scientifiche* per lo scambio di dati su studi e ricerche recenti; *simposi* di approfondimento su progressi avvenuti negli ultimi tre anni, in protetica e in ortotica.
Si tratta di un'occasione di eccezionale rilevanza per comunicare con colleghi del proprio campo, usufruendo di tutti i comforts offerti dal modernissimo Palazzo dei Congressi di Bologna, una città che è il centro dell'ortopedia e della protetica in Italia.
La traduzione simultanea in italiano, inglese, francese e tedesco darà la possibilità ai partecipanti di seguire i temi trattati in quattro aule diverse.

The International Society for Prosthetics and Orthotics (I.S.P.O.) in collaboration with INTERBOR (The International Association of Orthotists and Prosthetists) will organize the 3rd World Congress of I.S.P.O. in Bologna, Italy on September–28 October 4, 1980. The programme will provide substantial coverage of technical, scientific and administrative topics associated with amputation surgery, prosthetics, orthotics and related areas of orthopaedics and rehabilitation engineering, a fitting prelude to the Decade of the Disabled and to 1981, the United Nations Year of the Disabled Person. *Instructional courses* such as those presented at the 1977 (New York) congress will be offered so that physicians, surgeons, prosthetists, orthotists, therapists and rehabilitation engineers will receive information useful in a clinical setting.
The Congress will also provide exchanges of information on recent and current research and development, *symposia* to treat very special and sometimes controversial subjects in detail, and *plenary sessions* in which world leaders will present triennial reviews of progress in the major aspects of prosthetic and orthotic rehabilitation.
Simultaneous translation will be available in four different sites at the same time! Languages offered will include English, French, German and Italian. Translation into Spanish and Japanese is also being considered.

Die Internationale Gesellschaft für Prothetik und Orthetik (I.S.P.O.) veranstaltet in Zusammenarbeit mit der INTERBOR (Internationale Union der Orthopädietechniker) vom 28. September bis 4. Oktober 1980 in Bologna, Italien, den dritten Weltkongress. Das Programm sieht eine gründliche Behandlung technischer, wissenschaftlicher und verwaltungstechnischer Themen im Zusammenhang mit Amputationschirurgie, Prothesen, Orthesen und den verwandten Gebieten der Orthopädie und Rehabilitationstechnik vor. Wir glauben, dass das Programm auf Grund seiner kritischen Übersicht über neue Erkentnisse, seines wissenschaftlichen Inhalts und seiner umfassenden Berichterstattung über Erfahrungen, dem Geist des Behinderten gerecht wird und eine ideale Einleitung zum Jahr 1981, dem internationalen Jahr des Behinderten der Vereinten Nationen, darstellt.
Vorgesehen ist eine äusserst interessante und informative Reihe von *Forbildungskursen* so wie sie auf dem Kongress von 1977 (New York) geboten wurden. Diese Kurse bieten Ärzten, Chirurgen, Orthopädietechnikern, Therapisten und Rehabilitationsingenieuren Informationen, die in klinischem Rahmen nützlich sein können. *Wissenschaftliche Sitzungen* werden als Rundtischgespreche zum Informationsaustausch über jüngste und laufende Forschungs- und Entwicklungsarbeiten gegliedert sein; ausserdem werden *Arbeitsgruppen* und *Symposien* organisiert, so dass kleinere Gruppen besonders spezielle und manchmal umstrittene Themen im Detail behandeln können. In *Hauptsitzungen*, werden in der ganzen Welt anerkannte führende Persönlichkeiten eine Übersicht über die Fortschritte geben, die in den letzten drei Jahren hinsichtlich der wichtigsten Aspekte der prothetischen und orthetischen Rehabilitation gemacht wurden.
Diese Gelegenheit zum Gedankenaustausch mit Kollegen und Gleichgesinnten wird in ihren Möglichkeiten noch durch die ausgezeichneten Einrichtungen gesteigert, da ein attraktives, modernes Kongresszentrum zur Verfügung steht. Simultanübersetzungen werden gleichzeitig in vier verschiedenen Räumen angeboten und zwar in Englisch, Französisch, Deutsch und Italienisch.

La Société Internationale pour l'étude des Prothèses et Orthèses (I. S. P. O.), en collaboration avec INTERBOR (Union Internationale des techniciens orthopédistes) a choisi bologna comme le siége du 3éme Congrés mondial de l'I. S. P. O. qui aura lieu du 28 septembre au 4 octobre 1980. Par son haut niveau scientifique, assuré par des spécialistes connus dans le monde entier, par son examen des progrès récents dans tout le secteur de la réhabilitation soit du point de vue technique-scientifique soit du point de vue administratif, et par le vaste échange d'expériences, le Congrès répond complètment à l'esprit de la Décade des Handicapés et représente le prélude idéal à 1981, l'année des Handicapés proclamée par l'ONU.
On a au programme: des *cours d'instruction* pour offrir aux médecins, aux chirurgiens, aux techniciens orthopédistes, aux thérapeutes et aux bio-ingénieurs un ensemble d'informations ayant une utilité immédiate pour le milieu clinique; des *seances scientifiques* pour l'échange de données sur des études et des recherches récentes; des *symposiums* d'approfondissement sur des sujets spécifiques d'intérét commun; des *sessions plenieres* pour l'illustration de la part de spécialistes des progrès atteints dans les trois dernières années dans le domaine de la prothétique et de l'orthétique.
Il s'agit d'une occasion d'importance exceptionnelle pour contacter des collegues du méme secteur, en utilisant tous les conforts offerts par le moderne Palais des Congrès de Bologna, une ville qui est le centre de l'orthopédie et de la prothétique en Italie.
La traduction simultanée en italien, anglais, français, et en allemand permettra aux participants de suivre les sujets traités dans quattre salles différentes. On envisage aussi des traductions en espagnol et en japonais.

La Sociedad Internacional de Protesis y Ortesis (I. S. P. O.), en colaboración con la INTERBOR (Unión Internacional de los técnicos ortopédicos) ha elegido Bologna como sede del 3° Congreso mundial del I. S. P. O. que tendrá lugar del 28 septiembre al 4 octubre 1980. Debido a su elevado nivel cientifico, gracias a la participación de los más conocidos especialistas en campo mundial, debido a su actualización en los progresos más recientes de todo el sector de la rehabilitación, sea bajo el aspecto técnico-cientifico como administrativos, y por el vasto intercambio de experiencias, el Congreso respeta en lleno el espiritu de la Década de los Inválidos y representa un preludio ideal al 1981, año de los Inváido proclamado por la ONU.
En programa hay: *cursos de instrucciones* para ofrecer a los médicos, a los quirurgos, a los técnicos ortopédicos, a los terapeutas y a los bioingenieros un conjunto de informaciones de utilidad immediata para el ambiente clinico; *reuniones cientificas* para el intercambio de datos sobre estudios y búsquedas recientes; *simposios* de profundimientos sobre temas especificos de común interes; *asambleas plenarias* para la ilustración por parte de los especialistas de los progresos que se han producido en los últimos tres años sea en protesis como en ortesis.
Se trata de una ocasión de excepcional importancia para un intercambio de informaciones entre colegas del mismo campo, aprovechando todas las comodidades ofrecidas por el modernisimo Palacio de los Congresos de Bologna, ciudad que es el centro de la ortopedia y de la protética en Italia.
La traducción simultanea en italiano , inglés, francés y alemán dará la posibilidad a los partecipantes de poder seguir los temas tratados en cuatro aulas diferentes.
Se prevee también lá traducción en español y japones.

SATURDAY	SEPT. 27	**all day:** registration	
SUNDAY	28	**morning:** registration	**evening:** reception
		afternoon: opening ceremony opening of exhibition (open until 6 p.m.)	
MONDAY	29	**morning:** 8.00–10.00: instructional courses 10.30–12.30: plenary session	
		afternoon: paper sessions and symposia	
TUESDAY	30	**morning:** 8.00–10.00: instructional courses 10.30–12.30: plenary session	
		afternoon: paper sessions and symposia	
WEDNESDAY	OCT. 1	**morning:** 8.00–10.00: instructional courses 10.30–12.30: plenary session	
		afternoon: free	
THURSDAY	2	**morning:** 8.00–10.00: instructional courses 10.30–12.30: plenary session	**evening:** congress dinner
		afternoon: paper sessions and symposia	
FRIDAY	3	**morning:** 8.00–10.00: instructional courses 10.30–12.30: plenary session	
		afternoon: paper sessions and symposia	
SATURDAY	4	**morning:** plenary session closing ceremony	

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Prosthetics and Orthotics International, 1979, 3, 126–136

Total surface bearing self suspending above-knee sockets*

R. G. REDHEAD

The Biomechanical Research and Development Unit, London.

Abstract

A new type of above-knee socket has been designed to provide total surface support and to dispense with ischial bearing as the primary weight bearing area. The socket is based upon the hypothesis that if the soft tissues of an above-knee stump are adequately supported in a suitably shaped container they will behave under load as an elastic solid with low stiffness.

A method has been devised for taking the cast for the new type socket using an elastic sleeve as a "compliant socket". The grip of the elastic sleeve and the use of traction weights deform the stump tissues to the required shape while the cast is setting. The results of laboratory measurements of the transinterface pressures in these sockets under axial loading conditions have correlated well with the figures forecast by calculation. The new socket is now available to patients at a number of Centres in England.

Introduction

In 1967 the then Director of the Biomechanical Research and Development Unit at Roehampton, Dr. D. S. McKenzie, agreed to the start of a project to develop a total surface bearing self-suspending above-knee socket in which the load would be distributed as evenly as possible, so avoiding the discomfort associated with high loads concentrated in local areas. The project required that retention of the socket on the stump should be achieved without a significant adverse pressure gradient i.e. there should be no proximal constriction and there should be adequate stability of the stump tissues within the socket.

All correspondence to be addressed to Mr. R. G. Redhead, The Biomechanical Research and Development Unit, Roehampton Lane, London SW15 5PR, U.K.

At the start of the project the performance requirements for the stump/socket interface were defined as follows:

(a) To transfer the axial load of the body weight and bending and rotational loads arising during the walking cycle.

(b) To transfer the distraction loads due to the weight of the limb when the patient lifts the prosthesis off the ground.

(c) To eliminate lost motion between the surface of the stump and the socket wall.

(d) To reduce to a minimum the displacement of the femoral shaft within the soft tissues of the stump under the loads imposed during the walking cycle.

Axial compression loading

For the purpose of prosthetic fitting, the anatomy of the above-knee amputation stump was considered in a simplified form as consisting of an outer limiting envelope, the skin and deep fascia; a roof, the side wall of the pelvis; and a central rigid strut, the femoral shaft.

Contained within the outer envelope and surrounding the central strut is a composite of "soft tissues". At the "micro" level all the soft tissues are composed of a complex of fluid filled compartments, or cells. At the "macro" level the soft tissues are separated into various anatomical compartments by fascial layers and intermuscular septa. It was suggested that the shape of the soft tissues of the stump could be altered, within limits, by applying minimal stress without changing their volume. Figure 1 shows diagramatically the difference in the soft tissue distortion pattern when a proximal and a distal stress are applied to the free stump.

*Based on a paper presented at the ISPO International Course on Above-knee Prosthetics, Rungsted, November, 1978.

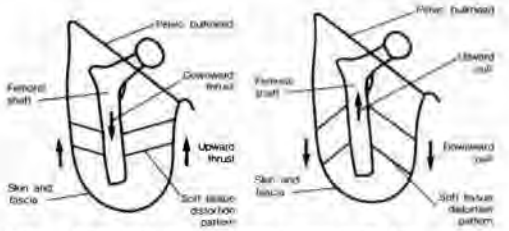


Fig. 1. Anterior views of free stump with, left, applied proximal stress and right, applied distal stress.

The changes in stump shape are limited by the arrangement of the tissue structures bounding the fluid compartments. Under minimal stress between points A and B on the stress/distortion curve (Fig, 2, left) the stump might be expected to behave as a "bag" containing fluid. When sufficient stress is applied to bring the tissue structures bounding the fluid compartments under tension the stress/distortion characteristics of the stump might be expected to change and exhibit a sharply increasing degree of stiffness in the direction of the applied stress.

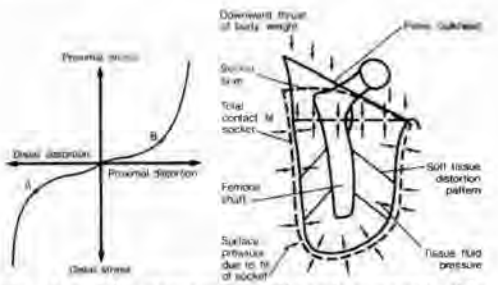


Fig. 2. Left, tissue/stress distortion relationship. Right, anterior view of total surface bearing self suspending above-knee socket.

If an above-knee stump were to be contained within a rigid total contact socket made to conform to the shape and volume of the stump between the two limits of distortion marked "A" and "B" on the curve in Figure 2, left, it was suggested that it should be possible to treat the stump, in respect of axial compression loads, as a fixed volume "bag of fluid" with a high bulk modulus which could be pressurized by the transference of the load of the body weight across the stump/socket interface. The patient would be supported upon the relatively incompressible volume of the stump contained within the socket (Fig. 2, right). Localized skeletal support in the form of ischial bearing would not be necessary, and the pressure would be evenly distributed across the whole of the stump/socket interface.

In these circumstances the forces at the interface should be acting normal to the surface of the stump in all areas, and there should be no element of shear stress at the interface to cause discomfort to the patient. The pressure required to support the patient's weight on such a socket would be independent of the surface area of the stump/socket interface, but would be related to the projected area of the oblique side wall of the pelvis, the "roof" of the stump.

This area would not be easy to determine, and so it was considered acceptable to take the area of the cross-section of the stump at the level of the posteromedial socket brim as being an approximation that could be measured. For example, if the circumference of the stump at this level was 46 cm (18 in)—then the area of the cross-section would be approximately 180 cm^2 (28 in^2) and a loading of 42 kN/m^2 (6 psi) would support the total body weight of a patient weighing 75 kg (168 lbs). The fraction of this total load supported by the area of the cross-section of the femoral shaft, would be transmitted back through the hip joint and might provide "feedback" of proprioceptive information.

If it was true to assume that the stump would behave as a hydrostatic system, the pressure throughout the system should be equal and therefore the pressure over all the area of the stump/socket interface should also be 42 kN/m^2 (6 psi), providing the load was a pure axial load and any intertial forces, bending loads or torque effects were excluded. The pressure would be expected to fall to 21 kN/m^2 (3 psi) if this patient stood with his weight evenly divided between both feet. To maintain a total contact fit under the distraction load of the weight of the prosthesis, it was suggested that the pressures across the stump/socket interface in the nil axial load state would have to be of the order of 2.1—3.5 kN/m^2 (0.3—0.5 psi).

Assuming that the arterial systolic inflow pressure to the stump, when a patient is standing, is about 150 mm Hg (21 kN/m^2 or 3 psi) the range of stump/socket interface pressures suggested seemed reasonable. It must be remembered that the patient only sustains all his weight on one leg transiently as he shifts his support from one foot to the other. The periods

of lower pressure across the stump/socket interface would permit adequate time for blood circulation through the stump. The variation in pressure should provide a pumping action that would promote the return of venous blood and lymph to the general circulation.

On the basis of the above hypothesis total surface bearing appeared to be a possible means of carrying the axial load of the body weight without the need to use skeletal support or support from soft tissue as in a plug fit socket.

Socket retention

It was suggested that retention of a self-suspending total surface bearing above-knee socket would depend upon two factors.

Firstly, friction at the stump/socket interface should retain the socket on the relaxed stump for levels of distraction loading up to the weight of the prosthesis. This factor could be augmented temporarily by voluntary contraction of the stump muscles to increase the pressure across the interface.

Secondly, for distraction loads in excess of the weight of the prosthesis when friction at the interface might not prevent the socket from slipping on the stump, atmospheric pressure could be an additional factor holding the socket in place. For the latter to be effective the socket wall would have to be airtight and a total contact fit maintained between the socket and the side of the stump. The end of the stump could not then separate from the bottom of the socket unless air was able to enter.

Under distraction loading the tissue distortion would take the form of an elongation of the stump which, if its volume remained constant, would be associated with a decrease in its diameter. If the socket was made to conform to the shape of the stump when it was not stressed or was stressed proximally, the application of a distraction load would cause the stump to draw away from the socket wall as the tissues elongated and the stump decreased in diameter.

To achieve the required fit the socket would have to be made to conform to the shape of the stump after a distal stress at least equal to the weight of the prosthesis has been applied to it, together with a radial compression of 2.1—3.5 kN/m^2 (0.3—0.5 psi) uniformly distributed over the sides of the stump. Under these conditions the stump would be distorted distally to the point "A" on the stress/distortion curve in Figure 2, left, where its stiffness to distal stress was rising steeply, and therefore any increase in distraction loading would cause little further distortion to occur. The surface of the stump should therefore stay in contact with the socket wall. In this manner the friction fit and the airtight seal along the length of the stump/socket interface should retain the socket securely on the stump without an adverse pressure gradient.

Stump/socket stability

Elimination of lost motion between the surface of the stump and the socket wall would be achieved if the socket was a total contact fit, but in addition there should be minimal internal lost motion between the femoral shaft and the stump/socket interface. Anteroposterior and mediolateral stability of the femoral shaft should be greatest if the horizontal distance from the shaft to the interface is a minimum in all directions. The effect of the philosophy proposed for the new socket would promote this situation.

It was also suggested that the increased stiffness of the soft tissues consequent upon distal stressing could be a factor maintaining skeletal stability within the stump. In respect of the bending loads, it was suggested that horizontal displacement of the distal femoral shaft might be resisted by a rise in pressure within the soft tissue on the side to which the bone is trying to move. These tissues could not displace to accommodate the femoral shaft because superficial distortion would be prevented by the fit of the socket, and internal displacement would be limited by internal tissue connections. These factors would be expected to modify the uniform pressure distribution exhibited over the stump/socket interface under axial load alone.

Vertical stability of the femoral shaft remnant, i.e. internal piston action under axial loading, should be achieved because the distortion resulting from the distal stressing of the soft tissue would prevent further upward movement of the skeleton in relation to the socket when an axial distraction load is applied (Figs. 1 left and 2 left). Downward movement of the skeleton in relation to the socket under an axial compression load would be prevented by the relatively incompressible volume of the stump being contained within the rigid, fixed shape, fixed volume total contact socket. These factors should ensure that there is minimal "internal

piston action" of the femur within the stump tissues. Rotational stability about a vertical axis would depend upon "locking" the socket to the stump by the shape of the socket brim.

If the above hypotheses are true the concept of total surface bearing does not appear to produce a conflict between the requirements for weight bearing, socket retention and stability.

Socket casting

Work done on total contact ischial bearing sockets has shown that it is essential to make these sockets to a cast of the stump. At present this is the only simple way for the prosthetist to record the shape/volume relationship of the stump.

If a socket is made direct from a simple "laid on" cast, it usually appears to be too large to be self-retaining, even when a casting jig has been used to impose a brim shape. This is because the cast reproduced the shape and volume of the stump either unstressed or under proximal stress if the cast was taken under conditions of weight bearing.

For a successful self-suspending socket to be made by these casting techniques, it is necessary to carry out a "reduction rectification" procedure on the positive cast to sculpture it to the shape and volume the stump would assume on the application of at least part of the distraction load due to the weight of the limb and a radial compression load to maintain the total contact fit of the socket. Sockets cast with a casting brim usually depend primarily upon the tightness of their top fit to hold the suction seal. Appoldt and Bennet (1967) have shown in their studies on the stump/socket interface pressures that these sockets often have an adverse pressure gradient along the length of the stump, being tighter at the brim than distally. The rectification procedures entail removing plaster in certain areas, either according to tables arrived at on an empirical basis, or depending upon the skill and experience of the individual carrying out the plaster work. Plaster may have to be added to the distal end of the cast to avoid the risk of excessive end loading and to accommodate the distal displacement of soft tissue. It was felt it would be difficult to achieve the degree of accuracy for the stump/socket shape/volume relationship for a total surface bearing socket if such techniques were used to define the final shape of cast.

The philosophy of total surface bearing self-suspending sockets called for the stump to be stretched distally when the cast is taken, making the stump longer and thinner as compared with its shape in the unstressed state. It was therefore suggested that if calculated stresses could be applied to the stump tissues, and the tissues allowed to deform to a point of balance with them, the shape could then be "fixed" by applying a plaster soaked wrap over the stump. In this manner the stump itself would dictate the physical dimensions of the socket necessary to produce the required conditions at the stump/socket interface.

To achieve the necessary distortion it was proposed that the stump should first be fitted with a "compliant socket" made from an open-ended sleeve of elasticated material, the stretch/tension characteristics of which could be related to stump diameters and so to the required radial loading. The radial pressure on the stump would be achieved by the hoop stress resulting from the elastic pressure sleeve being stretched over the surface of the stump. When an elastic material is stretched over a curved surface the radial pressure produced against the surface varies directly as the tension in the elastic and inversely as the radius of curvature of the surface, (Fig, 3, top). A factor contributing to distal stressing of

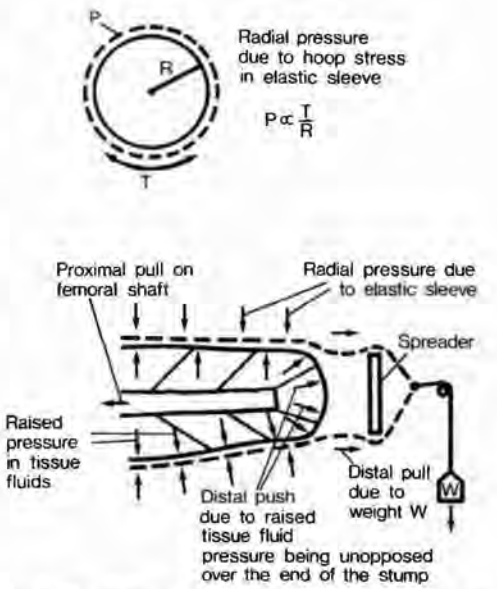


Fig. 3. Tissue stressing. Top, radial pressure. Bottom, additional distal force applied by means of a weight (see text).

the tissues would be the hydrostatic pressure raised within the tissues as the result of the radial pressure exerted by the elastic pressure sleeve.

This force, acting distally, would depend upon the area of the open distal end of the pressure sleeve and upon the shape of the end of the stump. This force alone might not be sufficient to provide adequate distal stretching of the tissues, and an additional distal force might have to be applied by pulling on the surface of the stump (Fig. 3, bottom). It was suggested that the additional distal stretching force should equal the weight of the prosthesis plus a safety factor. A total figure of 4½ kg. (10 lbs) might be suitable for the average British above-knee prosthesis.

The cast could be taken with the patient standing but it was suggested it might be better done with the patient lying down because:

It would be more comfortable for the patient and would make it easier for him to relax the stump muscles to allow the distal distortion of the stump tissues.

It would give the prosthetist ample time to work, without the patient becoming tired.

The development of gravitational oedema in the distal stump tissues would be avoided.

It should be easy to impose any desired deformation to brim shape by means of a simple casting board adjustable for height and angle of flexion, on which could be mounted adjustable formers as required. There should be no need to have a complete proximal casting brim with the associated risk of proximal constriction. The final brim shape could be achieved by rectification to the positive cast.

Alternative methods of obtaining the brim shape would be to cast the stump with the patient standing and to mould the negative cast by hand while the plaster cast was setting on the stump, or to combine the use of an adjustable brim shape with an elastic sleeve as a "compliant" socket.

Experimental procedure

The object of the investigation was to examine the distribution and level of loading across the stump/socket interface under axial compression load alone, excluding as far as possible any static bending loads in an anteroposterior or mediolateral plane, or any torque load about a vertical axis. The pressure transducers selected were Ferranti silicone etched diaphragm type ZPT50A (7/8" long × ¼" diameter), chosen because they exhibited high linearity and a good frequency response over the pressure range 0—350 kN/m² (0—50 psi), while being small in size and weight. Sixteen holders for the pressure transducers were built into the walls of each socket (Fig. 4). Nine transducers were available and plugs were screwed into the empty holders to maintain the suction seal. The sockets were fitted into trial legs incorporating facilities for alignment adjustment. A simple adjustable

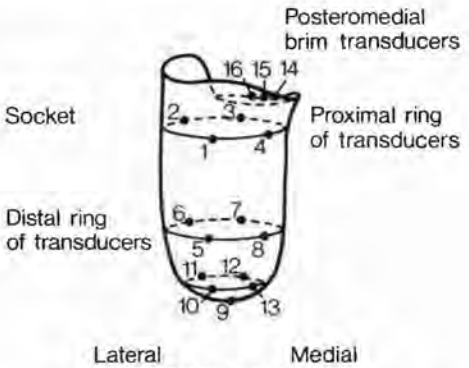


Fig. 4. Transducer locations.

socket mounting system was used that left the socket fully exposed (Fig. 5, left). The experiments were carried out using the apparatus shown in Figure 5, right.

The patient stood on the platform of the experimental rig with the prosthetic foot on a weighbridge, and with his own foot placed so that he was balanced and felt that his stump was


Fig. 5. Left, trial prosthesis showing adjustable socket mounting, transparent socket and built-in transducers. Right, subject on experimental rig, note overhead hoist and harness with weights attached.

pressing straight into the socket with no sense of twisting. The patient was asked to adjust his weight on the prosthesis until the weighbridge recorded half his body weight.

To increase the axial compression load on the prosthesis up to full body weight, an adjustable overhead hoist was used to load an increasing weight on to a strong shoulder/waist belt harness worn by the patient. When the added weight was equal to the patient's body weight, the patient's own leg and his artificial limb were each supporting an axial compression load equal to the patient's own weight. The added weight was then removed progressively.

To reduce the axial load on the prosthesis below half his body weight, the patient was lifted by the shoulder suspension harness attached to the overhead hoist. The weight on the prosthesis was decreased progressively until it was less than 5 kilograms.

The patient was then lowered until he was again standing with his weight equally divided between his two legs. Throughout the loading cycle a series of simultaneous readings were recorded from the 9 pressure transducers and from the weighbridge under the prosthetic foot.

When the 9 transducers were mounted in holders 1, 2, 3, 4, 5, 6, 7, 8 and 9 (Fig. 4), information was recorded on the level and distribution of pressure over the side walls of the socket with a spot check on the terminal pressure from the transducer in holder 9.

When the transducers were mounted in holders 1, 2, 3, 4, 14, 15, 16, 8 and 9 (Fig. 4), information was recorded on the level and distribution of pressure over the posteromedial socket brim with a check on the pressures on the side wall and distal end of the socket.

When the transducers were mounted in holders 1, 2, 3, 4, 9, 10, 11, 12 and 13 (Fig. 4), the terminal pressure distribution was recorded in more detail and a check was kept on the side wall pressures in the proximal part of the socket.

Clinical observations

During the experimental sessions the surface of the stump was examined through the transparent wall of the socket.

The socket was an accurate total contact fit on the stump at all levels of axial compression loading and there was no movement of the skin of the stump in relation to the socket wall. When there was no applied load the whole surface of the stump was a normal pink colour. As an axial compression load was applied, the surface of the stump, including the distal end, became uniformly pale. With the removal of the load a uniform pink blush reappeared over the whole surface of the stump. None of the patients complained of any discomfort at the stump/socket interface during the loading cycles.

Repeatability of the results

If the results of the experiments were to be meaningful they should be capable of being reproduced at repeated experimental sessions.

The first two graphs are examples of plots done to demonstrate this. Graph 1 shows the general pattern of pressure distribution at the stump/socket interface for various applied loads, and Graph 2 shows the results of a repeat experiment done 3 months later on the same patient wearing the same socket.

The main features of the plots for these two experiments are similar. The maximum difference in pressure recorded at any one of the transducers for similar applied loads during the two experiments was about 7 kN/m^2 (1 psi) while many of the transducers recorded almost identical pressures throughout the range of loading.

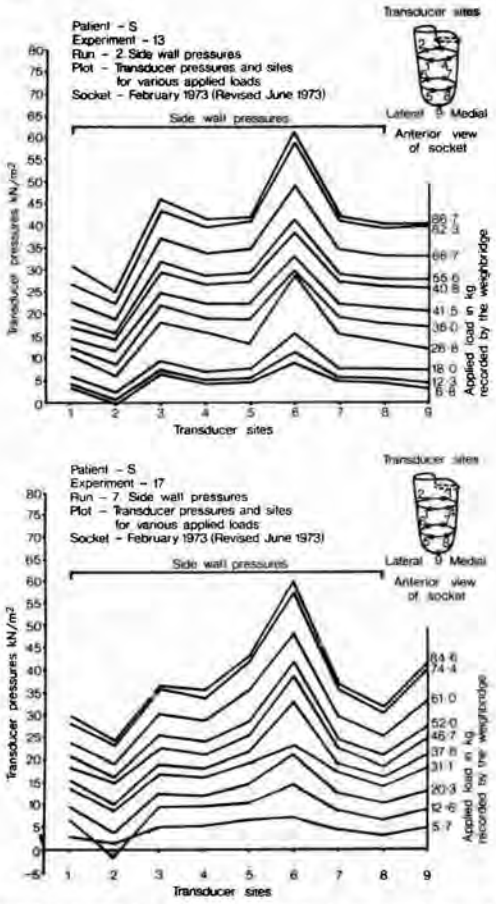
The effect of changes of limb alignment on pressure distribution at the stump socket interface.

A prime requirement of the experiments under discussion was that the applied load should be an axial compression load alone and that as far as possible other directions of loading should be excluded.

It was felt at the time of setting up the experiments that the patient's sensations were as good an indication as any that the stump was pushing straight into the socket under a direct axial compression load, without any unwanted bending loads.

To check whether this was a valid assumption, three loading cycles were carried out consecutively on one patient with the same application of the socket, each with the patient's feet separated by a different amount.

Graph 3 top, centre and bottom, show examples of the results obtained from a patient when his heels were separated by 14 cm, 24 cm and 57 cm.



Graphs 1 and 2. Plots to demonstrate repeatability of the experiment. Graph 1, top: general pattern of pressure distribution for various applied loads. Graph 2, bottom: repeat experiment three months later on the same patient with the same socket. The main features of the plots are similar (see text).

The overall pattern of the 3 plots is similar, but a closer examination of the pressures recorded under maximum load at transducer sites 2 and 4, and 6 and 8 in Graph 3, top shows that at the proximal pair of transducers there was a nett pressure difference of 7.8 kN/m² acting laterally, and that at the distal pair of transducers there was a nett pressure difference of 4.0 kN/m² acting medially. These figures indicate the presence of a lateral bending load.

If the pressures recorded at the same sites in the next two graphs are examined in a similar manner, the figures were 7.5 and 2.3 kN/m² for Graph 3, centre and 4.5 and 1.7 kN/m² for Graph 3, bottom showing that as the patient's feet were moved further apart the lateral bending load decreased. The nett pressure differences recorded in Graph 3, top at transducer sites 1 and 3 and 5 and 7 under the maximum applied load were 7.5 kN/m² acting anteriorly and 0.8 kN/m² acting posteriorly. The comparable figures are 6.8 and 0.9 on Graph 3, centre and 5.4 and 1.1 on Graph 3, bottom. These figures indicate the presence of a posterior bending load, but its relation to the anterior/posterior position of the foot cannot be inferred as the feet were kept level throughout the experiments presented on Graph 3, top, centre and bottom.

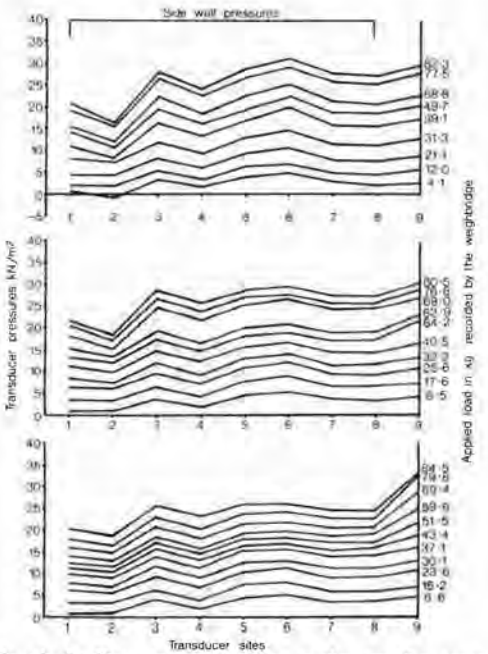
It was therefore concluded that it must not be assumed that the patient can detect the presence of minor bending loads. An examination of Graphs 1, 2 and 4 shows that bending loads were also present in these experiments.

Results

Pressure distribution

Graphs 1—4 show typical distributions of pressure over the side walls of sockets recorded at several different experimental sessions.

The same overall pattern is shown to a greater or lesser degree on all the plots. The lowest pressures were always recorded at transducer site 2 on the lateral side of the proximal part of



Graph 3. Consecutive loading cycles on the same patient with heels separated by different amounts. Top 14 cm, centre 24 cm and bottom 57 cm.

the socket, and the highest pressures were always recorded at transducer site 6 on the lateral side of the distal part of the socket. None of the stumps in question were myoplastic stumps and transducer holder 6 was positioned opposite the area on the stump overlying the lateral side of the distal end of the femoral shaft. It seems likely that a contributory factor to the higher pressures in this area was the small amount of soft tissue cushioning available to distribute the load.

On Graphs 1 and 2, under an applied load of about 85 kg. transducer 6 recorded 60 kN/m² (8.6 psi) and transducer 2 recorded 25 kN/m² (3.6 psi) giving a pressure range on these plots of 35kN/m² (5 psi). If the results from the patient shown in Graph 3, top are looked at in the same way, the maximum side wall pressue recorded at transducer 6 under an applied load of about 84.5 kg was 26.5 kN/m² (3.8 psi), and the lowest pressure recorded at transducer 2 was 18 kN/m² (2.6 psi), giving a pressure range on this plot of 8.5 kN/m² (1.2 psi). If future experiments could achieve a total absence of bending loads, these pressure differences might be expected to decrease considerably or even disappear.

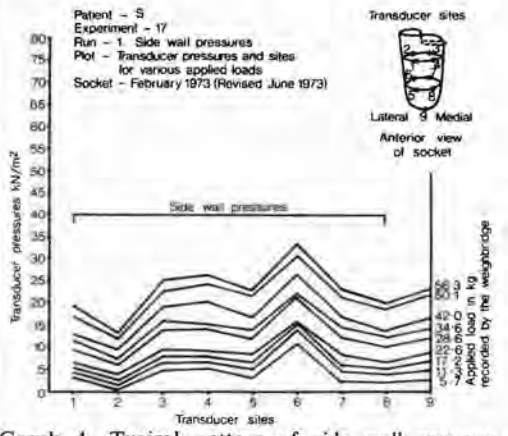
Graphs 4—10 show in more detail the results of an experiment on one patient carried out without removing the socket, making it possible to compare the various plots. Graph 4 shows the typical pattern of pressure distribution over the side wall of the socket previously described. Graph 5 shows a repeat run with transducers moved from holders 5, 6 and 7 in the side wall of the socket to holders 14, 15 and 16 around the posteromedial socket brim.

This plot demonstrates clearly that the pressures around the posteromedial brim of the socket were of the same order of magnitude as those over the side wall of the socket and the same as those recorded by the central terminal transducer in holder 9.

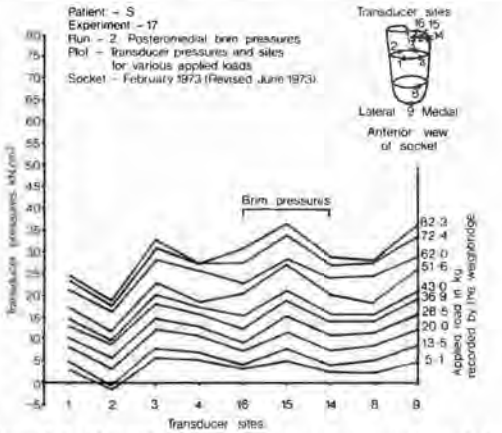
The average of the pressures recorded at the three brim transducers under an applied load of 82.3 kg was 32.2 kN/m² (4.6 psi) which compares with the average of the pressures recorded by the other 6 transducers of 28 kN/m² (4 psi).

The maximum pressure over the posteromedial socket brim recorded at transducer site 15 under an applied load of 82.3 kg was 36.6 kN/m² (5.2 psi) compared with 36 kN/m² (5.1 psi) at the terminal transducer.

Appoldt and Bennet (1967), in their studies of static socket pressures on a patient wearing a



Graph 4. Typical pattern of side wall pressure distribution.

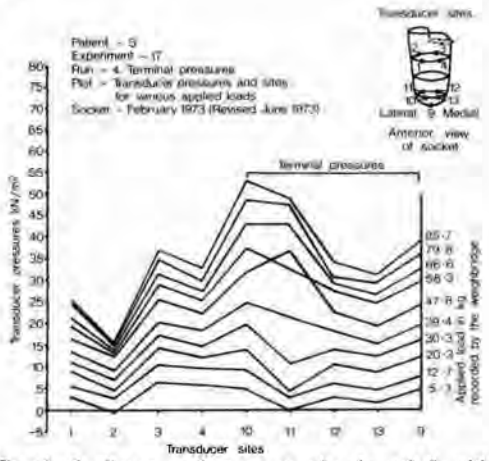


Graph 5. Repeat run with same socket as in Graph 4 with transducers moved from side wall holders 5, 6 and 7 to holders 14, 15 and 16 in the posteromedial brim.

brim bearing total contact socket, recorded a maximum pressure over the posterior socket brim of 42 kN/m² (6 psi) when half the patient's weight was on the limb.

In the total surface bearing socket under study in Graph 5 the peak pressure on the posteromedial socket brim under an applied load equal to about half the patient's body weight (43 kg), was 21.5kN/m² (3.1 psi) which is half the figure recorded by Appoldt and Bennett, although as they do not state the weight of their patient or the applied load, it is not possible to make an exact comparison.

Graph 6 shows that the highest pressures in this socket were recorded over the lateral side of the distal surface of the socket at transducer sites 10 and 11. The maximum pressure of 53.3 kN/m²



Graph 6. Same socket as graphs 4 and 5 with transducers in terminal holders 10, 11, 12 and 13.

(7.6 psi) occurred at transducer 10 under an applied load of 85.7 kg. The corresponding figure for transducer 11 was 48.6 kN/m² (6.9 psi).

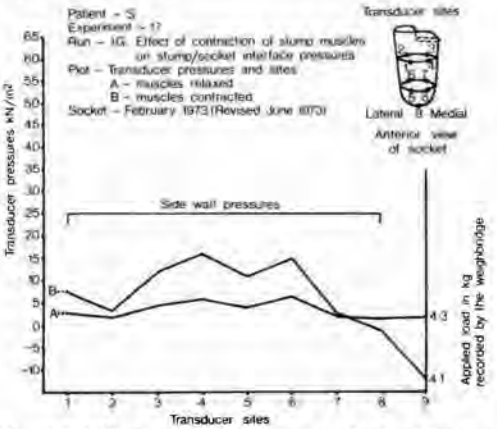
The other terminal transducers at sites 9, 12 and 13 recorded pressures of 38.6 kN/m² (5.5 psi), 34.2 kN/m² (4.9 psi), and 31.2 kN/m² (4.4 psi) respectively, which are comparable with 36.9 kN/m² (5.3 psi) and 32.5 kN/m² (4.6 psi) recorded at transducer sites 3 and 4 in the side wall of the socket.

The higher pressures recorded by transducers 10 and 11 may be related to the non-myoplastic nature of the stump, providing only skin and subcutaneous tissue to separate the end of the femur from the socket wall. The patient felt no discomfort over the end of the femur and there was no mark on the skin due to the higher pressure in this area when the stump was examined after removal of the socket at the end of the experiment.

The effect of muscular contraction

Graph 7 shows how the patient was able to raise the pressures generally over the side wall of the socket by tensing his stump muscles. With the muscles relaxed the average of the side wall pressures recorded at transducers 1—7 under an applied load of 0.5 kg (weighbridge reading of 4.3 kg less the weight of prosthesis of 3.8 kg=0.5 kg) was 3.6 kN/m² (0.5 psi), with a peak pressure of 6.1 kN/m² (0.9 psi) recorded at trasducer 6.

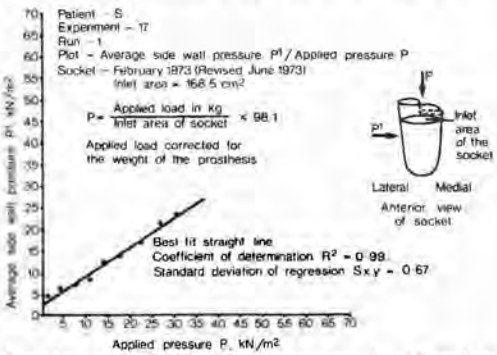
When the patient tensed his stump muscles, the average side wall pressure was raised to 9.6 kN/m² (1.4 psi), with a peak pressure of 15.6 kN/m² (2.2 psi) at transducer site 4. These increased pressures occurred in spite of a decrease in the applied load of 0.2 kg. At the same time, the terminal pressure recorded at transducer 9 dropped from 1.9 kN/m² (0.3 psi) to —12.3 kN/m² (−1.7 psi) as the tensed muscles attempted to shorten. The pressure recorded at transducer 8 dropped in a similar manner probably because of the retraction of the distal end of the adductor muscle group in this non-myoplastic stump.



Graph 7. Effect on side wall pressure of patient tensing his stump muscles.

The relation between side wall pressures and applied loads

Graph 8 shows a plot of the average of the pressures recorded by the transducers in the side wall of the socket against the calculated pressure which should result from the even distribution of the applied load over the area of the inlet to the socket. There appears to be a simple straight line relationship between the two sets of figures.



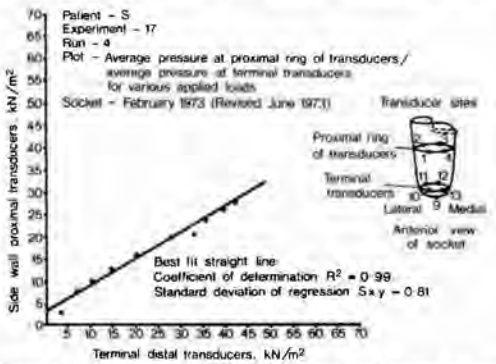
Graph 8. Average side wall pressures plotted against pressure resulting from even distribution of applied load over socket inlet area.

Previously it had been suggested that when there is no applied axial load the average pressure over the side wall of the socket should be between 2.1—3.5 kN/m² (0.3—0.5 psi).

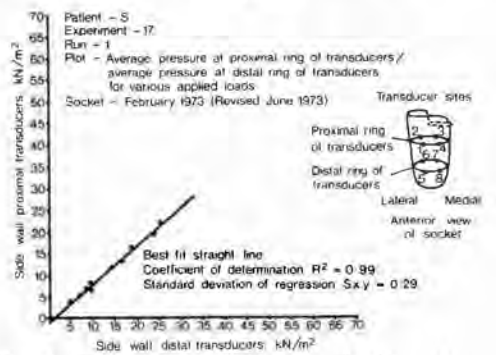
This best fit straight line cuts the vertical axis at 2.34 kN/m² (0.33 psi) which would be the average side wall pressure in this socket without an applied load. The ratio of the side wall pressure to the applied pressure when the applied load was 51 kg was 0.74:1 which is consistent with the tissues of the stump behaving as an elastic solid with low stiffness rather than as a fluid.

Pressure gradients along the long axis of the socket

Graph 9 and 10 examine the pressure distribution along the length of the socket. In Graph 9 the average pressures at the proximal ring of transducers in the side wall of the socket are plotted against the average terminal pressures. This best fit straight line cuts the vertical axis at 2.85 kN/m² (0.4 psi). For side wall



Graph 9. Average pressure at proximal transducers plotted against average pressure at terminal transducers.



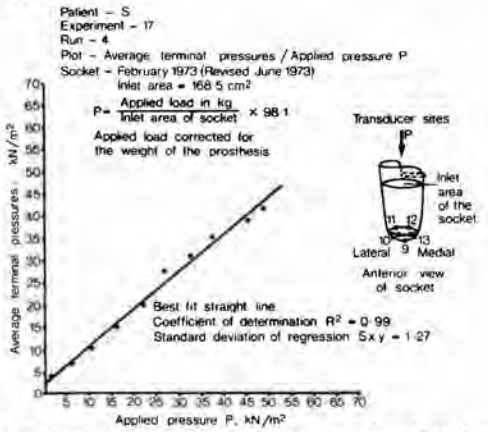
Graph 10. Average pressure at proximal transducers plotted against average pressure at distal transducers.

pressures of less than 7.5 kN/m² (1.1 psi), when the applied load is 20.3 kg, there is a small adverse pressure gradient between the terminal and proximal side wall transducers, but for applied loads of more than 20.3 kg there is a small positive pressure gradient.

Graph 10 examines the pressure distribution along the length of the socket wall by plotting the readings recorded at the proximal ring of transducers against those recorded at the distal ring of transducers. Throughout the range of loading the average pressures recorded at the distal ring were slightly higher than those recorded at the proximal ring, indicating a small positive pressure gradient along the length of the side wall of the socket.

The relation between terminal pressures and applied loads

Graph 11 is a plot of the average terminal pressures in the socket against the calculated pressures due to the distribution of the applied load over the area of the inlet to the socket. The plot shows that the ratio of the average terminal pressure to applied pressure is very close to 1:1, for example under an applied load of 51 kg it is 0.92:1.



Graph 11. Average terminal pressures plotted against pressure of applied load over socket inlet area.

Conclusions

The results of the experiments on these patients and their sockets indicate the following:

The experimental technique used was capable of giving consistent results from repeated experimental sessions.

Small changes in mediolateral limb alignment affect the pressure distribution in the socket.

The aim to produce a more uniform distribution of pressure at the stump/socket interface under static axial compression loads than exists in present brim bearing and total contact fit sockets, appears to have been achieved in spite of the presence of small lateral and posterior bending loads. In particular ischial bearing has been avoided. The higher pressures recorded in the region of the cut end of the femur may have been aggravated by the non-myoplastic nature of the stumps under examination. It is felt that these pressure inequalities would not have occurred if the load had been an axial compression load alone applied to a properly fashioned myoplastic stump in which the distal end of the femoral shaft had been stabilized and adequately covered with soft tissues.

The soft tissues of the stump behaved generally as an elastic/solid with a high bulk modulus and a low stiffness rather than as a fluid.

When there was no applied axial load, the stump tissues were pre-stressed by an average radial pressure of 2.34 kN/m^2 (0.33 psi), which accorded with the figures suggested in the original hypothesis.

The patient was able to increase the average radial pressure on the side wall of the socket by a factor of 2.7 when the applied load was 0.5 kg, thus demonstrating his ability to increase at will the security of socket retention on the stump.

There was no adverse pressure gradient recorded between the terminal transducers and the proximal side wall transducers at applied loads of less than 20.3 kg is not considered significant.

Clinical experience

Total surface bearing above-knee sockets were fitted experimentally to volunteer patients from 1969 to 1972, during this period some 200 casts were taken.

Since 1974 the sockets have been available for routine issue at a number of limb Centres in England.

The stumps of patients using these sockets have remained healthy and there is no evidence of progressive wasting of the soft tissues of the stump as a result of using total surface bearing sockets for up to 10 years.

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Alignment of the above-knee prosthesis*

J. FOORT

Department of Surgery, Division of Orthopaedics, University of British Columbia, Vancouver.

Alignment will be discussed in terms of what happens as a result of specified changes in the relative positions of the prosthetic components.

As a person walks on his prosthesis the conditions which predominate at various stages can be identified:

(a) the conditions of no-load, or *swing phase,*
(b) the conditions under which there is full loading or *stance phase,*
(c) the capacity to control the prosthesis easily or *voluntary control*
(d) the conditions under which little or no control can be exerted.

Alignment will be considered with these conditions in mind in the hope that visualization of what is happening as you work toward *optimizing comfort, saving energy and normalizing the appearance of gait* will be made easier.

As you watch the amputee walk, everything happens at great speed, and there is need for some sort of anchorpoints for your observations. You depend on motions—*motions en masse, motions between the prosthesis and the wearer, motions between the prosthetic parts.* You may however, develop the habit of feeling the effects of what you see as *forces.* As I watch a person walk with the view to changing his alignment I imagine *how the prosthesis or its parts would bend if they could, and how they would want to move to modify this tendency if they could.* From this I get clues as to what changes to make.

I have boiled it down to just a few things which I consider for change:

1. Length of the prosthesis.
2. Foot position—front, back, inward, outward.
3. Foot angle—toe up, toe down, toe in, toe out.

The stump is considered as the datum or reference system to which the position of parts is referred.

As for knee pivot position, (4) I think of this only in terms of the internal or external angulation of the axis from right angles to the line of progression. The forward or backward position has influence, but the prosthetist can easily pre-set this position with respect to the stump and leave refinement of stability to the positioning and angulation of the foot. Similarly, inversion and eversion relate more to cosmesis than function and can be neglected except in that sense.

Let us consider the effects of the changes we can make to the length, position and angle of the various parts; and as we do so, let us use the motions for clues while we sense these as forces necessary to give us the information we need for making a particular change.

1. Length of the prosthesis

The length of the prosthesis is affected not only by its axial elongation or shortening, but also by the inclination of the segments relative to one another. Thus, plantarflexing the foot makes the prosthesis seem longer at the end of the stride and shorter at the beginning of the stride. There is also a small effect which comes from inclining the shank toward the back or toward the front. Then, as changes are made in other segments of the prosthesis there may have to be a compensating change in the axial length of the prosthesis. But to be specific: *if the prosthesis is too long, the amputee will (a) rise up on the foot of the normal side in swing phase of the prosthesis or*

All correspondence to be addressed to Mr. J. Foort, Department of Surgery, Division of Orthopaedics, University of British Columbia, 2740 Heather Street, Vancouver, B.C. Canada, V5Z3J4.

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the prosthetic toe will scuff. (b) the pelvis will drop on the normal swinging leg side during stance phase on the prosthesis along with which there may be high forces in the crotch which cause pain, or (c) the amputee will space his feet to effectively shorten the prosthesis for stance phase and (d) swing the prosthesis through with it held out by hip abduction to help clear the foot and set it in the wide based position for the next weight bearing period.

If I shorten the prosthesis the amputee will (a) have increased voluntary control, (b) will usually shorten his stride, (c) will narrow his walking base, and (d) will lean laterally over the prosthesis to balance.

The best length depends on how long the stump is and how much voluntary control the amputee can and will exert over the prosthesis. A rule of thumb is that for a very long stump, equal length between the normal and prosthetic sides will be used unless there are other considerations which require it to be shortened (such as considerable activity on rough terrain). For progressively shorter stumps, the prosthesis is made shorter. The endpoint for shortening is determined by a balance between comfort, energy costs and cosmesis, factors which relate to amputee preference.

2. Foot position

As the prosthetic foot is displaced forward, the stride on the prosthetic side will be of INCREASED LENGTH. Conversely, *as the prosthetic foot is displaced toward the rear, the stride on the prosthetic side will be of DECREASED LENGTH.* Nor does it matter whether the foot position is altered by changing the inclination of the shank, as when a tilt is made in an alignment unit at the knee, or the foot position is changed by a translatory motion, as when a sliding mechanism is used in an alignment unit at the knee, even at the foot for that matter. It is the displacement of the foot relative to the stump that counts.

If the foot is displaced medially there are a variety of possibilities; *(a) the amputee may adduct the hip and walk with a narrow base.* Watch for gapping at the brim of the socket laterally and sense the low force situation there and how the forces would be high in the crotch and at the lateral side of the stump distally. He may catch himself early on the sound leg to "get away" from the crotch and distal pressures or *(b) he will abduct the hip, moving the foot laterally for relief and increased voluntary control; in this case he will lean over the prosthesis to gain balance or (c) he may stay erect, imparting a strong lateral push to his body with the sound leg as he enters weight bearing on the prosthesis.*

A variety of possibilities also exist if the foot is displaced laterally. The amputee will *(a) compensate by moving the foot toward the midline to decrease the walking base or (b) he will leave the foot displaced laterally and lean over the prosthesis to achieve balance. He may (c) leave the foot laterally positioned and merely stay erect, giving his body a lateral impulse with the sound leg as he enters the weight bearing phase on the prosthesis.*

You will notice that there is a degree of voluntary control over conditions which result from medial or lateral displacement of the foot. How the amputee responds to such changes in foot position depends to an important degree on the length and strength of his stump, or more importantly, on the capacity of the stump to tolerate the forces and to generate the forces for control of the prosthesis through hip muscle actions and shifts in the centre of gravity.

3. Foot angle

Toe down *imposes a tendency to SHORTEN the stride on the NORMAL side.* Toe up *has the opposite effect, LENGTHENING the stride on the NORMAL side.* You may have noticed among geriatric or weak amputees the tendency to step up to the level of the prosthesis with the normal leg so that the feet are together and then to step out again with the prosthesis. This asymmetry is his response to the forces being felt against the stump as the front lever of the foot creates an increased moment against the stump.

As you can see, the actions of the front and back levers of the foot are affecting the *timing* of the stride. You can easily sense this intuitively. Imagine the force-point moving from heel to toe as the prosthetic foot supports the amputee with the prosthesis at different angles of inclination as it passes from heel contact to toe support. Sense the "bending" effect this would have. And you must consider what is happening at the top end of the prosthesis too! As it happens, there is a corresponding force-shift from the *front* brim of the socket at heel contact when the inguinal crease area is a major weight bearing area to the *back* brim of the socket as weight is borne

through the toe of the prosthesis. *This synchronous tilting backward and forward of the line of action of forces through the prosthesis between heel contact and toe support has the effect of keeping forces acting on the knee pivot nearly constant when alignment is optimum, or at least within a good range for voluntary control over the knee.* Thus, when the foot angle as viewed from the side is correct, even when the line of action of forces falls *behind* the knee axis, stability is maintained. This is because during that period—when the short hind lever is in effect—both feet are in contact with the ground *(double support)* and hip extension on the prosthetic side can add a stabilizing force to the knee until, at the later stage when double support has ceased, the ground reaction point has moved sufficiently forward on the foot to set the line of action somewhat in front of the knee axis. You may have noticed how plantarflexing the prosthetic foot improves knee joint stability for amputees who fail to pull back on the hip to add the needed stabilizing force at the knee. In the German system of alignment, the amputee essentially toe-walks all the time, and the foot is displaced back to compensate. Foot angle and foot position as viewed from the side are significant factors in the comfort, energy consumption and cosmetic effect of the prosthesis.

I have often used toe-in to provide a quick check of whether or not the amputee needed more *medial placement* of the foot, and used toe-out for the reverse. From this you can easily deduce the significance of toe-in and toe-out on prosthetic function. However, in the extreme, the effects are opposite at heel-contact to what they are at toe-support as you can easily visualize. This sort of positioning is fixed by cosmetic requirements as a rule, and this is adequate when everything else is done well.

Summary

1. *A long prosthesis forces drop of the pelvis on the normal side, can cause crotch pressure and distal femoral pressure leading to discomfort, can reduce voluntary control, may force the amputee to walk with an abducted stump hip for stance phase and to swing the leg outward for toe clearance in swing phase or to compensate by rising up on the sound foot to clear the prosthetic foot.*
2. *A short prosthesis may shorten stride, increase listing over the prosthesis in stance phase, enhance prospects for a narrow walking base and increase the ease of balancing over the prosthesis in stance phase.*
3. *Moving the foot forward relative to the stump increases prosthetic stride length.*
4. *Moving the foot posteriorly relative to the stump does the reverse.*
5. *Displacing the foot medially has variable effects which may include walking with a narrow base, moving the foot laterally by hip abduction of the stump for relief of pain and increased voluntary control. Stump length and strength is a strong mediator in what choice the amputee makes as he tries to get comfort, save energy and maintain a cosmetic gait.*
6. *Displacing the foot laterally may lead to the amputee adducting the stump to reduce the width of the walking base, or force him to lean over the prosthesis during stance phase on it, or lift his body over the prosthesis by means of a strong lateral impulse from the sound leg as the prosthesis becomes weight-bearing with the torso erect.*
7. *Increasing downward inclination of the prosthetic toe shortens stride on the normal side.*
8. *Tilting the toe upward on the prosthetic side lengthens stride on the normal side.*
9. *Knee axis stability is increased for a larger percentage of the stance phase on the prosthesis when the prosthetic toe is inclined downward or when the prosthetic foot is moved forward with respect to the stump.*
10. *Toe-in and toe-out have comparable effects to moving the foot inward or outward except that the effects are lessened or reversed at heel contact, tending toward the effects indicated as the step advances.*
12. *Inversion-eversion changes are basically for cosmetic effects.*

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Modular prosthetics—a philosophical view*

J. FOORT

Department of Surgery, Division of Orthopaedics, University of British Columbia, Vancouver.

Introduction

This will be my definition of a *modular system:* "A system of elemental parts which can be combined in a variety of ways to arrive at the functional entity desired from among a variety of functional options which the system allows". The fewer parts required for the greatest variety of options, the more efficient is the modular system. (Imagine a system that could be adapted to both prosthetic and orthotic applications!). In prosthetics, the term modular is almost synonymous with tubular or endoskeletal structures. These designs have their origins in experimental devices such as the universities in the United States were making as far back as the 1950's. At Winnipeg, where the first system went into clinical use, the term "pylon prosthesis" was replaced with the term "modular prosthesis" when that term gained favour. The more recent designs are the systems we think of as modular now. I will accept these as representative of the modular concept, but not without discomfort. I believe other approaches would be more viable and should be entertained.

What I will do now is look at modularization from a variety of viewpoints so that users and designers may approach a little closer should they wish to improve or replace the current designs. Viewpoints considered will be those of;

1) the designer
2) the prosthetist
3) the amputee
4) the institution
5) the paying agency
6) the citizen.

1 The designer

The designer wants to make things easy for himself by designing a system which is (a) efficient in operation; (b) reliable over a long time span; (c) easy to put right when it goes wrong; (d) easy to manufacture. In order to meet such criteria he will adopt everything which is already available and tested for use in his system. For example, when the Winnipeg system was designed, the Berkeley pneumatic swing phase control unit was modified and used, the modification being related to the production requirements of the local manufacturers; the Northwestern University Hip Disarticulation Joint was included by the addition of a simple part; the wedge disc alignment method, modified to suit the system being designed, and the SACH foot were adopted. Tubing already in use in the Berkeley designs and hose clamps from the automotive industry were also adopted. The system was designed without other input. In addition the designer *must eventually* consider all of the people who will come in contact with his system, anticipating the different needs they have. However, the wants and needs are not immediately known. Experience *with* a new system precedes opinions *about* such a system. This means that a design must evolve over a period sufficiently long to allow formation of judgements and the resulting inputs. Only when a total picture has been formed can the designer finalize the system. Thus, the designer expects the system to "grow", expects feedback, and expects a certain degree of tolerance from those who commit themselves to having such new systems developed.

2 The prosthetist

What the prosthetist wants may vary according to his clients or those who influence the course of treatment and maintenance care. When I went to Winnipeg in 1963, I faced the incoming stream of amputees essentially alone,

All correspondence to be addressed to Mr. J. Foort, Department of Surgery, Division of Orthopaedics, University of British Columbia, 2740 Heather Street, Vancouver, B.C. Canada V5Z3J4.

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these were new amputees undergoing rehabilitation and entering an overloaded system. A system of prosthetic components was needed which would allow me to attach plaster sockets, temporary plastic sockets and definitive sockets; a system that would be sufficiently cosmetic to be acceptable to amputees and others who dealt with them at each stage; a system that would be included throughout the entire process from early initiation of rehabilitation to discharge and follow-up. My function as a prosthetist was to;

(a) reduce the time it took to initiate prosthetic management of the patients,
(b) accelerate the process once management was started
(c) speed up and simplify the process rehabilitation,
(d) improve the quality of prosthetic management.

I saw modularization comparable to that used in experimental devices as the answer. At the time, the only devices available were the standard Berkeley Adjustable Legs; heavy, clumsy, designed for short term and intermittent use, and unsuitable for out-patient use. The alternative needed would be readily available, fast and easy to adjust, inexpensive, functionally acceptable, durable and sufficiently cosmetic. In the beginning, the system designed was too fragile for use except within the hospital, however it was gradually improved until it was possible to make definitive prostheses using the various components for a large proportion of the amputees being served. Criticisms relating to noise, failures, damage to clothes and appearance were overcome sufficiently to allow use of the system as the major procedure in the treatment scheme. Other criteria met included fitting under clothes unobtrusively, being tamper proof, being sufficiently light-weight, being no more expensive than functionally equal alternatives and being easy to manufacture. Besides meeting these sorts of requirements, before the prosthetist is satisfied there is one more important factor. The system must be sufficiently *universal* in application before he can risk committing himself to it. New systems which face the world alone often fall over this rock. A prosthetist facing a system which he has not experienced before and for which he has no parts can be thrown into confusion when faced by it on a prosthesis from another region. He may try to talk the amputee seeking help out of it or refer him back to distant lands for maintenance. The prosthetist wants something familiar and available. That is why he wants to be introduced to a new system (whether he knows it or not!) by means of a co-ordinated effort which includes him, the designer and the educator.

If we consider systems now available and think in terms of what the prosthetist might demand of them, I would say that his main demand would be for a system which will not threaten his relationship with his client. He will not go on using devices which fail to meet such criteria as I have indicated. The system must stay mechanically and cosmetically viable for the longest period of time, be easy to use, be easy to maintain, etc. The people who will benefit from using the system in their prostheses must be sufficiently numerous to make stocking of parts and systems practical. Or alternatively, he must have quick and easy access to the parts and system. While he accepts that *the criteria for using the system for prescription purposes are defined by those people on whom the system works well,* he cannot afford to scatter his efforts among many systems which all do essentially the same thing. Considering that the bulk of the amputees passing through a service where modular prosthetics can be used are geriatrics, there are reasons for optimism with regard to modular systems. The prosthetist will accept them for the suitable cases when his job is made easier and that is something the designer and producer must take seriously.

3 The patient

The patient wants a system which is;

(a) very cosmetic
(b) very light weight
(c) never fails him
(d) noise free
(e) adjustable by himself, and which he can maintain to some extent, and why not?
(f) quick to service
(g) low cost
(h) offers a variety of functions from which to choose
(i) organized to permit him to change sockets for improving his comfort
(j) organized to stay with him through all phases of his management and follow-up care
(k) organized to introduce no hitches in the various stages of his care.

4 The institution

Modular prosthetics suits institutions, such as Rehabilitation Centres, very well. All of the patients needs can be met right in the centre. In my Manitoba experiences I found that the managers were enthusiastic for getting the patient through the process smoothly. To them, the less visible the process and the prosthetists were the better they liked it. In fact, I got the feeling that if a system could be devised which would eliminate the prosthetist some would be even better pleased. The prosthetist can hold a prominent position in rehabilitation more by facilitating passage of amputees through the system than by providing various and novel options. When the managers in Manitoba saw how easily the modular system could facilitate the process they were keen to have the prosthetic services within the institution. What they discovered was that prosthetics was a small enough part of the array of needs that are met by such a service that the modular prosthetic system was not enough. What was needed was *total modularization and standardization.* This would utterly suit institutions. Cost is not the problem so much as speed. Nothing must bog down the flow. Follow-up must be unproblematic and undemanding, otherwise the growing cadre of "treated" patients will return to retard the process. Designers must be aware of such factors and make their designs suit the needs of various institutions within which the systems will be used including home and work-place.

Managers prefer a single system which has a predictable cost and which can be made available without controversy or delay. They do not want to be confused with a variety of terms and descriptions. Designers can satisfy this need to quite an extent. I feel that designers have been remiss in not developing the various modular systems along lines which make the systems compatible in terms of interchangeability of parts and functional elements. An example of what *should* be done is illustrated by the Winnipeg designers who are making their polycentric knee compatible with the Bock system. But see how it is; various pipe sizes; various bolts, nuts, screws; different alignment methods; different attachment systems, clamps, locks. What a hodge-podge! No wonder the concept moves slowly into service. There must be greater integration, more coherence and a minimum of parts covering a maximum of situations before the institutions can embrace modularization more securely.

5 The paying agency

When the state or other agency pays for the service which might include provision of modular prostheses, what is most wanted is predictability. When users complain or costs vary the agencies become resistant. Designers must aim to overcome such difficulties through design. Nor can the agencies be expected to initiate services along new lines, they are essentially conservative, perhaps necessarily so. The demand must come from without. Only when it can be demonstrated that the agencies will *gain* from use of a system without penalty to their clients will they shift from one system to another or include an additional system. But the paying agencies have real stake *in finding out!* Therefore, they should, along with the institutions which would carry out the processes necessary, support evaluation programmes with their funds for new things, including modularized prosthetic systems.

Correctly developed, modular prosthetics will decidedly be to the advantage of the paying agencies.

6 The citizen

The citizen hardly cares. He pays willingly for an adequate service because it is an infinitesimal part of the costs of living, also, he is not unsympathetic. His views of any assistive device relate more to how it distracts him from any illusion he prefers to maintain. Thus, he will see prosthetics as a limp or a noise or a bulge, and we can say, his interest is primarily one of cosmetics. This is where modular prosthetics is falling down too. The state must give the needed support for the solution to such aspects of the problem as the manufacturers cannot solve. Designers must be supported while the problems are being solved. As far as the amputee goes, speaking from the point of view that he too is a citizen, he has a right to expect that better systems be developed with community or state funds. He has a right to regain lost function, have improved comfort and gain or remain in social dignity. To view the shortcomings in modular systems one need only think of women and children or to think of the working man in relation to modular components. At least we have seen a move

toward soft covers. Soon the Winnipeg Carver will make a restoration in a more automated way, on demand for widespread distribution, giving a product very nearly matched to the natural contours of the missing limb, possibly leading us to banks of prefabricated covers for a certain proportion of the amputee population. But other options are required. The hard user will not accept soft covers which disintegrate within weeks. Designers must pay attention to the development of two piece systems to add to the developing armamentarium.

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Modular assembly above-knee prostheses*

J. S. TAYLOR

National Centre for Training and Education in Prosthetics and Orthotics, Glasgow.

Introduction

The main aim of a modular system of assembly is to provide functional prostheses to the patient faster and more easily than has been possible up till now.

The waiting time in the past could extend to periods of months for completion of a prosthesis, when there was a heavy demand on the service.

The ideal modular lower limb assembly prosthesis, as has been agreed at the many conferences on the subject (Cosmesis and Modular Limb Prostheses, 1971; Lower Limb Modular Prostheses, 1973; Standards for Lower Limb Prostheses, 1978), would also have the following advantages:

(1) Possibility of alignment adjustments throughout the effective life of the prosthesis.
(2) Simple change of components to give flexibility in prescription.
(3) Easy disassembly of components permitting rapid replacement of worn or damaged parts.

The first requirement, that of alignment, has not been accepted by all manufacturers; two systems which come to mind—those designed by Hosmer and United States Manufacturing Company—require the alignment device to be removed after completion of dynamic alignment. Other systems such as Blatchford and Otto Bock have the alignment facility as part of the final structure. When looking at the other two advantages of the supposed ideal modular prostheses, it would seem that the systems which are designed with built-in alignment devices are at a distinct advantage.

Most of the following comments will be directed to four systems which I have been involved with over the last six years—two of which have built-in alignment systems and two which transfer-out the device.

The systems are the Blatchford or British system; Otto Bock, West Germany; Hosmer and United States Manufacturing, both designed and manufactured in the United States of America.

These systems have been the subject of an evaluation programme (Solomonidis, 1979) and, for the last three years, Blatchford and Otto Bock prostheses have been regularly supplied to patients in our clinic.

Prostheses for the above-knee amputee have always tended to be rather heavy and it was confidently expected that modularization would ease this but any gain has been minimal.

An interesting feature emerged when the weights of the completed systems were recorded; it has been argued long and often that a weight penalty would be incurred by leaving the alignment device in the finished structure, in fact the two systems which had the alignment devices as part of the structure, proved to be considerably lighter.

Blatchford	2.90 kg. to 3.65 kg.
Otto Bock	3.30 kg. to 3.50 kg.
Hosmer	3.65 kg. to 4.45 kg.
U.S. Manufacturing	3.70 kg. to 4.73 kg.

The conventional prosthesis worn by the patient taking part in the series—which included all-metal limbs—ranged from 2.5 kg. to 4.04 kg.

It would seem then, that in the present state of design, to have the alignment facility as part of the structure gives two advantages; less weight and a reduction in the manufacturing process.

Before the introduction, in the 1950's of the adjustable leg which was designed at the University of California, Berkeley, the fitting

All correspondence to be addressed to Mr. J. S. Taylor, National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, 73 Rottenrow East, Glasgow G4 0NG.

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and alignment of conventional limbs could only be achieved with considerable difficulty. The adjustable leg constituted a significant improvement on conventional fitting methods, in that it allowed the prosthetist to make easy changes in alignment, while observing the patient's gait. On completion of alignment, the adjustable portion was transferred out and replaced by a conventional knee and shank.

The introduction of modular assembly prostheses has assured alignment capability as a standard feature, whether the alignment unit is a permanent part of the structure or not, this in itself is a tremendous step forward.

While the time spent by the prosthetist on alignment has not necessarily reduced with these aids, there is no doubt that greater alignment accuracy has been made possible. Some of the alignment systems are somewhat awkward to handle, although one can adapt to them in time.

The Otto Bock arrangement which has alignment capability at two levels, below and above the knee, is particularly useful from the prosthetist's point of view—and when used in conjunction with the static alignment apparatus, makes this a very satisfactory combination. The use of the static alignment apparatus makes bench alignment so accurate that only a minimum of adjustment is required during the dynamic stage. This is perhaps something that other designers may consider with a view to reducing the size and weight of alignment couplings.

Socket shape and manufacture has not changed a great deal as a result of modularization, but the method of attachment to the knee units vary to some extent.

The Blatchford system was originally designed with metal sockets in mind hence the duralumin struts (Fig. 1), perhaps if plastic sockets had been standard, as surely they will be in the near future, this may have influenced their design. Blatchford have also designed an attachment to accommodate plastic and wood sockets but in many cases this arrangement does not allow close enough access to the knee centre. This is not necessarily wholly the fault of the prosthesis, as many above-knee stumps are fashioned without knowledge of prosthetic design and are as a result too long. Given a better shaped stump many of the so-called design faults would disappear, with benefit all round.

Fig. 1. Blatchford modular assembly prosthesis showing duralumin struts for socket attachment.

The other feature which makes modular assembly prostheses so acceptable is the availability of different knee mechanisms. Some systems are far more complete than others in this respect and have units which can cater for almost any eventuality. As far as the prosthetist is concerned, the easier the exchange of knee mechanisms during the alignment procedure, the more acceptable is the system.

The ease of change of components makes it possible to try different prescription combinations and also to effect repairs very quickly, and as even a new socket can be made and instantly attached, it lessens the need for a second or spare prosthesis in many cases. This could lead to an overall reduction in delivery times to patients, one of the original aims of modularization.

Cosmetic restoration is one area which is presenting a lot of problems to designers—whether it should be one piece or two piece, soft or firm, etc. The cosmesis should not be too difficult to work with. If it is to be modular it should not take up too much technician's time to fashion it, and the final shape should approximate to the remaining leg (Fig. 2). It has been my experience that no matter the material, shape is the most important feature, although in a percentage of amputees softness is an added bonus.

Fig. 2. Otto Bock above-knee modular prosthesis with built-in alignment unit and one piece soft cosmetic cover.

Use of a one piece cosmesis is desirable but not at the expense of knee function; this, of course, is no problem with a large percentage of the elderly above-knee amputees who would use a fixed knee and a one piece cosmesis suits their needs. However, in the present state of development the one piece soft cosmesis is not totally adequate for the young active amputee, as regards to both function and to a greater extent, durability. There is no doubt, however, that these problems will be overcome in time.

The complete system should be able to cope with all levels of amputation, not all can do this at the moment. The adult hip disarticulation patient is adequately satisfied by some systems but children, who form a fair percentage of amputees at this level, cannot be easily fitted as there are no small sizes of socket attachment.

In conclusion, during the last two years our clinic and workshop have used modular construction for all above-knee prostheses; production can be satisfied with a small number of technicians and storage of components is not a problem as sources of supply are good. In general it can be said that modular assembly makes production possible in smaller, self-contained units so long as component supplies can be readily obtained. In this connection it makes for easier handling to have a system of ordering and pricing such as is available with the British and West German modular systems.

If one could get the ideal modular system, that is one which could satisfy the needs of all patients, production and storage would be easier still and smaller production units attached to clinics would be able to function quite satisfactorily.

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Economics of modular prostheses*

A. STAROS

Veterans Administration Prosthetics Center, New York

Introduction

Some clarification is needed about the meaning of "modular" in prosthetics. To describe the endoskeletal modular prosthesis as simply "modular" is not fully correct. The crustacean (wood) *exoskeletal* array used for many years in above-knee limb construction was modular, at least in the assembly process. For this, a fitter or his technician took a knee-shank set-up, a foot-ankle set-up, and a socket and assembled these in a modular fashion. Each of the major elements was in fact a module. Nevertheless, the value of modularity was lost whenever replacement of individual components was required; it was quite difficult to replace a major component such as the knee-shank set-up in this system without re-making the whole prosthesis. The nature of the design, in wood finished with plastic laminates, prevented easy replacement.

Thus a particular significance to the modularity of current *endoskeletal* designs is associated with the quickness with which individual components might be replaced. This "plug-out, plug-in" capability simplifies servicing and maintenance of prostheses requiring either changes in function or replacement due to wear or failure.

Fitting with endoskeletal prostheses

The use of modular hardware in the endoskeletal above-knee prosthesis does not produce savings in prosthetist time; he or she is still required to cast the patient, design and conduct the fitting of the socket, and perform dynamic alignment at complexity levels no different from that used for exoskeletal systems. However, the technician, who in most facilities supports the prosthetist, will save time, especially if the modular system includes alignment devices incorporated within the prosthesis. Alignment transfers, shaping, and finishing are not necessary; assembly is quicker with no glued joints consuming waiting time. But any saving in technician time is approximately equal in value to the increment in initial cost of the modular hardware over the cost of similar wood exoskeletal components.

In the non-labour-intensive developed countries of the world, the choice between endoskeletal and exoskeletal designs offers no economic differentials, at least in the fitting and assembly processes. But, in the developing world, the procurement of the higher cost of "modular" metal hardware would be financially (and socially) unsound since use of labour-intensive methods is indicated for prosthesis production.

Repair and maintenance

Servicing the modern automobile would be an economic impossibility were it not for modularity. Local (neighbourhood) automobile servicing depends on persons who are not experts in rewinding generators or rebuilding transmissions but who can provide, through the modular approach, replacement of the defective part at that site; more specialized facilities may be involved in complex rebuilding tasks.

Thus it is primarily in the repair and maintenance that modularity may produce some economic advantages. The time of the people providing such service is saved, as is the very valuable time of the automobile user or the prosthesis wearer whose economic welfare is most likely very dependent on the quick availability of a functioning appliance.

All correspondence to be addressed to Mr. A. Staros, Director, Veterans Administration Prosthetics Center, 252 Seventh Avenue, New York, N.Y. 1001, U.S.A.

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More complicated prosthetic systems may be expected in the future. In these there will be an increased need for special "fitting" considerations, particularly in the selection of the proper components for the prosthesis. An economic value from modularity during the initial fitting and assembly processes may then be realized but, with fitting practices today, the economic advantage in the use of modular systems comes solely in repair and maintenance.

Advantages of interchangeability

A very significant advantage can nevertheless be experienced in the prescription process today with one particular endoskeletal design; the multiplex system designed by the U.S. Manufacturing Company in response to a U.S. Veterans Administration standard. This unit permits an interchange, in the same metal shank, among different kinds of fluid knee controls as well as a special mechanical friction system. The dimensions (and angles) of the triangle in the antero-posterior plane formed by the knee axis, the top of the piston rod and the lower attachment of the cylinder have been standardized (Fig. 1). Manufacturers of the various units co-operated with the Veterans Administration in making their knee controls conform with this standard geometry.

This design permits quick interchange among these systems as prescription is contemplated. Since there are variations in the functions among the fluid controlled units, a trial with actual performance analysis can be quickly organized.

This system also typifies the advantage which will be evidenced when a fluid control malfunctions; replacement is very easy. The "plug-out, plug-in" possibilities when selecting rotators for prostheses should also not be overlooked; future designs will consider this.

On a subject not directly related to the main thrust of these comments, we have successfully experimented with the use of graphite fibre-epoxy composites to achieve weight reductions in certain prosthetic components. For example, the present aluminium multiplex frame weighs approximately 700 g. A graphite-epoxy frame weighs about 450 g. (Fig. 2). Models of these units are now being tried on patients.

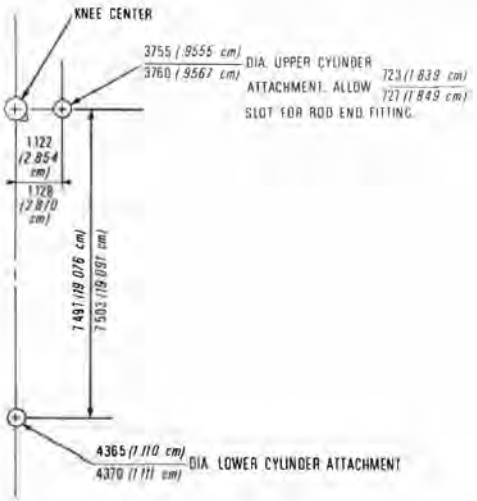

Fig. 1. Dimensions for fluid knee controls.

Fig. 2. Graphite composite multiplex pylon.

Fig. 3. Graphite composite rotator.

Graphite-epoxy is also being used in rotators (Fig. 3) and in the SACH foot keel; by this latter application the total foot weight can be reduced by about 20%, a significant functional gain at the end of the long lever that is the lower-limb prosthesis.

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Lightweight prostheses*

A. B. WILSON, Jr.

Rehabilitation Engineering Center, Moss Rehabilitation Hospital, Philadelphia.

Because the research programme in artificial limbs in the United States was initiated primarily for the benefit of young amputees from World War II and because amputations due to vascular disease were not as common as is now the case, very little attention was given to the effects of weight of the prosthesis for the first years of the programme. This is not to say that weight was not considered at all. It was always recognized that artificial legs could be too heavy even at weights considerably less than the lost limbs, and efforts were made to keep the weight reasonable. There were also those who felt that artificial legs could be too light which probably contributed to the lack of interest in weight reduction.

The fact that during the course of evaluation of the Henschke-Mauch devices (Lewis and Bernstock, 1968) amputees reported feeling that the heavier new limbs felt the same or lighter, no doubt owing to better fitting, alignment, and function, added no incentive to conduct studies concerning reduction in weight of lower-limb prostheses.

However, about 1960, the so-called geriatric amputee problem began to be recognized and the report of nearly every workshop and conference since that time has included a recommendation that lighter limbs are needed for the older weaker patients. But until recently, very little attention has been given to this matter.

The reasons for this are not apparent. However, experience with Dr. J. Barredo (1975), a BK amputee himself and a retired physicist, the work of Wollstein (1972) (Fig. 1), and the introduction of vacuum-formed polypropylene lower-limb orthoses prompted experimentation at the Moss Rehabilitation Center in 1975 that has produced an extremely light below-knee prosthesis (Wilson et al, 1976).

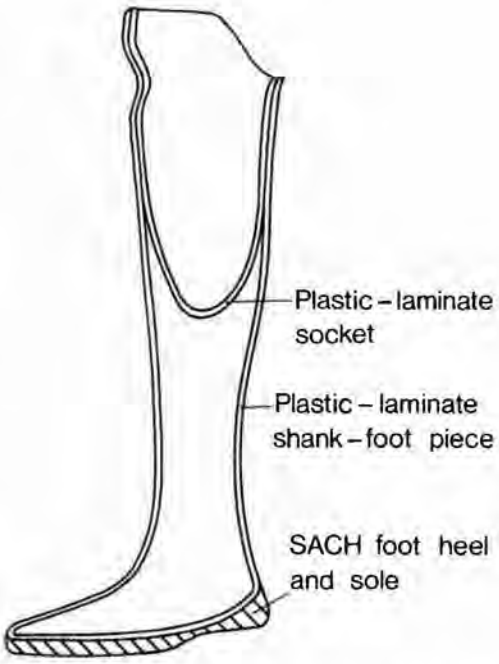

Fig. 1. Cross section of the Wollstein below-knee prosthesis.

The crustacean design was selected because it is inherently lighter for the same structural strength than the so-called pylon construction. The result is a below-knee prosthesis that weighs two-thirds less than the standard PTB with the same or better function since suspension becomes less of a problem (Fig. 2).

All correspondence to be addressed to Mr. A. Bennet Wilson Jr., Director, Rehabilitation Engineering Center, Moss Rehabilitation Hospital, 12th Street and Tabor Road, Philadelphia P.A. 19141, U.S.A.

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Fig. 2. Ultralight below-knee prosthesis on patient. Note that suspension is by flexible supracondylar tabs and that auxiliary suspension is not needed.

Although the design was developed primarily for the geriatric amputee the experimental units were tried on a 22 year old, very energetic male with a below-knee amputation due to a motorcycle accident. This patient was selected because his work schedule made him available during the day and because it was felt that he could report his findings more objectively than most of the old patients that were available.

We were pleasantly surprised to find that not only the initial patient but other young patients appreciated the substantial reduction in weight.

At the present time the Rehabilitation Engineering Center at the Moss Rehabilitation Hospital with the co-operation of six local private prosthetics facilities, the Veterans Administration Regional Amputee Clinic, and several private clinics, is in the last stages of a clinical study involving approximately 36 patients nearly all of whom had previously been satisfied with the conventional PTB prosthesis. There seems to be no doubt that the reduction in weight is greatly appreciated. An unexpected result is the favourable reaction of the patients to the "feel" of the polypropylene sockets.

Our next project will be to study the effects of weight and weight distribution in above-knee legs. The first step will be the development of a limited function leg as light as can be devised. We hope that the immediate result will be a design that will be useful to the geriatric amputee. The second phase will include the introduction of more functional knee-control devices and measuring the biomechanical results as weight is increased in different areas. This approach has been proposed before, but it now becomes feasible to provide at a reasonable cost extremely light legs to which weight can be added gradually to study the effects of weight change, since here-to-fore it has been very costly to provide such experimental devices. Furthermore, with the advent of our force-line visualization equipment (Wilson et al, 1979) and our proposed accelerometer configuration to provide quickly an index of efficiency (Moss Rehabilitation Hospital, 1979) the likelihood of success is increased greatly.

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The establishment of prosthetic services in African countries*

S. HEIM

German Agency for Technical Co-operation Ltd. (GTZ), Eschborn

I have now worked for more than 10 years in several African countries and think it is not necessary to cover the problems which Eric Jensen has already pointed out, because mainly the same social problems are encountered in African countries as in South America.

I should like to focus more on the possibilities of doing something positive in those countries, perhaps with a view to setting up orthopaedic training centres, orthopaedic workshops etc.

There is one thing that should be mentioned first, in most African countries the medical and technical services in our field are State Services. These services are being built up and to an ever increasing extent things are being provided by the government of the different states. So in Africa there is perhaps a better situation in the initial phase than Mr. Jensen had in South America.

In 1973 we handed over to a Tunisian director the orthopaedic workshop in Tunis. Five years later I had, for the first time, the occasion to visit this former Technical Co-operation project. Over a three-week period I was able to examine and judge thoroughly the work done by the Centre. There is now a limb fitting and brace centre employing 7 prosthetist/orthotists and about 40 orthopaedic technicians with a relatively high technical level.

The workshop, situated in a really modern building, has a branch facility about 150 km away and is setting up another one in the southern part of Tunisia.

As far as I know no other African country has such an ideal symbiosis between orthopaedic hospital, surgery, physiotherapy and technical service. In principle no difficulty should exist today in providing orthopaedic appliances to any patient in Tunisia—to rehabilitate and reintegrate him into the working process and into society. And yet it was realised based on critical observation, that the achieved results do not correspond one hundred percent to the anticipated ones. It was hoped that the staff—who were professionally very well trained—would create their own ideas, would enlarge and expand the social service and would to a greater degree adapt the technical needs to the conditions of the country.

In fact this has not been the case. It was noticed that provision of orthopaedic appliances was in some cases extremely good, that a fairly large group has been treated regularly and with good results and that the external view of the whole facility was very positive. However, further development has not taken place during the last 5 years. The technical staff has only preserved what was handed over to them by the foreign experts five years ago.

Up to the present, there are only small signs that the received technical knowledge is being adapted to the needs of the country. Earlier methods are still being copied. Today fewer patients from rural areas are being really properly cared for. Putting the purely technical service aside, the social service in this Centre does not meet the expected level. On the contrary, the work has become worse compared to former times. The present social worker is only doing administrative work, no decisions are taken about any social problem cases and no attempts are made to speak to and advise patients, to make home visits and eventually to begin the total rehabilitation process.

All correspondence to be addressed to Mr. S Heim, German Agency for Technical Co-operation Ltd. (GTZ), Postfach 5180, D-6236, Eschborn 1, Germany.

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This example from a period covering nearly 10 years suggests that we reconsider our attitude with regard to the follow-up care of such projects. It is just impossible to set up something completely new such as an orthopaedic service in a short period and to then leave it alone.

Even so, as far as I am aware Tunisia is the country with the widest-developed orthopaedic care services in Africa. In most African countries there is no normal orthopaedic care service, that means no continued service for manufacturing and repairing prosthetic appliances. By this I do not only mean a more or less sufficiently large facility in the capital of the country, but a service able to serve every district and every population group. In most capitals limb fitting centres etc. can be found. These attempts, as I would like to call them, correspond in no way with the demands of the population, with the technical and medical responsibility and above all with the social duties. In most cases only a few possibilities of providing orthopaedic appliances are considered. Sometimes, only one or two types of prosthesis are available and are given to all patients if possible. The existing professionals, having been trained only to provide those types of prosthesis, abandon the work if a slight deviation is made or if problems crop up. That means that only a small group of patients can be properly cared for.

From the statistical point of view these centres are not bad—because the total number of patients is so high that even though they select only a particular type of patient they are fully employed. But, compared with the population and the different types of orthopaedic patient, they are by no means what they could and should be.

To solve these problems, the demands made on professionals—the prosthetists—in African countries are much higher than perhaps those made on equivalent technical personnel in the developed countries. A technician in a normal European Centre has less problems with the medical part of the work. Very often, the African prosthetist has not only to take care of his own work, but partially to take over the work of the rest of the rehabilitation team.

Another wide field not sufficiently taken into consideration is the social aspect of prosthetic care. In orthopaedic centres I rarely met a functional social work department with social workers caring for the disabled. Even when social workers were present, they only had the functions of a postman, they carried files from one place to another. There was hardly any proper social work, such as giving advice to patients, discussing financial problems, co-ordinating with other departments and so on.

This is a very sad outcome, because nearly every African country has a social ministry.

The orthopaedic technician has very often to take over this work and stands in the place of the social worker to tackle these immense social problems.

Nevertheless, it should be pointed out that the financial problems regarding costs are not the most important, because in the countries concerned it is often possible to overcome those problems.

Taking into consideration all these facts, there must first be assured an all round training of specialists in prosthetics in order to set up a good service. This does not mean a short term or narrow-gauged training that cannot give the desired basis. Only the well and fully trained professional has a chance in Africa to fill the gap and, last but not least, to find an appropriate salary and rank in government administration.

From the above mentioned training requirements it is evident that the planning and operation periods of such technical projects have so far been too short. It has been equally proved that it is not possible to set up a prosthetic service if the foreign expert and adviser cannot stay and work on the spot over a longer period of years, so as to show the lines along which work has to be continued.

There is no use in setting up theories—even well presented—if the African whether he be an official, the patient—or maybe the future expert, is not able to see clearly the technical potentials.

It is known that 90 per cent of the African population live in rural areas, and this must be taken into consideration when planning future services. This requires, besides the technical work, arrangements for effective social services. In the rural areas, more attention is to be paid than in the capital cities to not only the financial problems, the climate and way of life, but also to the level of education and language and the religious customs and transportation problems of these patients.

A good social network is needed over the country which is able to give, together with the technicians, education and advice to handi-

capped people; so that, for example, prostheses could be repaired by local handicraft workers and patients far away from the centre would be in the position to help themselves.

Only continuous care, education and advice covering patients, chiefs, politicians and the whole population, will guarantee the process of orthopaedic care in rural areas.

So far, a very important member of the team, the specialized MD has only been mentioned in passing. That does not mean that this is not a problem in Africa. As already mentioned, the prosthetist has to take over responsibilities in medical areas which exceed the normal level. The problem of the MD in developing countries is far reaching and I think we are very far from solving those problems. When creating a prosthetic service in such countries all the available medical potential should be included to see how further training can be given to them in order to win them over for constructive co-operation.

Permit me finally to portray some ideas regarding the level of orthopaedic care. I have often tried to propose very simple types of prosthesis for patients living in rural areas or mountain regions, to avoid difficulties in maintenance.

With these proposals I had the peg-leg in mind, which I find is sometimes indicated. Not even one patient during my 12 years in Africa has accepted such a leg. Even if he was very poor, he refused to accept this kind of prosthesis as he felt he was being discriminated against. He liked to be treated like every other patient. People in Africa are generally very sensitive if they feel they are being treated as second class individuals. This also applies to the field of orthopaedic treatment. They carefully evaluate the situation, can be convinced by understandable examples but refuse anything that can be considered second rate.

May I give you another example of the difficulties we have to deal with.

In Togo there are a lot of leprosy patients. The mutilated fingers do not permit the use of crutches; these people are no longer full, active members of the community. It was the intention to provide some of those patients with PTB prostheses. They all accepted the idea, but nobody wanted to be the first i.e. to be an experiment. After long discussions one of them was amputated and fitted with his artificial limb. He was very carefully observed by his co-citizens—only after his return to the tribe and after proving that his PTB was usable did the other patients come into the hospital.

Unfortunately, we had to give one of this group a conventional BK prosthesis with a thigh corset. New discussions arose, he wanted to have a leg like all the others. There is no ethnic or religious resistance to be noticed against certain kinds of orthopaedic treatments. It is only the intense desire of African patients to be treated equally and not to be discriminated against. In Africa I have never met groups refusing prostheses for any religious reason. I have drawn this picture to underline that it is not enough to set up an orthopaedic service in a developing country and then leave it alone.

Perhaps I should give you another example.

I recently visited a certain country and had discussions with the Under-Secretary of State. He talked to me about their orthopaedic care services, hospitals and so on. We talked about difficulties, what should and what could be changed by eventual assistance etc. I asked him suddenly if he would send his own children to the orthopaedic workshop for treatment. I had no idea of his family situation and was unaware that he had a child suffering from polio. He was astonished and answered, 'You see, I have to send my child to London. The local services are not good enough to provide the service I want'.

Thus, you see, that is the problem. The level of those centres is not high enough to encourage all sections of the population to go there. Our duty is to treat everybody in the local centres, from the highly educated to the rural population.

Conclusion

After having set up orthopaedic care centres in developing countries following the rules pointed out, they should be given the possibility of continuing the process of development by themselves by supporting them with continued follow-up care by means of international professional groups or bilateral agreements.

This is, in my opinion, one of ISPO's duties in the future.

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Prosthetics and orthotics in Latin America*

E. JENSEN

Pan American Health Organization, Lima

Latin America can be divided into four parts. Bordering with the United States is Mexico, next there is Central America then the Northern and Southern parts of South America. Europeans tend to be concentrated in the South where there are many Germans, British, French, Italians and Danes. The further North you travel the more you meet the Indians and their culture and the more you encounter different attitudes.

Throughout Latin America there are three different types of services for the amputee: private shops or laboratories, state supported laboratories and government controlled laboratories. Patients can also be divided into three groups:

1. Patients who are financially independent; these usually obtain services from private prosthetic/orthotic laboratories or travel abroad for treatment.
2. A very large group of middle or working class patients who are covered by insurance or social security as well as receiving support from their families; they are normally sent to a state supported laboratory, such as my institution, for service. This type of assistance is provided by the majority of the Latin American countries. (In some hospitals or rehabilitation centres where social security services are provided, doctors—mostly physiatrists, issue prescriptions for prostheses or orthoses and send their patients to places from a list of private or government laboratories. The final checkout is done in the institution where the prescription is issued, usually without the participation of the prosthetist.)
3. The great majority of patients are in the low income group and they generally use the services of the government prosthetic/orthotic laboratories at the Rehabilitation Centres.

Many of the patients in the third group fail to take good care of their stumps, due to lack of training, following discharge from hospital. When they come to the Rehabilitation Centre their stumps are often in very bad condition with contractions, heavy subcutaneous tissue, neuromas etc. Consequently, many patients have to start protracted pre-prosthetic treatment or undergo revision surgery. However, the indigent patient cannot afford to pay for a long course of treatment as his income is very low and, because there is no birth control, his family may be large. As a result the patient may insist that the prosthesis be finished quickly so that he can return home and resume supporting his family. Many of these patients will return to the clinic due to changes in their stump. They will complain that the prosthesis was improperly fitted, and insist on getting a new prosthesis. This situation is very common in most of the Latin American countries and results in much wasted time and material. It would be very useful if prosthetic clinics could be attached to the orthopaedic hospitals, but this will be difficult to arrange. It would also be very helpful if more of the new amputees could be fitted with rigid dressings. The use of a temporary pylon prosthesis would also greatly benefit the patient and I am very happy to note the work that is being carried out elsewhere on temporary sockets for the primary amputee. We are particularly interested in the lightweight polypropylene prosthesis. Many Latin American countries are producing this material and the new prosthesis may partially answer our problems of low budgets and difficult

All correspondence to be addressed to Mr. E. Jensen, OPS/OMS Los Cedros 269, Casilla 2117, Lima, Peru.

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importation of sophisticated materials. At the moment nearly eighty percent of all amputations are carried out at above-knee level and only a very small number are below-knee, we would like to see that changed.

In some cases patients are unable to pay for the prosthesis. Many patients from rural areas have to travel a long way to the specialized services, which are usually located in the capital cities, and do not have enough money to pay for food and accommodation while being treated. When his funds run out, the patient may have to abandon treatment and return home perhaps without his prosthesis.

In order to solve these problems the governments have assigned an annual budget to the Rehabilitation Centres, including the prosthetic/orthotic departments. However, due to scarcity and failure to provide the centres with regular supplies of materials and the fact that the prosthetic/orthotic services do not have an independent renewable budget, it is still difficult for many patients to get a prosthesis made. We are trying to have a separate budget for the prosthetic/orthotic laboratories so that we can get a little closer to a good prosthetic service. Another problem faced by patients is the difficulty of finding a job; it is easier for some amputees to inspire compassion by showing their stumps or crippled limbs and begging for support.

In spite of the incorporation of the prosthetic/orthotic services in the Rehabilitation Centres, technical difficulties arise because these services are managed by physiatrists who are not skilled in amputee care. The surgery itself is carried out by orthopaedic surgeons. Another problem which requires attention is that the prosthetist/orthotists are not recognised as professionals by the medical staff and their opinions with regard to prescription formulation and prosthetic management are not accepted.

In my twenty years experience in training prosthetists in Latin America I have observed that they are highly qualified and have great interest in their profession. Unfortunately the administrative authorities are still reluctant to recognize them as professionals despite the fact that two permanent schools exist for paramedical staff in rehabilitation; both of which are recognized by the Ministry of Health and the Ministry of Education.

The problems of the prosthetists start when they graduate. The Departments of Personnel and Administration usually classify prosthetists as technicians because they work in a workshop. This unhappy situation results because, although the governments spend large sums training prosthetists, they fail to classify them accordingly; in addition, the salaries paid are inadequate. Consequently many prosthetists change their job in favour of a better position and the prosthetic/orthotic services are undermanned. In some Latin American countries only limited training is provided to meet the most urgent needs of the service. In the rural areas and other places with poor access to services use is made of simple devices made in the community from local materials.

It is worth mentioning that despite the difficulties and obstacles good results have been obtained in certain aspects of the service. This progress is related to improvement of the teaching system which began with the training of resident rehabilitation doctors in prosthetics and orthotics and the introduction of simple, inexpensive methods, using local material to avoid expensive importation of materials and components.

New ways of solving the problems outlined are being applied. A prosthetic/orthotic educational programme has been organized through the Pan American Health Organization for doctors dealing with cardiovascular disease, orthopaedics and rehabilitation. The programme is concerned with the importance of preserving articulations, especially the knee joint; the use of immediate rigid dressings, early prosthetic fitting, stump shape, prescription and information on developments related to new materials and designs. The School of Prosthetics/Orthotics of the New York University Post Graduate Medical School has co-operated with PAHO, WHO and the Member Governments in programming and implementing these courses.

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Ideas on sensory feedback in hand prostheses

P. HERBERTS and L. KÖRNER

Department of Orthopaedic Surgery, University of Göteborg, Sweden

Abstract

Development of systems for sensory feedback in hand prostheses has not been as successful as that of modern prosthesis control systems. The discrepancy is partly caused by an insufficient analysis of the concept of sensory feedback and by neglect of knowledge on the physiology of kinesthesis. In the present paper modern theories on physiologic kinesthesis are briefly summarized and the implication of these theories on the development of prosthesis sensory feedback systems are discussed. It is concluded that the future development of sensory feedback systems for hand prostheses should be directed towards increased utilization of the physiologic kinesthesis resulting from operation of the prosthesis control systems. This can be obtained by further development of the control systems. One promising approach in this direction is the use of a proportional control signal based on signal acquisition through pattern recognition of multiple myoelectric signals. Development of artificial systems for feedback should be restricted to situations when feedback emerging from the prosthesis control is insufficient. The importance of simplicity and reliability of feedback systems is stressed as well as the necessity to maintain prosthesis self-containment even after application of a feedback system.

Introduction

The development of control systems for motorized prostheses based on detection of myoelectric signals has been rapid and successful following the first use of such systems in the late nineteen fifties (Battye el al., 1955). Clinical success with powered prostheses has been reported (Schmidl, 1973; Lewis et al., 1975; Herberts et al., 1978), but other authors have expressed doubts about the value of these devices (Mooney, 1976). Beyond doubt, many patients using myoelectrically controlled prostheses have been pleased with them, despite several limitations in the prosthetic function (Herberts et al., 1979). A generally recognized drawback in powered prostheses in comparison to conventional cable-operated devices is that the lack of feedback makes control outside the field of vision very difficult. In addition, the control of strength in the grip is insufficient. These findings have led to attempts to develop artificial feedback systems for use in myoelectric prostheses (Kato, 1970; Mann, 1973; Clippinger et al., 1974; Prior & Lyman, 1975; Rohland, 1975; Shannon, 1979). Most feedback systems described up to now have not reached a development stage allowing routine clinical use outside the laboratory. Clinical follow-up has been reported in only small series. Measurements of the performance of amputees using powered prostheses with feedback show that it is close to their performance with conventional cable-operated grips (Mann & Reimers, 1970). The purpose of this paper is to discuss some basic clinical and physiological principles relevant to feedback in externally energized prostheses. Possible fields for future research in this area are indicated.

The concept of sensory feedback

A hand prosthesis is mainly a machine to replace the prehension of the lost hand. The loss of a hand, however, also implies the loss of an important sensory function. The full sensation of a normal hand, as pointed out by Moberg (1964), is a very complex quality which does not lend itself easily to replacement by artificial devices. Sensory feedback is not intended to replace normal hand sensation. We think that sensory

All correspondence to be addressed to Lars Körner, M.D., Department of Orthopaedic Surgery, Ostra sjukhuset S–416, 85 Göteborg, Sweden.

feedback in prosthetics should have the same meaning as feedback in control engineering; i.e. a way to compare the output of a machine with the input (Fig. 1). Feedback is an aid to increase the accuracy of the control system and of prehension, Therefore, the term "artificial touch" is inadequate and may lead development of prosthetic feedback systems in wrong directions.

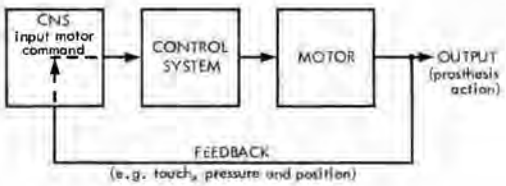

Fig. 1. Concept of feedback as discussed in the text.

In order to effectively enhance control of the prosthesis the sensory feedback should consist of several components. One component is kinesthetic information about position, movement and force in the prosthesis joints, i.e. proprioceptive feedback. Another component is information about the effects of the action of the prosthesis on the outside world; i.e. somatosensory feedback concerned with, for example, force in the grip and detection of slippage of handled objects. Most feedback systems developed today convey somatosensory information only. The somatosensory part of feedback information can not replace proprioceptive information and vice versa. The special importance of feedback concerning handling of objects in the grip has been pointed out by Forchheimer et al., (1978).

Existing feedback systems

Feedback systems for externally powered prostheses have been used since the early nineteen sixties (Tomovic & Boni, 1962). Most systems have employed some kind of artificial stimulation controlled by a transducer situated in the grip of the prosthesis. The block diagram of Figure 2 illustrates some different solutions of the feedback problem as described in the literature. The most commonly used artificial feedback systems have worked with vibratory stimuli (Alles, 1970; Mann, 1973) and electrical stimulation (Kato, 1970; Clippinger et al., 1974; Reswick et al., 1975; Rohland, 1975; Anani et

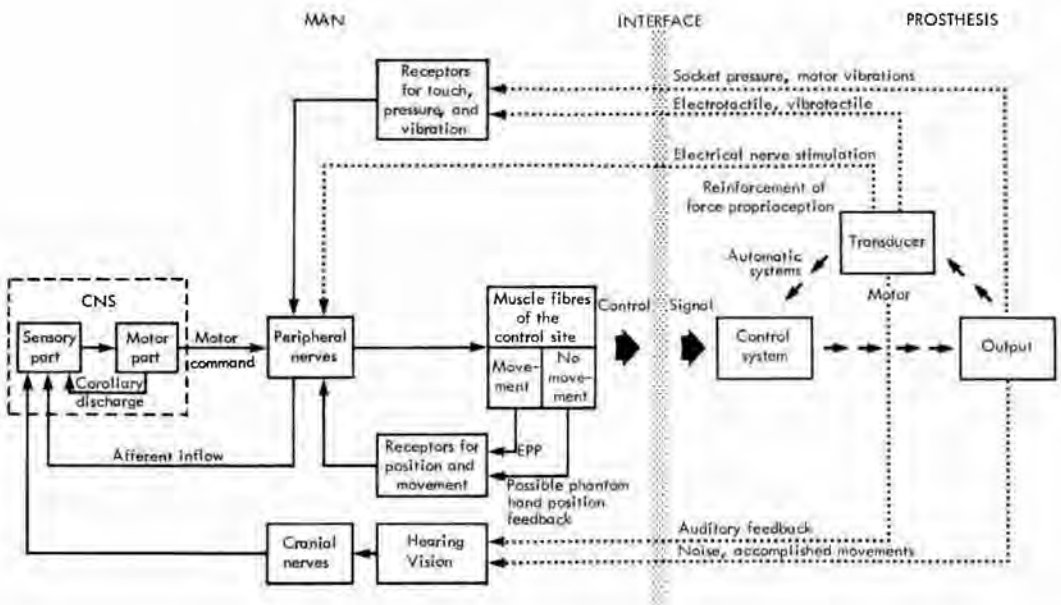

Fig. 2. Different means of providing feedback from externally powered prostheses.

al., 1977; Shannon, 1979). The clinical application of artificial sensory feedback systems has often been unsuccessful (Reswick et al., 1975; Mooney, 1976; Reswick & Nickel, 1977). Rejection of systems has been caused by technical problems such as fragility, interference with the control systems and lack of miniaturization. It is obvious that in some applications the need for a feedback system has not been present, which of course, has led to rejection.

Kinesthetic mechanisms and prostheses feedback

Aids for the handicapped designed to take advantage of physiologic mechanisms will have a greater chance of being accepted by patients than devices based on entirely artificial grounds (Herman, 1973; Hirsch & Klasson, 1974). This is especially true of the complicated myoelectric prostheses that include a feedback system. In the design of such devices thorough knowledge of kinesthetic mechanisms in man is mandatory for success. Mann (1973) and Simpson (1974) described feedback for prostheses using the physiologic signals that result from the actions of the human body necessary to control the prostheses. Clinically this has resulted in the most attractive and successful systems so far described. Practical experience with such devices shows the need to take physiologic kinesthetic mechanisms more into consideration in the future development of feedback systems.

Physiologic kinesthesia and forearm amputation

The knowledge of neural mechanisms behind the sensing of muscular effort and the sensing of position has been revised during the nineteen seventies. Previously, it was widely believed that the sense of position was based entirely upon information from joint and skin afferents (Rose & Mountcastle, 1959). Indirect evidence is now against this view. Investigation of cat knee joint receptors has not revealed the presence of any receptors capable of signalling absolute joint angles (Clark & Burgess, 1975). Sense of position is not affected by total joint replacement (Grigg et al., 1973). In addition, it was shown by Goodwin et al., (1972) that artificial stimulation of muscle spindles induces illusions of movements. This indirect evidence opposing the traditional views on kinesthesia led to a re-evaluation of the classical experiments concerning the sensing of movement and position when finger movements were performed with blocked skin and joint sensors. Through such experiments it has been convincingly shown by Gandevia & McCloskey (1977) that the sense of position has its neural mechanisms partly in muscle spindles. By performing weight-matching tests, McCloskey & Gandevia (1978) have shown that the estimation of heaviness and the sense of effort have their neural mechanisms largely in the central nervous system.

Substantial parts of the tendons and muscles executing movements of the normal hand and fingers are left intact after a hand amputation. The new evidence presented above leaves no doubt that these remaining structures contain receptors responsible for important components of the sensing of position and movement in the normal hand. The information from these receptors is also available to the amputee as is the sensing of effort.

These physiologic facts should be taken into consideration in the design of prosthesis feedback systems.

Sensing force and effort

The force proprioception system developed by Mann (1974) for the Boston arm works with the sensing of effort accompanying the muscle work at the control site necessary for generating myoelectric signals. A negative feedback signal proportional to the force resisting the movement is added to the myoelectric signal that controls the prosthesis. When the prosthesis is loaded an increased myoelectric signal is required to achieve movement. The result is an augmentation of the sensing of effort of the muscles at the control site. Such an augmentation is necessary to make the force clearly perceivable. Since the sensing of effort has its neural mechanisms in the central nervous system, it can be assumed to be influenced by signals originating from different muscles relevant to one specific movement (Herman, 1973). Such a convergence can explain why previous attempts to utilize the sensing of effort for prosthesis feedback purposes without strong augmentation have been disappointing. In these attempts subjective sensations resulting from activity in single muscles only have been applied. Single muscle activity is an unphysiologic phenomenon and probably lacks cortical representation (Radonjic & Long, 1970). Therefore, if the control of prostheses can be related to physiologic movements rather than

to actions in single muscles, more feedback information can be expected to emerge from the control system.

Hand prostheses controlled by pattern recognition of multiple myoelectric signals have been described by, among others, Herberts et al., (1973). The pattern recognition approach permits control of the prosthesis through and integration of signals from several muscles relevant to one specific movement. It is therefore plausible to assume that a proportional control signal derived from signal processing according to the pattern recognition method should yield significant amounts of information about effort and force. Our preliminary data support this hypothesis.

Sense of position

Muscle afferents have been shown to play a significant part in the perception of movements (Mattews, 1977). In a clinical follow-up study of unilateral below-elbow amputees using myoelectric prostheses we found that all non-congenital amputees (35 patients) had a distinct phantom image (Table I). All the patients except two stated that they could easily move the perceived phantom and that they could feel how the phantom was moved. The neural basis for the perception of movement of the phantom image is considered to lie in the muscle afferents from the distal part of the amputation stump (Henderson & Smyth, 1948). Control of prostheses using the pattern recognition method is based on the principle that specified movements of the phantom hand shall result in corresponding movements of the prosthesis. Our preliminary data show that it is to a certain extent possible to relate a proportional control signal achieved through the pattern recognition method to specified positions of the phantom hand. Therefore it seems reasonable to postulate that the sense of movement and even the sense of position of the phantom hand can be used to convey prosthesis feedback. Further development of the prosthesis control systems for feedback purposes is, however, necessary if this goal is to be achieved.

Table I
Features of phantom perceptions in 35 unilateral, non-congenital below-elbow amputees.

	No. of patients
Presence of phantom perception	
Constantly	26
Can be evoked	9
Ability to perform distinct movements with phantom hand	
Yes	33
No	2
Length of phantom extremity	
Equal to non-amputated side	16
Shorter than non-amputated side	13
Can not tell	6
Extent of phantom extremity	
Complete hand	21
Parts of hand	4
Variations in extension	8
Unknown	2
Pain in phantom	
None	24
Mild, occasional	6
Severe	5

Extended physiological proprioception

A different way to utilize the physiologic actions at the prosthesis control site was described by Simpson (1974). In prostheses designed for amelic children the movements and the positions of the clavicle are translated to movements and positions of the prosthesis. Through this proportionality between the movements of the clavicle and the prosthesis the angle of the prosthesis in space can be determined by the normal kinesthetic mechanisms of the child. The brain will then very easily adapt to the length of the terminal segment (the prosthesis) in much the same way as the golfer will adapt to the length of his club. The phenomenon is called extended physiological proprioception (EPP) and gives the amputees significant amounts of feedback information in addition to excellent efferent control.

Areas for future research

Future approaches to feedback in hand prostheses should be directed towards the design of systems which can be included in self-contained prostheses. The importance of self-containment for patient acceptance is clearly documented (Childress, 1973). The experiences of Mann and Simpson indicate that feedback systems based on the physiologic actions necessary to control the prosthesis are consistent with high patient acceptance and with prosthesis self-containment. The new evidence on neural mechanisms underlying kinesthesia gives further support to the opinion that development of control systems providing feedback information

as well will lead to success. One promising approach in this line of research is the use of pattern recognition of multiple myoelectric signals to generate a proportional control signal.

In accordance with the discussion above, a proportional control signal derived from several muscles with relevance to one specific movement should yield substantial amounts of force proprioception in addition to sensing movement and possibly also sensing position.

Even if important feedback information can be achieved from a properly designed control system, it is necessary to leave room for systems working with purely artificial stimulation as well. Such stimulation is needed to convey the somatosensory components of kinesthesis. In the design of purely artificial sensory systems it is equally necessary not to endanger self-containment and reliability of the whole prosthesis system. From this point of view electrical stimulation seems to have advantages over mechanical or auditory stimulation. Electrical stimulators can readily be miniaturized, their energy consumption is low, and they are reliable. If the electrical stimulation is applied to the nerves of the amputation stump the sensations will be felt by the amputee in specific parts of the phantom image (Anani et al., 1979). This creates a convergence of feedback and control functions to the phantom image. Such convergence can be expected to increase the accuracy of prosthesis control (Weissenberger & Sherridan, 1962).

Acceptance by the patient is the only important criterion to determine if an effort to replace a lost function with an artificial device is successful. Patient acceptance is a complicated concept which is not determined only by technical and cosmetic characteristics of the rehabilitation aids. Social and economic factors are equally important in addition to the psychologic attitude of the amputee towards his handicap (Höök, 1976). This means that complicated rehabilitation aids can never be prescribed without a thorough analysis of the psychologic and socio-economic situation of each individual patient. However, in order to meet the requirements of most patients, rehabilitation aids must be reliable, easy to use, and inconspicuous. Therefore, the aim in designing prosthesis feedback systems must be as much to maintain prosthesis self-containment and self-suspension as to provide significant amounts of feedback information.

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Gait training for the below-knee amputee*

C. VAN GRIETHUYSEN

Bioengineering Unit, University of Strathclyde, Glasgow.

Introduction

Statistics show that the real incidence of amputation occurs between 60 and 75 years (Kay & Newman, 1975). Hence most of the time the therapist is dealing with older patients who have a poor vascular condition which is often accompanied by other complications such as diabetes, hypertension or cerebro-vascular accidents. The level of amputation is highly dependable on tissue viability (Murdoch 1977). In the older amputee, preservation of the knee joint makes a great difference to his rehabilitation prospects and his ability to walk greater distances without expending even more energy. The intensity of treatment has to be carefully adjusted to each individual patient.

Pre-operative treatment

In this phase attention is directed to muscle reinforcement and the maintenance of full joint range movement of all extremities; the patient is taught to balance on one foot and how to transfer from bed to chair and vice versa, as far as his condition allows.

Post-operative care

As soon as the condition of the patient permits bed exercises are started for muscle strengthening; special attention is paid to preventing contractive deformities, especially of the preserved knee joint. Exercises are carefully and gradually increased in intensity.

If a rigid dressing has not been applied the patient is taught to touch and exert slight pressure with his hands on his stump. This not only prepares him to tolerate his prosthesis but also teaches him to accept his stump as part of his body on which he has to rely for walking.

All correspondence to be addressed to Ms. Corla van Griethuysen, Bioengineering Unit, Wolfson Centre, University of Strathclyde, 106 Rottenrow, Glasgow G4 0NW.

Gait training

As soon as possible, standing exercises in the parallel bars are started, with or without the prosthesis.

Once the patient has been fitted he learns to flex and extend his knee to work the stump comfortably in the socket. He is taught to bring his weight over the prosthesis and gradually reduce the support provided by the parallel bars.

It is very important that the patient learns to bring his full weight over the prosthesis before he starts walking and learns to have confidence in his new leg.

It is helpful to do these exercises in front of a mirror so that the patient can correct his posture with his weight evenly distributed over both legs.

Once the patient can accomplish the balance exercises properly, with only slight support from the parallel bars, walking training can be started between the bars which, at this stage, are only for security and to give the patient more confidence.

Initial walking exercises

Firstly, the patient is taught to raise each knee alternately and to try to take weight on his limb as much as possible. Secondly, with the prosthetic foot five to ten centimetres ahead of the sound foot, he is asked to flex his knee with progressively more weight on it until the prosthetic foot is flat on the floor (Fig. 1, left). He is then asked to do the reverse and extend the knee fully, at the same time bringing the weight back to the sound leg. This teaches the patient that he can use and trust the knee of his limb. Thirdly, he brings the prosthetic foot forward again as described previously, then steps forward with the normal foot at the same time extending the knee on the amputated side while keeping the foot flat on the floor. This is repeated with

*Based on a paper presented at the 7th Scientific Meeting of the United Kingdom National Member Society, ISPO, Sheffield, September, 1979.

the sound foot in front. When he has mastered these exercises, free walking in the parallel bars is undertaken. It is very important that the patient learns to use his knee right from the start, as once a bad habit is established it is very difficult to correct the fault.

Fig. 1. Left, one stage in initial walking training. The prosthetic foot is advanced and the knee flexed, progressively more weight is applied until the foot is flat on the floor (see text). Right, demonstration of standing up from a wheelchair. The wheelchair brakes should be applied before the patient attempts to stand.

Advanced walking exercises

Once the patient walks well in the parallel bars and feels confident, walking with crutches or walking sticks can be started; first with one crutch and supported by one parallel bar, then with two crutches or walking sticks but still inside the parallel bars for security. The patient is taught to walk with a four point gait. Once the patient feels confident with his sticks, walking outside the parallel bars can be started. Psychologically this is an important point in his treatment and the patient will take time to adjust to the greater space around him where people might pass by closely.

The patient is taught how to stand up from his wheelchair while holding his sticks (Fig. 1, right) and how to stand up from a chair without arm supports (Fig. 2).

Gradually the walking exercises are extended to out-doors, to walking with one stick, stepping over obstacles, ascending and descending inclines and ascending and descending stairs.

Dependent on age and ability, the patient is taught to ascend and descend stairs by using the prosthetic leg and sound leg alternately. Young patients are taught to run, jump and practice some equilibrium exercises on a beam.

Fig. 2. Standing up from a chair without arm supports. The patient leans forward until his weight is above the feet then pushes up into the standing position.

Gait deviations

To give correct walking training the therapist should be able to recognize gait deviations and their causes. Firstly, let us look at the components of a desired gait:

1. Foot flat on the ground during midstance
2. Lateral stability of the knee
3. Controlled knee use
4. Smooth movements
5. Minimal lateral bending of the trunk
6. Even step length, timing and arm swing.

The aim of the therapist is to achieve a gait pattern for the patient which corresponds as closely as possible to the desired gait. The degree of success which is attained is dependent upon the condition and motivation of the patient and the input of the therapist.

Deviations during stance phase

Between heelstrike and midstance—excessive or insufficient knee flexion.

Excessive knee flexion can be caused by:

1. Excessive dorsiflexion of the prosthetic foot or excessive anterior tilt of the socket.
2. Excessive anterior placement of the socket with respect to the foot.
3. Too stiff plantarflexion bumper or heel cushion.
4. Posterior displacement of cuff suspension points.

(This deviation can also be caused by a flexion contracture of the knee).

During normal locomotion, the knee is in approximately full extension at heelstrike; following this, the knee begins to flex and continues until foot flat (approx. 15–20 degrees).

Insufficient knee flexion can be caused by:

1. Excessive plantarflexion of prosthetic foot or insufficient anterior tilt of the socket.
2. Excessive posterior placement of the socket with respect to the foot.
3. Too soft plantarflexion bumper or heel cushion.
4. Anterodistal socket pain

(When the quadriceps contracts, pressure between the anterodistal surfaces of the stump and the socket increases considerably. This may cause stump discomfort at heelstrike.)

5. Weakness of quadriceps muscles. The patient compensates automatically by shortening the prosthetic step length, or by digging the heel into the ground with increased hip extension resulting in a lumbar lordosis, or with anterior bending of the body or with a combination of these.
6. Habit pattern; sometimes caused when the patient goes over from a BK prosthesis with a thigh corset to a PTB prosthesis or when during initial walking training there was not enough emphasis on using and trusting his knee.

During mid-stance there is sometimes an excessive *lateral thrust* of the prosthesis, this derives from the tendency of the prosthesis to rotate around the stump; when this occurs the medial brim presses against the stump while the lateral brim tends to gap. The patient sometimes compensates by bending the trunk laterally to reduce the pressure on the medial brim.

Between midstance and toe off, the centre of gravity passes over the MP joints. If this occurs too soon it results in insufficient anterior support and allows *premature knee flexion or "drop off"*. This can be caused by:

1. Excessive anterior placement of the socket with respect to the foot.
2. Posterior displacement of the toe-break or keel.
3. Excessive dorsiflexion of the prosthetic foot or excessive anterior tilt of the socket.
4. Excessively soft dorsiflexion bumper.

If the bodyweight is carried over the MP joints late the reverse occurs causing *delayed knee flexion*; the amputee experiences the sensation of walking uphill. This can be caused by:

1. Excessive posterior placement of the socket with respect to the foot.
2. Anterior displacement of toebreak or keel.
3. Excessive plantarflexion of the prosthetic foot or excessive posterior tilt of the socket.
4. Excessively hard dorsiflexion bumper.

Certain deviations may be observed during the swing phase of gait, such as:

1. Piston action due to improper suspension or poor socket fit.
2. Circumduction, when the patient does not sufficiently flex his knee out of habit, or when the prosthesis is too long.
3. Toe stubbing, due again to a too long prosthesis or insufficient knee flexion.

When a therapist recognises deviations which may be corrected by alterations to the limb alignment she should contact the prosthetist so that any necessary adjustments can be made. If a prosthesis is not well aligned the patient can suffer pressure sores and will develop bad walking habits. It is therefore of prime importance that the therapist and the prosthetist communicate with each other as members of the clinic team for the benefit of the patient.

Good walking training which is adjusted to the capability and age of the patient will enable an optimum gait to be achieved which might make all the difference between the individual staying at home bound to a chair, or leading a relatively normal existence.

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Book Review

Orthopaedic Engineering, Harris, D. and Copeland, K. (Eds.) Biological Engineering Society, London, 271p. £10.00.

This volume comprises the proceedings of the Conference held in Oxford at the Department of Orthopaedic Engineering in 1977.

The material is in four main sections, Spinal Research, Three Dimensional Measurement of the Body, Gait Analysis and Orthotic Design and Fabrication.

- The editors have deliberately decided to publish a range of papers in these topics and there is no coverage for other areas in biomechanics such as prosthetics or orthopaedic implants.

There are a total of forty-three chapters and the authors comprise forty scientists or technologists and twenty clinicians from the United Kingdom, there are eighteen from overseas and, of these, six are from North America. The coverage of the professional material is therefore strongly biased towards United Kingdom fields of activity and professional expertise.

The book is A3 in format and the print, though small, is clear and easily legible. Each chapter is well illustrated with line diagrams and where photographic records are presented these are reproduced with satisfactory clarity.

In the spinal research area the papers represent the many different attacks which are currently made on this problem; for example, cadaver investigation and the determination of mechanical properties of the relevant tissue materials; measurement of the biomechanics of the lumbar spine in the living individual and studies of the back muscle activity involving myo-electric techniques as well as investigations of work factors related to industrial incidence of back pain and the effect of spinal orthoses.

In the section on three dimensional measurement the principal problem considered is the measurement of the shape of the trunk and relative movements between spinal segments. The techniques presented are studio photography, Moiré topography, together with ultra-sound and the use of x-rays in static and dynamic analyses.

In the gait analysis section descriptions are given of the use of television cameras interfaced with computers, together with polarized light goniometers, mechanical goniometers as well as force platforms. The applications of the studies relate to, for instance, the interpretation of joint loading, the assessment of patients with arthritic conditions, as well as the investigation of racehorse injuries. There is one paper on prosthetics investigating the angular accelerations of below-knee limbs in order to determine optimal moment angle relationships in respect of knee flexion in the swing phase. The performance of below-knee amputees is assessed with a view to acquiring information relative to the design of mechanisms for application to above-knee prostheses.

In the orthotics design and fabrication section, there are studies of the loads transmitted by particular leg orthoses together with the clinical assessment of particular systems and materials of construction, with presentations relating to specific methods of orthoses construction.

There are also papers relating to the problem of support of the severely disabled patient, taking account of the requirements of stability and the avoidance of pressure sores.

This volume comprises a useful summary of the research activities in its chosen areas and it will be of considerable use to those requiring an introduction to these particular areas of study. It will be a useful book to have on the bookshelf for a limited number of years, recognising the general characteristic of the transient nature of books of this type, reporting the results and discussion of professional conferences and symposia. In due course the material will be stabilized in authoritative text books and the reader will be looking for the research presentations of the future for guidance in the developing fields.

J. P. Paul.
Glasgow, Scotland.

Obituary

We regret to announce that Hasan Hosni died suddenly of a heart attack at the age of 55 in September 1979.

Hosni, an excellent orthopaedic surgeon, served in the Egyptian Army with the rank of General. His untimely passing, which occurred on a holiday tour of Yugoslavia, the home country of his wife, will be felt as a serious loss to the Egyptian Orthopaedic Association. He was a pioneer in technical orthopaedics in Egypt and was responsible for establishing a modern comprehensive rehabilitation centre in Cairo. Recently he was co-operating with Mrs. Sadat in the planning and construction of a large new rehabilitation hospital in Cairo.

Hosni was an active Fellow of ISPO since its foundation. He was Chairman of the Egyptian National Member Society and a member of the International Committee of ISPO. As Regional Consultant for the Middle East he was also a member of the Executive Board. Hosni was principally responsible for the organization of the ISPO International Conference in Cairo in 1972, which was sponsored by Egypt and the U.S.A. In this international work he gained many good friends by his charm and intelligence.

ISPO wishes to honour the memory of Hasan Hosni and express sympathy to his widow and family.

Prosthetics and Orthotics International, 1979, 3, 168

Letter to the Editor

Dear Sir,

Your readers may be interested to learn that as a result of my participation in the ISPO Course on Above-knee Prosthetics at Rungsted and meeting so many enthusiastic members of ISPO, I was encouraged to form a committee from the staff of Nordland Sentralsykehus, Bodø to organize a multidisciplinary course on lower limb prosthetics to take place in Norway.

The course, which was entitled "Aspects of lower-limb amputee rehabilitation", was held at Nordland Sentralsykehus from 23–25 August, 1979 and was a great success. Those attending represented all the professions involved in amputee rehabilitation.

Bodø is situated on the coast of Northern Norway. The people of this region live on islands and by fjords, often in very isolated conditions. They make their living from fishing and farming and this presents problems of rehabilitation which are different from those of the more central parts of Northern Europe.

Topics covered in the course included amputation levels and techniques; pre-and post-operative care; normal and amputee locomotion; biomechanics; components and materials; principles of prosthetic fit and alignment; occupational and physiotherapy at each level of amputation; psychosocial aspects of amputation, and social rights. Ample time was allowed in the programme for discussion so that participants could compare techniques and experiences.

Lectures were given by surgeons, physicians, therapists, prosthetists and bioengineers. ISPO was well represented among the course speakers who included George Veres and Martin Goplen, respectively Chairman and Secretary of the Norwegian National Member Society—Bente Berild, Bodil Bjastad, Øystein Johansen, David Simpson (who had travelled from Glasgow to take part) and myself. You can be sure that opportunities were taken to state the aims of our Society and stress the benefits of membership.

Associated with the course was an exhibition displaying the range of prosthetic mechanisms available in Scandinavia and during the course each exhibitor made a presentation of their most recent developments.

The companies which took part and also generously sponsored the course were:

Otto Bock Scandinavia AB
Camp Scandinavia AB
Lindemann Ortopaedic
Oslo Bandasje og Proteseverksted

The social programme included a visit to Saltstrauman, where the participants had a meal and witnessed the spectacle of the most powerful tide race in the world. The course dinner was conducted with traditional Norwegian hospitality, culminating in dancing through the short Arctic night.

Participants on the course numbered 45 almost half of whom were members of ISPO. The course was considered so successful that we hope to organize a similar one for next year.

I hope that some of our speakers will submit their useful papers to "Prosthetics and Orthotics International" for consideration.

Turid Sandstrand
Head Physiotherapist
Nordland Sentralsykehus, Bodø.

Prosthetics and Orthotics International, 1979, 3, 169–173

Calendar of events

National Centre for Training and Education in Prosthetics and Orthotics

Short Term Courses 1980

NC 203 Knee-Ankle-Foot and Hip-Knee-Ankle-Foot Orthotics for Orthotists; 14–25 January, 1980.
NC 302 Lower Limb Prosthetics for Therapists; 28 January–1 February, 1980.
NC 206 Upper Limb Orthotics for Orthotists; 11–15 February, 1980.
NC 102 Lower Limb Orthotics for Physicians and Surgeons; 18–22 February, 1980.
NC 204 Patellar-Tendon-Bearing Prosthetics for Prosthetists; 3–14 March, 1980.
NC 205 Above-knee Prosthetics for Prosthetists; 17–28 March, 1980.

Further information may be obtained by contacting Prof. J. Hughes, Director, National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, 73 Rottenrow East, Glasgow G4 0NG, Scotland. Tel: 041–552 4400, extension 3298.

New York University Post-Graduate Medical School

Short Term Courses
Courses for Physicians and Surgeons

741 C Lower Limb Prosthetics; 25–29 February, 1980.
751 B Lower Limb and Spinal Orthotics; 24–29 March, 1980.
744 B Upper Limb Prosthetics and Orthotics; 7–11 April, 1980.
741 D Lower Limb Prosthetics; 28 April–2 May, 1980.
751 C Lower Limb and Spinal Orthotics; 5–10 May, 1980.

Courses for Therapists

745 A Upper Limb Prosthetics; 14–18 January, 1980.
742 B Lower Limb Prosthetics; 3–14 March, 1980.
742 C Lower Limb Prosthetics; 14–25 April, 1980.
752 C Lower Limb and Spinal Orthotics; 5–10 May, 1980.
745 B Upper Limb Prosthetics; 2–6 June, 1980.
757 B Upper Limb Orthotics; 9–13 June, 1980.

Courses for Orthotists

758 Upper Limb Orthotics; 7–18 January, 1980.
756 Spinal Orthotics; 27 May–6 June, 1980.
753 Lower Limb Orthotics; 7–25 July, 1980.

Courses for Prosthetists

746 Upper Limb Prosthetics; 21 January–1 February, 1980.
743 Above Knee Prosthetics; 9–27 June, 1980.

Course for Rehabilitation Counsellors

750 B Prosthetics and Orthotics; 23–27 June, 1980.

Requests for further information should be addressed to Professor S. Fishman, Prosthetics and Orthotics, New York University Post-Graduate Medical School, 550 First Avenue, New York, NY 10016, U.S.A.

Northwestern University Medical School

Short Term Courses
Courses for Physicians, Surgeons and Therapists

702 A, 703 A Spinal, Lower and Upper Limb Orthotics; 20–24 January, 1980.
702 B, 703 B Spinal, Lower and Upper Limb Orthotics; 11–15 February, 1980.
602 C, 603 C Prosthetics: Lower and Upper Limb; 3–7 March, 1980.
602 D, 603 D Prosthetics: Lower and Upper Limb; 17–21 March, 1980.
702 C, 703 C Spinal, Lower and Upper Limb Orthotics; 7–11 April, 1980.
602 E, 603 E Prosthetics: Lower and Upper Limb; 5–9 May, 1980.
631, 632, 633 Management of the Juvenile Amputee; 12–15 May, 1980.
602 F, 603 F Prosthetics: Lower and Upper Limb; 9–13 June, 1980.
602 G, 603 G Prosthetics: Lower and Upper Limb; 23–27 June, 1980.

Courses for Orthotists

731 Review Course in Orthotics for Orthotists.
751 Fitting and Fabrication of the C.T.L.S.O.; 16–20 June, 1980.

Course for Prosthetists

641 Review Course in Prosthetics; 31 March–2 April, 1980.

Course for Rehabilitation Personnel

640 Orientation to Prosthetics and Orthotics; 25–28 February, 1980
Further information may be obtained by contacting Mr. C. M. Fryer, Director, Prosthetic-Orthotic Center, Northwestern University Medical School, 345 East Superior Street, Room 1723, Chicago, Illinois 60611, U.S.A.

Oxford Study Days 1980

Rehabilitation after Stroke; 25–26 February, 1980.
Principles of Research for the Remedial Professions; 26–27 March, 1980.
Information: Secretary, Demonstration Centre, Mary Marlborough Lodge, Nuffield Orthopaedic Centre, Headington, Oxford.
Closing date for applications: two months before the date of each course.

1980
4th World Congress of the International Rehabilitation Medicine Association, Stockholm.
Information: International Rehabilitation Medicine Association, CH–7310 Bad Ragaz, Switzerland.

1980
Chartered Society of Physiotherapists Congress.
Information: Miss M. Auld, Physiotherapy Dept., Royal Infirmary, Aberdeen.

Spring 1980
8th Annual New England Bioengineering Conference, Cambridge, Massachusetts.
Information: Institution of Electrical and Electronics Engineers, Conference Services, 345 East 47th Street, New York, NY 10017, U.S.A.

4 January, 1980
Measurement of Forces and Pressures in and on the Body (BES).
Information: K. Copeland, RCS, Lincoln's Inn Fields, London WC2.

18–21 February, 1980
Finite Elements in Biomechanics, University of Arizona.
Information: University of Arizona, Tucson, Arizona, U.S.A.

6–8 March, 1980
Second Course on Locomotor System Biomechanics, St. Rafael Hospital, Barcelona.
Information: Srta. Margarita Garcia, Secretaria del 11 Curso de Biomecánica, Servicio de Cirugía del Aparato Locomotor, Hospital de San Rafael, Pso. Valle de Hebrón, s/n. Barcelona, 35 España.

23–29 March, 1980
Bio Engineering 80, London.
Information: BES, K. Copeland, University College, Gower Street, London WC1, U.K.

24–28 March, 1980
Aids for the Disabled, Dusseldorf.
Information: International Trade Fairs Ltd., 2 Old Bond Street, London W1, U.K.

31 March–1 April, 1980
Mechanical Factors and the Skeleton—the effects of forces and movements on bones, joints and connective tissue. A two-day scientific meeting to be held at the Oxford Orthopaedic Engineering Centre, Nuffield Orthopaedic Centre, Headington, Oxford.
Information: J. D. Harris, Director, Oxford Orthopaedic Engineering Centre. Tel: (0865) 64811, ext. 514/510.

8–12 April, 1980
1st World Congress on Biomaterials, Vienna.
Information: K. Copeland, University College, Gower Street, London WC1, U.K.

10–15 April, 1980
Third International Congress on Physically Handicapped Individuals who use Assistive Devices Houston, Texas, U.S.A.
Information: William A. Spencer, M.D., President, Scientific Programme Committee T.I.R.R. Texas Medical Centre, P.O. Box 20095, Houston, Texas 77025, U.S.A.

May, 1980
World Congress on Accident Prevention, Amsterdam, Netherlands.
Information: International Social Security Association, Case Postale 1, Geneva 22, Switzerland.

7–9 May, 1980
Engineering Aspects of the Spine, London.
Information: Institution of Mechanical Engineers, 1 Birdcage Walk, London SW1, U.K.

19 May, 1980
Practical Management of Handicapped Children.
Information: Castle Priory College, Thames Street, Wallingford, Oxford, OX10 0HE, U.K.

21–23 May, 1980
Fourth Annual International Rehabilitation Film Festival, Fordham University Complex, Lincoln Center, New York.
Information: Tim Moses or Beth George, Rehabfilm, Box F1, 20 West 40th Street, New York, NY 10018. Tel. (212) 869–0460.

23–27 May, 1980
Physical Medicine and Rehabilitation Stockholm.
Information: Dr. W. Moritz, Sydsvenska Sjukgymnast Institutet 5–220 05 Lund 5 Sweden.

29 May–2 June, 1980
Stress and Performance in Children with Cerebral Dysfunction.
Information: Castle Priory College, Thames Street, Wallingford, Oxford OX10 0HE, U.K.

June, 1980
24th Annual National Rehabilitation Conference, Arlington, Texas.
Information: D. Hall, 912 Rosevale Drive, Hewitt, Texas, U.S.A.

15–20 June, 1980
3rd Annual Interagency Conference on Rehabilitation Engineering, Toronto, Canada.
Information: Joseph E. Traub, Rehabilitation Services Administration, HEW, 220 C St. S.W., Washington, DC 20201, U.S.A.

20–27 June, 1980
International Conference on Rehabilitation Engineering, Sheraton Centre, Toronto, Canada.
Information: CMC, Inc., 5401 Kirkman Road, Suite 550, Orlando, Florida 32805, U.S.A.

21–22 June, 1980
Plastics in Medicine and Surgery, Enschede Netherlands.
Information: J. N. Ratcliffe, Plastics and Rubber Institute, 11 Hobart Place, London S.W.1.

22–27 June, 1980
Rehabilitation International 14th World Congress, Winnipeg, Canada.
Information: Mr. Jack Sarney, Canadian Rehabilitation Council for the Disabled, Suite 2110, Yonge Street, Toronto, Onatario M5E 1E8, Canada.

23–28 June, 1980
Helen Keller Congress, Boston, Massachusetts.
"Blueprint for the Future".
Information: American Foundation for the Blind, Inc., 15 West 16th St., New York, NY 10011.

July, 1980
Helen Keller Centennial World Conference on Deaf-Blindness, Hannover, West Germany.
Information: World Council for the Welfare of the Blind, 58 Avenue Bosquet, 75007 Paris, France.

August, 1980
National Multiple Sclerosis Society Conference, Denver Colorado, U.S.A.
Information: Sylvia Lawrie, 205 East 42nd Street, New York, NY 10017, U.S.A.

17–22 August, 1980
8th International Congress of Physical Medicine and Rehabilitation on the theme "Disability-Prevention and Management by Rehabilitation Medicine" Stockholm, Sweden.
Information: Secretary, Dr. I. Swedborg, Avd. for Fys. Med. och Rehab., Karolinska Sjukhuset S–104 01, Stockholm, Sweden.

Change of dates: 25–29 August, 1980
Physical Medicine/Rehabilitation: 8th Congress of the International Federation of Physical Medicine, Stockholm, Sweden.
Information: Physical Medicine, c/o Stockholm Convention Bureau, Jacobs Torg 3, S–111 52, Stockholm, Sweden.

September, 1980
Mediterranean Conference on Medical and Biological Engineering, Marseilles, France.
Information: Prof. G. Kapham, Faculté de Médecine (Nord), Boulevard P-Drummard, 13326 Marseilles Cedex III, France.

Mid-September, 1980
American Orthotic and Prosthetic Association Meeting, New Orleans, U.S.A.
Information: M/s. S. I. McCamley, American Orthotic and Prosthetic Association, 1444 N. Street NW, Washington, DC 20005, U.S.A.

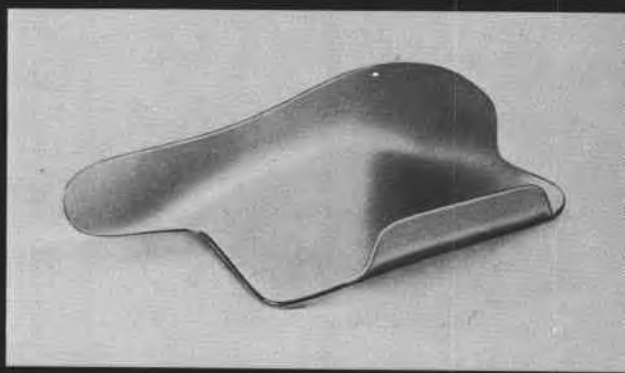
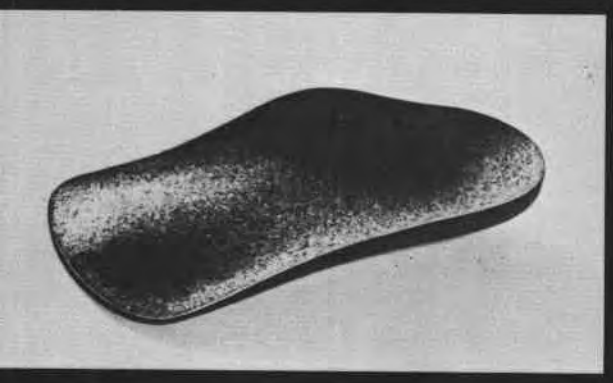
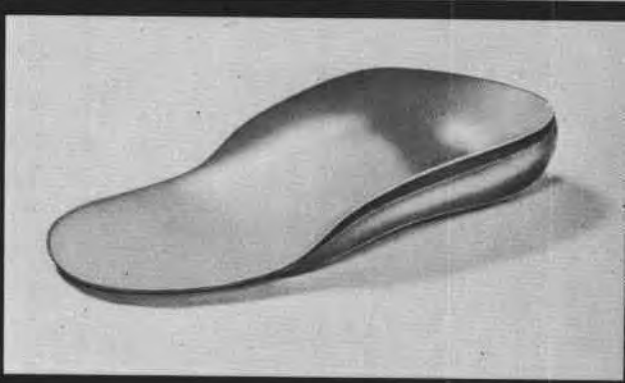
27 September–1 October, 1980
33rd Annual Conference on Engineering in Medicine and Biology, Washington DC, U.S.A.
Information: M/s. P. I. Horner, Administrative Director, Alliance for Engineering in Medicine and Biology, 4405 East-West Highway, Suite 404, Bethesda, Maryland 20014, U.S.A.

28 September–4 October, 1980
ISPO 3rd World Congress, Bologna, Italy.
Information: Studio B.C., via Ugo Bassi 10,40123 Bologna, Italy.

17 October, 1980
B.E.S. Scientific Meeting and A.G.M.
Information: K. Copeland, R.C.S. L.I.F., London W.C.2.

16–21 November, 1980
First International Convention on Medico-Legal Aspects of Disability, Tel Aviv, Israel
Information: Convention Secretariat, P.O. Box 3059, Tel Aviv, Israel.



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(2) Authors' names, initials and titles. The present address of any author if different from the place where the work was done, may be shown as a footnote.
(3) Department(s) in which the work was done.
(4) The name and full postal address of the author to whom correspondence and requests for reprints should be directed. This will appear as a footnote.

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All illustrative material should be lightly marked on the back in pencil with the figure number in arabic numerals, title of paper, authors' name and a clear indication of the top of the figure. The approximate location in the text should be marked. Figure captions should be typed on a separate sheet. Tables should be used only when necessary to clarify important points. Each table should be typed on a separate sheet and numbered consecutively in arabic numerals.

References

References in the text should follow the author/date system for example: Peizer (1971). If there are more than two authors—Solomonidis *et al.* (1974). References at the end of articles should be listed on a separate sheet in alphabetical order of (first) authors' name, as follows: Marx, H. W. (1974). Lower limb orthotic designs for the spastic hemiplegic patient. Orthotics and Prosthetics, 28(2), 14–20. Journal titles must be given in full.

References to articles in books should include author, year of publication, article title, book title edition, editor (if different from author) first and last pages, publisher and place of publication. For example, Hughes, J. (1975). Recent developments in prosthetics and orthotics. Recent Advances in Orthopaedics (2) Ed. McKibbin, B., 196–216, Churchill Livingstone, Edinburgh.

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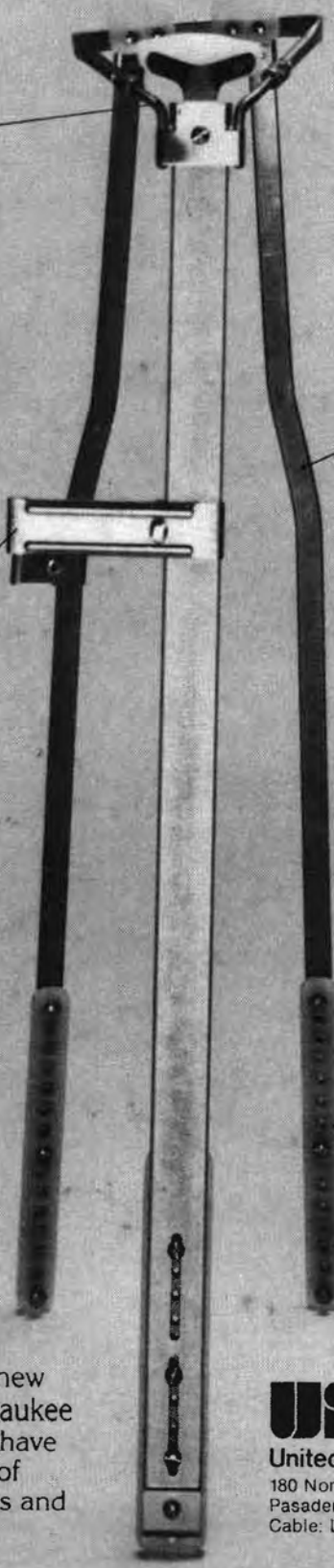

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