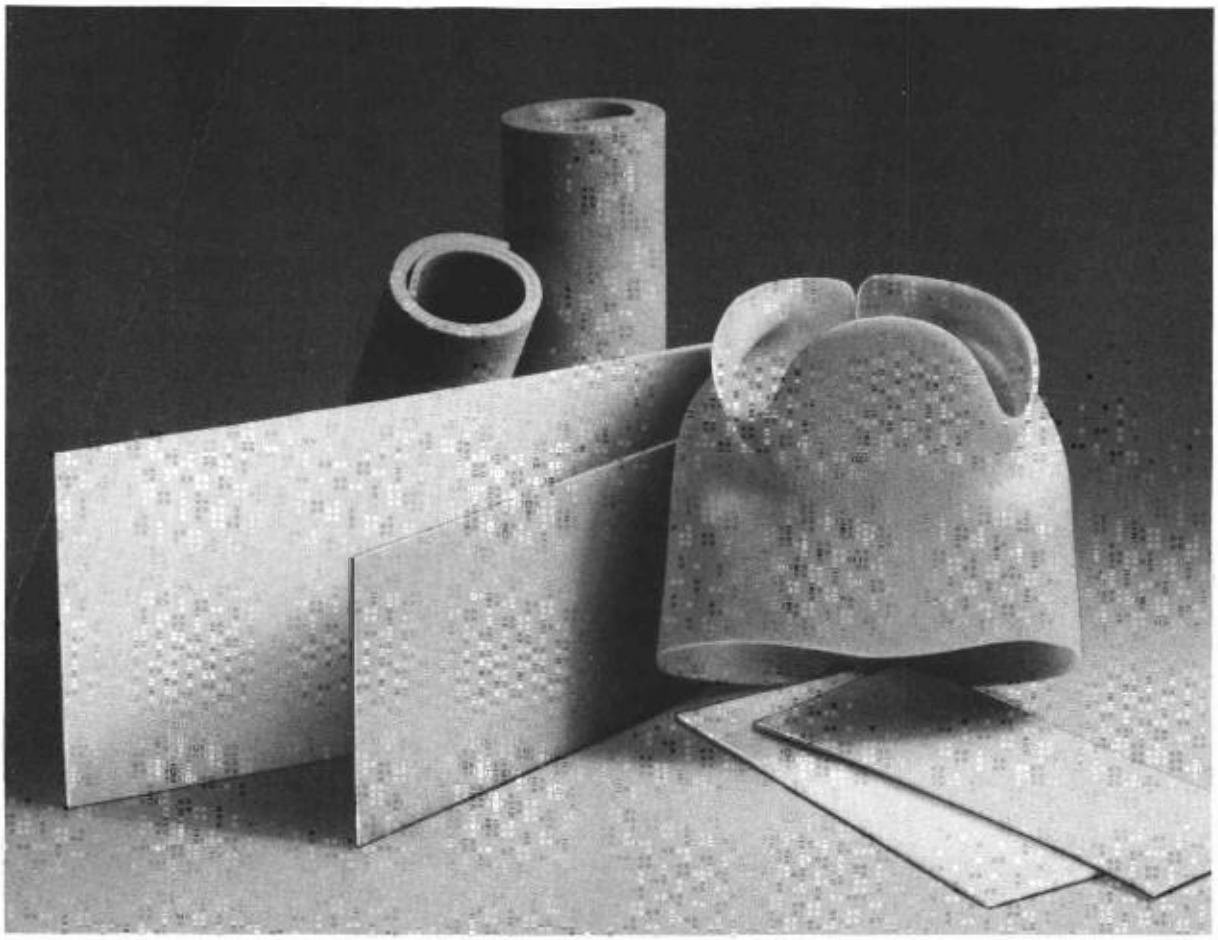
The Journal of the International Society for Prosthetics and Orthotics

Prosthetics and Orthotics International

August 1982, Vol. 6, No. 2



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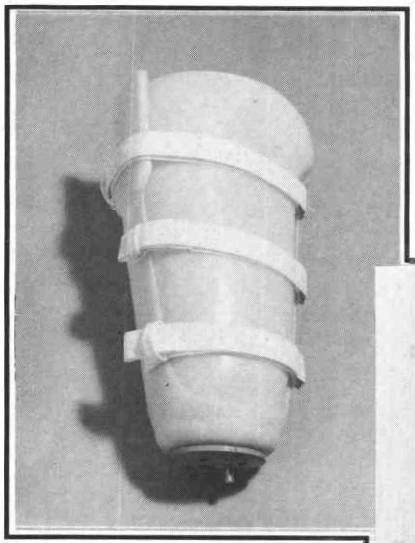

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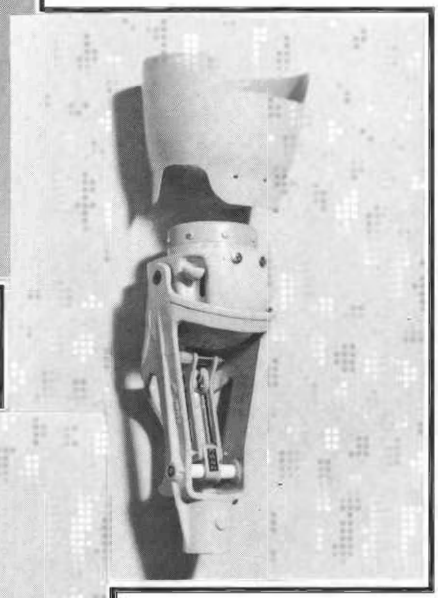

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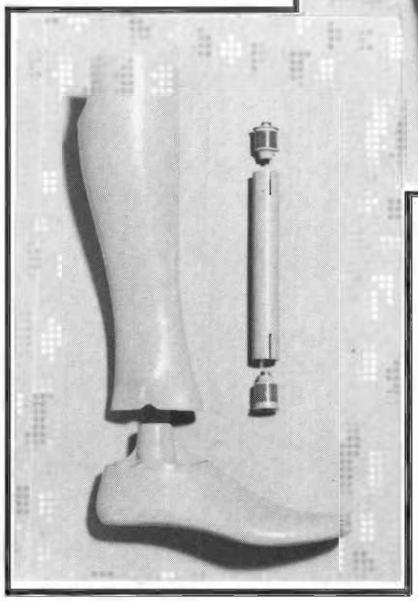

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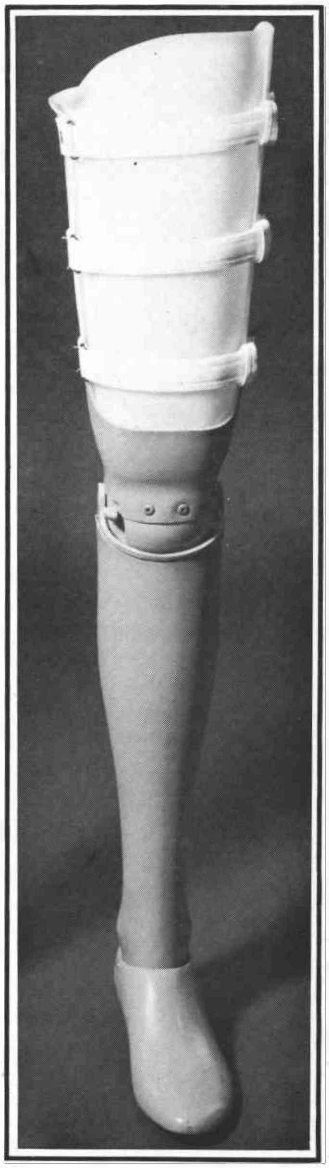

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The Journal of the International Society for Prosthetics and Orthotics

August 1982, Vol. 6, No. 2

Contents

Executive Board of ISPO

Prosthetics and Orthotics International, 1982, 6, 67–68

Editorial

The month of May in Denmark is a delightful time with sunshine and Spring blossoms. This year was no exception, and from the Society's point of view was further embellished. First, the Advanced Course in Below-Knee and Through-Knee Amputations and Prosthetics was held in Køge at The Hotel Hvide Hus. It proved to be an unqualified success with over 100 participants and a number of commercial sponsors who presented a first class exhibition covering the whole range of products relevant to the discussions. The course itself was at a high level with world authorities in their various specialties debating with vigour and good humour important issues in patient management. It was in many ways reminiscent of the courses held some years ago in the Orthopaedic Hospital, Copenhagen. For those of us who were privileged to attend those courses it was both welcome and refreshing. I believe it was especially pleasing to our founding President, Knud Jansen.

I now turn to the second matter of importance to the Society which happened in the month of May. The Executive Board learned that Knud Jansen was about to retire from active practice, although we found it difficult to imagine that he will no longer see his patients, teach his students, perform

operations, edit journals, chair committees and act as guide, philosopher and friend to many hundreds of colleagues throughout the world. It is with the greatest pleasure that I can announce that the Board decided that we would mark Knud's retirement by his appointment as The President Emeritus of our Society. I am glad to say that he graciously accepted this honour along with "The President's Chair" as a token of our appreciation for all that he has given of himself to the Society. The Executive feels confident that the membership of the Society will approve of its action and is relieved to know that Knud Jansen will continue working for the disabled, and particularly for those in the developing countries as Vice-President of World Orthopaedic Concern and as Chairman of ISPO's Task Force for the Developing Countries.

George Murdoch

Immediate Past President.

Just before going to press we have learned with regret of the resignation of Anthony Staros as President of ISPO. This is due to a change in his responsibilities. We are sure members will join us in wishing him well in his new activities and thanking him for his hard work for the Society in the past.

Prosthetics and Orthotics International, 1982, 6, 69–74

Fourth World Congress

5–9 September, 1983, London

Secretary General—George Murdoch

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GENERAL ARRANGEMENTS

Time and place

The Fourth Congress of the International Society for Prosthetics and Orthotics will be held in the Imperial College of Science and Technology, London from 5th to 9th September, 1983.

Imperial College of Science and Technology

The College is situated in a very pleasant area of London close to Hyde Park with a wide selection of hostels and hotels nearby. The venue offers excellent conference facilities combined with a large exhibition area for both the trade and scientific exhibits.

Programme

The Congress programme will offer morning plenary and afternoon concurrent sessions covering prosthetics, orthotics and all other aspects of rehabilitation engineering. Amongst the main topics to be discussed are amputation surgery, neuro-muscular disorders relative to extremity orthotics, spinal disorders, spinal cord injury, multiple sclerosis, arthritis and the multiply handicapped person. The official language of the Congress will be English and there will be no simultaneous translation.

Social programme

A programme of social events will be organized for all participants and will include a welcoming reception and a Friendship Evening encompassing the unique character of Imperial College itself and other Institutions within the city of London. In addition one evening will be devoted to a cruise on the River Thames.

Exhibition

The College offers excellent facilities for the exhibition and displays and we anticipate a large number of companies and organizations will participate. Exhibitors will be offered spaces with or without a shell scheme in two halls in very close proximity to each other. Continuous coffee and tea will be available within the exhibition area and pub-style lunches—sandwiches, beers and wines—will be available for purchase there.

Plans of the proposed layout will be available shortly when bookings may be made.

Delegate tours
Wednesday afternoon has been kept free of sessions so that all participants may have an opportunity to explore London. For those who are unfamiliar with the London scene there will be a selection of interesting tours to choose from covering both the City itself and sites close by.

Accompanying persons programme
In addition to the Tours on Wednesday afternoon there will be a programme of tours and visits for accompanying persons to places of interest in and around London.

Accommodation
There is a wide range of hotels and University Halls of Residence costing upwards of £12 per night and within walking distance of Imperial College. Block bookings have been made for this Congress so that delegates may take advantage of favourable rates.

Pre- and Post-Congress visits and courses
A programme of professional pre-and post-Congress visits and courses is being developed which will be offered along with the Course registration.

For further information and details of advance bookings please contact:

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(Tear-off slip on page 74)

Provisional programme

Plenary sessions
The five plenary sessions will take place during the mornings of the five week days of the congress. Each will comprise an authoritative review paper in the subject area followed by supporting papers. The first session (Monday) will follow immediately after the opening ceremony and will commence with the Knud Jansen lecture.

First Plenary Session: Monday 5th, 11.15–1.00
—Lower Limb Amputation

Second Plenary Session: Tuesday 6th, 10.15–12.00
—Lower Limb Disabilities

Third Plenary Session: Wednesday 7th, 10.15–12.00
—Upper Limb Prosthetics and Orthotics

Fourth Plenary Session: Thursday 8th, 10.15–12.00
—The Severely Disabled

Fifth Plenary Session: Friday 9th, 11.45–1.00
—Spinal Problems

Concurrent sessions

Submitted papers and panel discussions

There are seven afternoon periods for concurrent sessions—two on Monday, Tuesday, Thursday, and one on Friday. Two of the concurrent sessions on each of the seven occasions will take the form of a panel discussion. Five of the panels will concentrate on the subjects of the plenary sessions and for continuity will employ the same chairmen and include plenary speakers as panellists. The remaining three concurrent sessions each afternoon are devoted mainly to submitted papers. These papers, which will be vetted as far as possible from abstracts, are expected to report innovation and to promote discussion.

Panel discussions

	Early afternoon	*Late afternoon*
Monday	Lower Limb Amputation Surgery	The Special Problems of the Elderly
	Back Pain: Management & Social Problems	Establishing a Service in Developing Countries
Tuesday	Lower Limb Disability	Gait Analysis
	Prosthetics and Orthotics Education	The Patient's Viewpoint Combined with Psycho-Social Considerations
Wednesday	No afternoon session	
Thursday	The Severely Disabled	Upper Limb Prosthetics and Orthotics
	Arthritis and the Hand	The Foot and Footwear
Friday	Spinal Problems Biofeedback	

Poster sessions

Poster session will be held in parallel with each of the concurrent sessions.

Films and Videotapes

Presentations in the form of film and videotape will be invited for screening on weekdays nominally from 1.30 to 6 p.m. and in parallel with the submitted paper sessions. A schedule will be prepared by the AV team when all the offered titles and timings are known.

Instructional courses (Table 1)

These courses will run all day Sunday (8 hours) and from 8 to 10 a.m. on each of the five weekdays (10 hours).

Table 1—Instructional Course Programme

	Sunday 0900–1300 1400–1800	Monday 0800–1000	Tuesday 0800–1000	Wednesday 0800–1000	Thursday 0800–1000	Friday 0800–1000
Lower Limb Prosthetics	Introductory Biomechanics and Normal Locomotion (2 hrs) Below-Knee and Syme Prosthetics (6 hrs)	Above-Knee and Knee Disarticulation Prosthetics (2 hrs)	Above-Knee and Knee Disarticulation Prosthetics (Cont) (2 hrs)	Above-Knee and Knee Disarticulation Prosthetics (Cont) (2 hrs)	Hip Disarticulation Prosthetics (2 hrs)	Management of the Bilateral Amputee (2 hrs)
Orthotics	Spinal Orthotics (4 hrs) Lower Limb Orthotics (4 hrs)	Lower Limb Orthotics (Cont) (2 hrs)	Lower Limb Orthotics (Cont) (2 hrs)	Upper Limb Orthotics (2 hrs)	Upper Limb Orthotics (Cont) (2 hrs)	Upper Limb Orthotics (Cont) (2 hrs)
Rehabilitation Engineering	Communication Aids (4 hrs) Clinical Gait Analysis (4 hrs)	Wheelchairs (Incl. Adaptations and Prescription) (2 hrs)	Wheelchairs (Incl. Adaptations and Prescription) (Cont) (2 hrs)	Seating for the Severely Disabled (2 hrs)	Seating for the Severely Disabled (Cont) (2 hrs)	Functional Electrical Stimulation (2 hrs)
Rehabilitation	Rehabilitation of Stroke Patients (4 hrs) Scoliosis (4 hrs)	Gait Training (2 hrs)	Cerebral Palsy (2 hrs)	Cerebral Palsy (Cont) (2 hrs)	Spina Bifida (2 hrs)	Spina Bifida (Cont) (2 hrs)
Orthopaedics	Fracture Bracing (4 hrs) Amputation Surgery (4 hrs)	Amputation Surgery (Cont) (2 hrs)	Amputation Surgery (Cont) (2 hrs)	Amputation Surgery (Cont) (2 hrs)	Paediatric Problems (Perthes: CDH: Clubfoot) (2 hrs)	Extension Prostheses (2 hrs)
Other	Partial Foot Prosthetics (2 hrs) Upper Limb Prosthetics (6 hrs)	Upper Limb Prosthetics (Cont) (2 hrs)	Footwear and Adaptations (2 hrs)	Footwear and Adaptations (Cont) (2 hrs)	Footwear and Adaptations (Cont) (2 hrs)	Footwear and Adaptations (Cont) (2 hrs)

Provisional programme

SUNDAY	SEPT. 4	**morning:** 0900–1300: Instructional courses
		afternoon: 1400–1800: Instructional courses
MONDAY	5	**morning:** 0800–1000: Instructional courses 1015–1115: Opening ceremony 1115–1300: First plenary session
		afternoon: 1400–1700: Concurrent sessions*
TUESDAY	6	**morning:** 0800–1000: Instructional courses 1015–1200: Second plenary session
		afternoon: 1330–1700: Concurrent sessions
WEDNESDAY	7	**morning:** 0800–1000: Instructional courses 1015–1200: Third plenary session
		afternoon: free
THURSDAY	8	**morning:** 0800–1000: Instructional courses 1015–1200: Fourth plenary session
		afternoon: 1400–1700: Concurrent sessions
FRIDAY	9	**morning:** 0800–1000: Instructional courses 1015–1145: World assembly 1145–1300: Fifth plenary session
		afternoon: 1400–1600: Concurrent sessions

*Concurrent sessions will include submitted papers, panel discussions, poster sessions and film/videotape programme.

CALL FOR PAPERS, EXHIBITS, FILMS, AND POSTERS

Papers, Exhibits, Films/videotapes and Poster presentations are invited on prosthetics, orthotics and rehabilitation engineering. Particular areas of interest for the Congress are amputation surgery, neuro-muscular disorders relative to extremity orthotics, spinal disorders, spinal cord injury, multiple sclerosis, arthritis and the multiply handicapped person. Please complete this form, detach and mail promptly. PLEASE PRINT.

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Prosthetics and Orthotics International, 1982, 6, 75–78

The development of an assessment chair

G. I. BARDSLEY and P. M. TAYLOR*

Limb Fitting Centre, Dundee
**School of Bioengineering, University of Dundee*

Abstract

Initial difficulties in producing seats for the physically disabled have led to an investigation of the process of seat prescription. An adjustable assessment chair was developed for the purpose of identifying patients' seating requirements.

The main variables of the chair's configuration are controlled by a number of hydraulic and mechanical systems.

Different support surfaces can be attached to the chair to simulate different seat characteristics. At present a bead bag vacuum consolidation system is used to simulate moulded seats.

Experience to date has shown that the assessment chair performs a valuable clinical role in the provision of seating.

Background

The early stages of a seating research programme at Dundee Limb Fitting Centre involved the production of prototype chairs for physically disabled patients. It soon became apparent that because of their very individual characteristics and problems, these patients require a highly personalized form of seating. Experience has enabled the accurate selection of the required type of seating system but difficulties were still encountered in identifying the exact configuration of seating which would fulfil the individual's particular requirements.

An iterative empirical approach to the problem was found to be applicable. The subject was supported in one position and their reactions observed for a period of time. The support was then modified, depending upon the observations, to produce a new position. The process was continued until the requirements of body configuration, stability, comfort, function, etc. were all fulfilled or acceptable compromises achieved.

Initially subjects were positioned manually by the assessor physically holding them. This technique, however, is limited to children whose body weight can be handled. In addition, its accuracy is limited by the time that positions can be held, by the inadequacy of control of posture and by failure to properly simulate the support available in the final product. Seats produced in this way often had to be returned or modified because of assessment inaccuracies.

Consequently, some form of device was considered desirable to assist the process of identifying optimum configurations. In the past, a few adjustable chairs have been designed for this purpose or as definitive chairs, however they tend to be limited in the seating variables which they can control and limited to the production of one particular type of seating (University of Tennessee, 1977; Wijkmans and de Soete, 1978) (Commercial Information).

Objectives

In view of these observations, it was decided to design and construct a device which was sufficiently adjustable to simulate all configurations of seating systems likely to be prescribed. The primary application of the device would be to act as a clinical tool for determining optimum seating configurations for individual patients, however it was also envisaged as acting as a research tool for developing new seating systems and for investigating sitting posture.

All correspondence to be addressed to Dr. G. I. Bardsley, Dundee Limb Fitting Centre, 133 Queen Street, Broughty Ferry, Dundee DD5 1AG, Scotland.

Design concepts and criteria

The concept of the device was based upon a highly adjustable framework which would accept a series of different interfaces between the subject and framework (Fig. 1). The framework would control the configuration of the seat and the interfaces would simulate the characteristics of the prescribed seating system each system having its own series of interfaces.

All joint angles, relative positions, and longitudinal dimensions of the seating components were to be variable to accommodate all anticipated postures and body sizes, the latter ranging from the obese adult down to the baby first learning to sit upright. In addition the height and attitude with respect to the ground of the whole seating assembly were to be adjustable. In view of the expected range of body weights, it was envisaged that some adjustments would require power assistance.

All mechanisms were to be kept away from the seating surfaces to permit easy access to the chair and its occupant. Since the device was being designed for use in the clinic, its appearance and method of adjustment had to be acceptable to the patients for whom it was intended. It was anticipated however that it would be operated only by skilled clinical personnel.

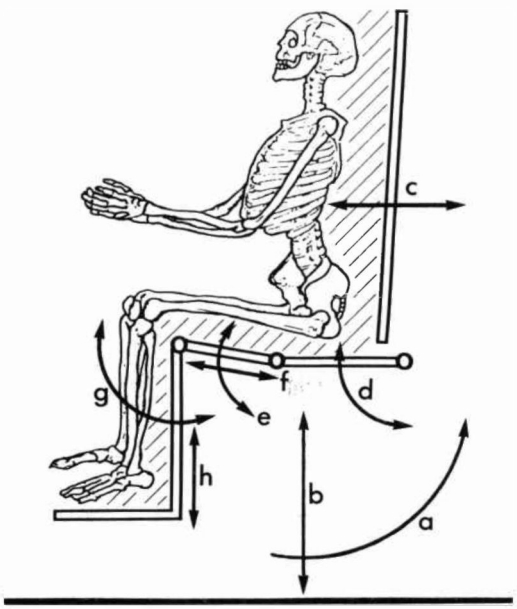

Fig. 1. The design concept.
a—chair attitude
b—chair height
c—backrest position (seat length)
d—pelvis backrest angle
e—hip angle
f—thigh angle
g—knee length
h—shank length
Adjustments e—h are independent for left and right legs. Adjustments a—e are powered. The seat interface is indicated by the diagonal lines.

Design details

Base frame

The base frame of the chair is a parallelogram linkage which permits independent control of the height and the anteroposterior attitude of the seating with respect to the ground (Figs. 2 and 3). The height of the pelvic support may be varied from 14 cm to 75 cm above the ground and the ankle of the backrest from 90° to 10° with respect to the horizontal. The rear of the base frame supports the electrical and hydraulic control gear for the chair.

At present there is no provision for mediolateral adjustment of the chair attitude. A few situations have arisen where this control would have been useful and its inclusion in the design is now being considered.

Seating framework

The main seating framework is divided into three articulated sections comprising the pelvic platform, and left and right thigh supports. This arrangement permits independent control of the pelvis and the left and right thighs. The two thighrests may be locked together if independent operation is not required and their lengths are adjustable with two different assemblies being used to span both adult and child sizes. The depth of the pelvic platform is governed by the position of the backrest which can be brought

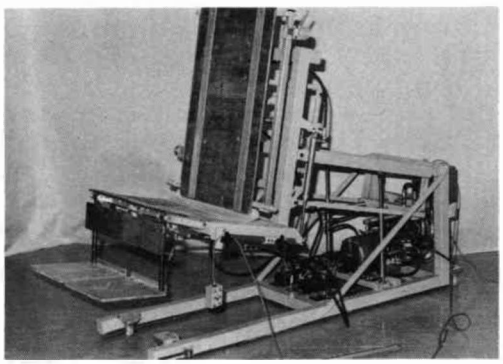

Fig. 2. The finished assessment chair.

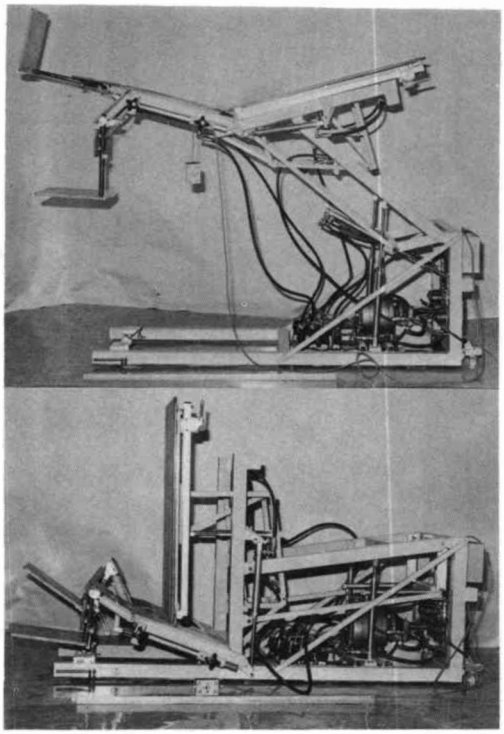

Fig. 3. Top, seat in maximum extension. Bottom, seat in flexion and reduced in size.

forward for this purpose. Adjustable head and trunk lateral supports can be fitted to rails on the backrest. The ends of the thigh supports have attachments for shank and footrests. These can be adjusted to accommodate a range of knee angles and shank lengths. Finally, adjustable arm supports are currently being developed for assisting independent egress and ingress.

Powered adjustments

Adjustments involving significant proportions of body weight are powered to ease operation of the chair. These consist of chair height and attitude, backrest position, pelvic support angle, and left and right thigh support angles. Power is derived from an electrically driven hydraulic pump acting on 20 mm diameter hydraulic cylinders at 80 bar (1200 psi). The movements are controlled by electro-mechanical valves operated by 6 switches at the end of a remote control cable (Fig. 4, top).

Interfaces

One of the major objectives of our current research programme has been the development of improved techniques for producing moulded seats. Consequently the first interface developed for the chair was the bead bag vacuum consolidation system used to simulate the support provided by a moulded seat (Nelham, 1975). The system consists of 6 bags whose firmness is controlled independently by a six channel vacuum pump (Fig. 4, bottom). This arrangement permits separate adjustment to the support of different body sections. The bags are placed directly onto boards fitted to the chair framework.

Other interfaces are to be developed to simulate other seating systems. For example, a series of cushions, armrests and lateral trunk supports is in the process of being developed to simulate seating for the elderly. A further series of preformed plastic modules is planned to simulate a modular seating system for handicapped children similar to that currently in use in Memphis, Tennessee (University of Tennessee, 1977).

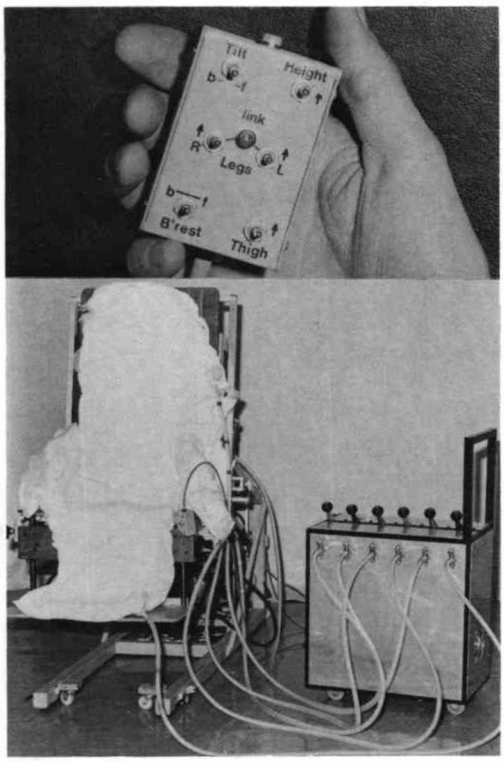

Fig. 4. Top, the control box. Bottom, chair in use with bean bag vacuum consolidation system.

Conclusions

Previous to this work, the authors were uncertain as to whether an assessment chair as described above would be a help or a hindrance in the clinical provision of seating. To date, seventeen moulded seats have been made using the chair. In all cases it has proved to be of great assistance, permitting quick, accurate postural adjustments, several of which cannot be contemplated without this type of control. For example, the anteroposterior attitude of a chair with respect to the horizontal has been found to be critical to within a few degrees for many cerebral palsy patients, as it has a major influence upon head control and overall muscle tone. The chair permits the fine control necessary to identify the optimum setting of this variable.

The design of the device has ensured that all mechanisms are unobtrusive and do not impede access to the subject. Despite its unusual appearance, all subjects have accepted the use of the chair and some children even enjoyed the experience. In summary, the assessment chair has proved to be a valuable clinical tool in the provision of seating for the physically handicapped. Although extra time is required in the assessment stages, it would appear that better seats can be produced and final modifications which were required in previous techniques may now be avoided.

Acknowledgements

The seating research programme is financially supported by the Chief Scientist's Office of the Scottish Home and Health Department.

Mr. Taylor has been funded by the Manpower Services Commission and the Norman Fraser Design Trust.

Facilities have been provided by Tayside Health Board.

REFERENCES

NELHAM, R. L. (1975). The manufacture of moulded supportive seating for the handicapped. *Biomed. Eng.*, **10(10)**, 379–381.

UNIVERSITY OF TENNESSEE (1977). Specialized seating for the severely handicapped. In: *Report of activities, 2.* July 1, 1976—October 31, 7–9.

WIJKMANS, D. W. AND DE SOETE, H. W., (1978). Further developments on the vacuum form fixation techniques. In: *Progress Report No. 6,* Institute of Medical Physics, TNO, Da Costakade 45, 3521 VS, Utrecht.

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Prosthetics and Orthotics International, 1982, 6, 79–87

The physical effect of lumbar spinal supports

N. D. GREW[1] and G. DEANE[2]

Oxford Orthopaedic Engineering Centre, University of Oxford

Abstract

A study has been performed to investigate the physical effects of lumbar spinal supports. Two groups were studied, a group of normal male subjects and a group of male low back pain patients. Five different spinal supports were investigated and their effects upon the skin temperature, spinal movements and intra-abdominal pressures of these individuals were examined. The results show surprisingly similar patterns for the widely varying designs of support. The findings also suggest that the longer term wearing of a spinal support results in a degree of physical dependence. The results of this study are aimed at improving the prescription and use of spinal supports in the treatment of low back pain.

Introduction

Lumbar spinal supports form a major part of the treatment of low back pain to the extent that each year over quarter of a million are prescribed in England and Wales. Supports available for prescription fall into the general categories of spinal brace and fabric corset, although there are many types and constructions. The basis upon which a support is prescribed is unclear because little is known about the performance of such orthoses in terms of their mechanical and other effects upon the wearer. Perry (1970) showed that almost all orthopaedic surgeons prescribed external supports at least occasionally in their treatment of low back pain. The most common diagnosis for which a corset is prescribed was found by Ahlgren and Hansen (1978) to be chronic lumbago and the main reason for the patients using this form of treatment was that it provided "support", with or without the relief of pain.

The following possible effects of spinal supports may have an important role in terms of their therapeutic value:

(a) Limitation of movement.
(b) Alterations of intra-cavity pressures.
(c) Modification of muscle actions.
(d) Warming of skin.

The use of a rigid brace for restriction of movement and stabilization is widespread (Perry, 1970) but even the more flexible supports are obviously intended to modify movements in some way. The efficacy of spinal braces has been questioned (Norton & Brown, 1957) in particular with regard to intervertebral movements. While gross movements are prevented, individual vertebral movements are sometimes increased.

The abdominal cavity, sometimes in conjunction with the thoracic cavity, is pressurized voluntarily when the spine is put under stress (Bartelink 1957; Eie & Wehn 1962; Davis & Snoup, 1964; Kumar & Davis 1973). This activity has a direct effect on the spinal loading by introducing a distending force anteriorly. This force produces an extension moment about the lumbar spine which reduces the tension required in the posterior spinal muscles. An inflatable corset increases the resting abdominal cavity pressure by about 10–15 mm Hg, but does not raise the peak pressures seen during a controlled lift (Morris et al, 1961). The effect of normal spinal supports was studied here.

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The effects upon muscle activities of wearing a spinal brace have been investigated using EMG (Waters & Morris, 1970). The findings were not consistent and indicated little or no change in activity. The effect of an inflatable corset is to reduce muscle activity considerably, especially during activities which apply high loads to the spine (Morris et al, 1961).

Dixon et al. (1972) showed that some forms of chronic low back pain respond as well to the wearing of a woolly belt as to an ordinary corset.

They concluded that an increase in the lumbar skin temperature was the cause of this symptom relief. Lumbar and thoracic skin temperatures were recorded in the tests of corsets and braces made in this study.

Materials and methods

The supports which were studied are described in Table 1. Eight patients and 10 healthy volunteers were studied while wearing these supports. The 10 healthy men who did not suffer from low back pain had ages ranging from 25 to 46 years, with a mean of 37 years. The patient group was composed of eight men who suffered chronically from low back pain, whose symptoms had settled and who had been wearing lumbar spinal supports regularly for at least three months. Their ages ranged from 30 to 61 years, with a mean of 42 years.

Table 1. Description of the supports tested.

Symbol	Description
NS	No support
SE	Semi-elasticated, narrow corset. Padded lumbar insert semi-conforming to lordosis. Rigid anterior section.
NF	Narrow fabric corset with some posterior strengthening.
LF	Long fabric corset extending from pelvis to thorax. Some steel posterior strengthening and some padding.
RB	Leather covered steel brace. Pelvic and thoracic hoops linked by longitudinal members. Anterior abdominal pad.
PJ	Polythene jacket.

Each subject performed the tests first of all without a lumbar spinal support to enable base line readings to be made, and then repeated the tests wearing a variety of supports. The study sought to answer the following questions:

(a) How do the individual support types influence the measured parameters and can they be characterized in terms of their effects upon the wearer?

(b) Are low back pain patients for whom a spinal support is a regular part of treatment affected in a different manner to the normal group?

Three parameters were chosen for measurement, these being range of lumbar spinal movement, skin temperature, and intra-abdominal pressure.

Lumbar spine movement

Lumbar spine movements are difficult to measure because they are usually accompanied by hip movements. These must either be measured separately and taken into account, or prevented. The latter course was decided upon and the method adopted uses a pelvic constraint frame (Fig. 1). It is accepted that by adopting this approach, the movements being measured were not strictly normal, but nevertheless would represent the degrees of immobilization provided by the orthoses. Another feature of the technique was that, because it isolated the movements of the lumbar spine, the results were more likely to be repeatable. The movement of the lumbar spine was measured by means of the vector stereograph (Morris & Harris, 1976). This

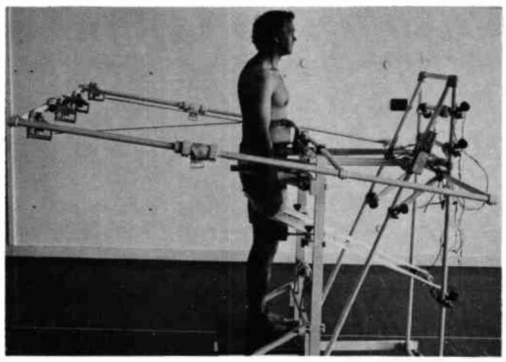

Fig. 1. The pelvic constraint frame.

instrument records movement of a selected point by electrical recording of the lengths of string attached to that point. In this study two strings were used and a two-dimensional recording was used to record the movement in a horizontal plane (Fig. 2). The errors involved in this simplification were minimized, as described by Grew and Harris (1979). The stereograph strings were arranged to intersect at the level of the spinous process of T12 and were held in place on the back by a belt (Fig. 1). Once held in the

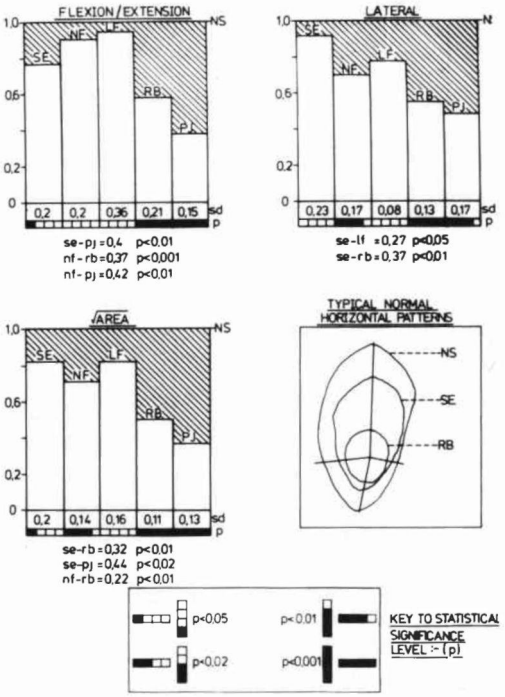

Fig. 2. Normalized spinal movement results from normal subjects.

frame with the stereograph attached, the subject performed a sequence of movements listed in Table 2.

All movements were to the limit of comfort and in all but the circumductions the subject passed through and paused at the neutral, upright posture. The stereograph outputs were recorded on a tape recorder for subsequent playback into a computer.

Table 2. Sequence of spinal movement instructions.

Terminology	Instruction
Neutral	Stand comfortably
Flexion	Flex fully forward
Circumduction	Move to your left rotating forward-left, left, backward-left, back, backward-right, right, forward-right, forward
Extension	Lean back
L. Lateral Bend	Lean left
R. Lateral Bend	Lean right
Circumduction	Flex forward and then move to your right rotating forward-right, right, etc to forward position
Neutral	Stand upright comfortably

Recording of pressure and temperature data

Intra-abdominal pressures and skin temperatures were recorded continuously throughout a sequence of activities. In order that the subject remained free to carry out these activities a portable tape recorder (Oxford Medical Systems "Medilog") was used. This was a four-channel cassette device, three channels being used for data recording as detailed below, and a fourth recording a voltage step change, controlled by a switch on the recorder and used as an event marker.

Intra-abdominal pressure

Previous studies of the pressure within the abdominal cavity have investigated both muscle-induced and hydrostatic pressure changes at various levels. Because the abdominal cavity is largely fluid-filled, hydrostatic pressure differences exist which are consistent with an average fluid density similar to that of water (Rushmer, 1946). Previous measurements of active abdominal pressure have been made in both the stomach and rectum, a constant hydrostatic difference being noted between the two measurements during each activity (Adno, 1956). The technique adopted in this study was measurement in the rectum by means of a catheter with a strain gauged pressure transducer at its tip (Gaeltec type 3EA/ICT). The catheter was inserted to a distance of about 15 cm, to ensure that the tip was within the abdominal cavity.

Maximum baseline drift of pressure measurement was about 1 mm Hg per test, but sensitivity was unaltered. Calibration was performed before each test by means of a graduated water column.

Skin temperature

To assess the warming effects of each support, skin temperatures were measured at two sites, one in the lumbar region directly under the support, and a second in the thoracic region well above the area of influence of the support and used as a control. Thermistors (Oxford Medical Systems type YS1) were taped to the skin in these areas, porous tape being used to minimize its influence upon the local skin temperature. Recording on to the cassette was via amplifiers, set to cover temperature ranges of 26°C–37°C (lumbar) and 21°C–36°C (thoracic). Calibration was by means of a thermostatically controlled water bath and negligible drift was observed.

Experimental procedure

The subject was instrumented for temperature and pressure recording. Once reclothed and acclimatized to the feel of the measurement apparatus he performed a sequence of timed activities (Table 3), lasting approximately 20 minutes, once without a support and then again while wearing a variety of lumbar spinal supports. A spinal movement study was performed under each condition.

Table 3. Sequence of timed activities.

Symbol	Activity	Time
L	Lying supine Standing from lying	3 min
W	Walking on a horizontal surface Ascending and descending stairs	2 min
S	Sitting in an upright high-backed chair Lifting 5kg between high and low shelves	2 min
FL	Lifting 15kg with legs straight, flexing at hips	
UL	Lifting 15kg with torso upright, flexing knees and hips	
H	Holding 15kg while standing upright Lifting 15kg from the side	30 sec
Then S, W, and L repeated in that order		6 min

Each subject underwent a single measurement session. This was important to ensure that results from each condition would be most readily comparable. Unfortunately, this meant that a long session was required in order that a complete set of measurements could be made from one subject wearing all of the supports in turn. For reasons of time, only four out of the "normal" group completed such a session. In the case of the patient group each subject, in addition to being measured without a support, was measured while wearing his prescribed support (usually a rigid brace) if a suitable one could be found. It was found that patients could not readily tolerate more than three corset tests, each of which took about 20 minutes. Two patients did not take part in the movement studies.

Data analysis

Movements

The recorded vector stereograph readings were played back into a computer file through an analogue to digital converter, and transformed into Cartesian coordinates. A further program used the data to plot out the movements of the top of the lumbar spine, as seen from above, also calculating the mean area enclosed by the circumduction manoeuvres (Fig. 2). The scale of the plot being known, any further measurements could be made by hand from the locus of movement.

Temperature and pressures

The cassette recordings were stored on computer file for analysis. The event marker channel was used to identify periods of controlled activity and for each of them print out the length of time the event took, the mean intra-abdominal pressure, the standard deviation, the maximum and minimum pressures. Over the combined periods of activity the mean, maximum and minimum temperatures at the lumbar and thoracic levels were also calculated and printed out. A further program displayed the pressure pattern from any selected event on a visual display unit and the operator then identified special features within the activity, such as pressure peaks on lifting. For each of these sub-events the time, mean, maximum and minimum pressures were printed out.

Averaged results are used throughout this report and where the effects of spinal support or posture within a group are examined the mean of the differences is given. This enabled the paired 't' statistical test to be used. Where results from different groups were being compared the difference of the means was used and in this case the straightforward 't' test was adopted.

Results

Table 4 gives complete averaged results from both normal and patient groups and results from a further spinal movement study.

Lumbar spinal movements

Three measurements of the movement locus were made to characterize each pattern; the total lateral range (L), the total flexion/extension range (F/E), and the square root of the mean area enclosed by the circumduction loci (A). The first two define the movement in specific directions, while the latter is a linear measure of overall functional range (Fig. 2). To establish the repeatability of the technique adopted, 15 normal male subjects were measured while

wearing no support. Each subject was measured four times in all on separate days. The scatter of results showed variations of less than 10%

Because the linear ranges of movement are dependent upon the anatomical dimensions of the subject as well as the support being worn, the measurements for each subject were normalized by dividing them by the corresponding value when no support was worn. Figure 2 illustrates the averaged, normalized results from the normal group; the smaller the bar the greater the restraint imposed by the corresponding support.

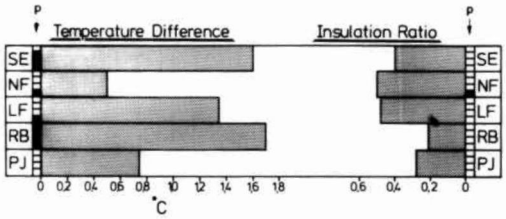

Fig. 3. Mean skin temperature results normalized by subtraction of "no support" figures.

Skin temperatures

It was assumed that for the purposes of analyzing skin temperature results all subjects could be taken as a single group, low back pain not being associated with local rise in skin temperature. To determine the degree of warming in each test the difference between the mean lumbar and thoracic skin temperatures was recorded (TD), being positive with a higher lumbar temperature. The thoracic skin temperature served to allow for the effects of ambient conditions upon the skin. By comparing for each individual the differences from each test involving a support with that from the test when no support was worn, the influence of the supports upon lumbar skin temperature could be seen. Figure 3 illustrates the mean of the temperature differences after subtraction of the corresponding value when no support was worn. The longer the bar the greater the warming effect. Figure 3 also illustrates the means of the differences in a factor termed "insulation ratio" (IR). This is a measure of the relative extent to which the lumbar skin is insulated from temperature changes seen at the thoracic level. It was calculated for each test by dividing the range of temperatures measured at the lumbar level by the range of temperatures measured at the thoracic level (IR). Thus if both had the same range the ratio would be unity and the lumbar skin would not have been insulated at all. The mean insulation ratio when no support was worn was greater than unity (see Table 4) and this indicates that in normal clothing the lumbar skin saw more temperature variation than the thoracic over the sequence of activities. However, the corresponding mean temperature difference was zero and this indicates that the two skin temperatures were not consistently different. In Figure 3 the values illustrated are averages of the insulation ratios after the subtraction of the corresponding value derived from the tests when no support was worn. The longer the bar, the greater the insulation.

Intra-abdominal pressures

There are two possible ways in which the supportive influence of the intra-abdominal pressure could be affected by a spinal support. The first is passive and involves the increase of resting or mean pressure levels, and the second is active and involves the modification of the reflex pressurization when the spine is put under mechanical stress. For the purposes of examining the passive influence four activities were selected from the sequence performed by each subject, these being lying, sitting, walking and holding a weight whilst standing upright (L, S, W, and H respectively). For each subject the mean intra-abdominal pressure during each of these activities in each test was determined. The averaged results are given in Table 4.

In order to examine the influence of supports upon the reflex abdominal pressurization two types of lift were examined, these being lifting with legs straight, bending at the hips, and lifting with an upright torso whilst flexing the knees and hips (FL and UL). From the results from each subject each time they performed these lifts, the peak pressures recorded were noted. The average values appear in Table 4.

The averaged results from the two groups are given in Figure 4. In this figure both results from wearing no support and the averaged effects of spinal supports are given. The longer the bars the higher the pressures, the dotted lines being the results from the recordings made while spinal supports were being worn. Where significant, other results are also listed.

Table 4. Summary of averaged results.

			Normal Group						Patient Group				Trial	
			NS	SE	NF	LF	RB	PJ	NS	NF	LF	RB	NS	
Spinal Movement (cm)	$\sqrt{A}$	m	12·6	10·0	9·0	10·6	6·0	4·4	7·4	7·8	9·0	6·0	12·7	
		sd	3·7	3·5	2·8	1·2	1·4	1·1	4·6	2·9	3·2	2·0	2·2	
	L	m	11·7	10·6	8·0	9·5	6·0	4·9	9·2	8·0	9·5	6·7	13·9	
		sd	3·4	4·0	2·8	2·8	1·1	1·1	3·4	3·1	0·7	2·3	2·9	
	F/E	m	19·3	14·4	16·8	17·5	11·0	6·8	14·4	14·0	14·0	9·5	21·3	
		sd	5·5	3·1	4·8	0·7	4·4	1·5	5·4	5·6	7·1	4·1	2·6	
		n	10	9	6	2	9	5	6	4	2	3	15	
Intra-Abdominal Pressure (mm Mercury)	L	m	16·9	19·3	20·0	19·5	19·5	19·4	8·0	16·0	11·0	17·0	Mean	
		sd	13·6	16·8	15·9	16·2	15·1	16·7	2·5	5·5	5·7	10·0		
	S	m	28·7	32·6	32·6	36·0	33·9	35·5	24·0	28·0	25·0	32·0		
		sd	13·5	17·3	19·8	20·7	16·4	19·3	5·0	8·0	7·8	5·5		
	W	m	29·0	34·3	32·3	32·3	33·5	33·8	24·0	26·0	24·0	33·0		
		sd	14·1	16·2	19·5	16·2	15·8	18·6	5·5	6·0	8·4	5·0		
	H	m	30·7	33·8	32·4	33·1	34·2	37·0	30·0	33·0	30·0	38·0		
		sd	17·4	18·0	19·7	16·2	17·4	20·5	6·0	9·0	6·4	11·0		
	FL	m	49·0	47·0	47·0	44·0	47·0	45·0	62·0	64·0	84·0	73·0	Peak	
		sd	24·0	22·0	29·0	15·0	17·0	19·0	33·0	28·0	41·0	26·0		
	UL	m	58·0	50·0	55·0	54·0	55·0	51·0	63·0	70·0	86·0	70·0		
		sd	25·0	21·0	21·0	12·0	18·0	26·0	21·0	22·0	13·0	24·0		
		n	10	10	8	6	10	7	8	5	2	6		
Skin Temperature (C°)	TD	m	0·0	1·24	0·56	1·3	1·46	0·50						
		sd	1·10	0·60	0·62	0·57	0·74	0·39						
	IR	m	1·15	0·73	0·72	0·67	0·69	0·69						
		sd	0·60	0·39	0·18	0·23	0·35	0·10						
		n	14	9	11	7	12	5						

Differences between supports

The results from the normal group are taken as representing the properties of the supports in terms of their effects upon an individual at first encounter and without the influence of pain. The latter influence is considered insignificant in the case of the surface temperature measurements. Therefore, when considering this question the results from the two groups are combined for the purposes of discussing the skin temperatures, whereas the results from the normal group are only discussed where intra-abdominal pressure and spinal movements are concerned.

The supports all warmed the lumbar skin, but those with added padding over this area warmed the most (Fig. 3). However, the greater the difference in temperature between the skin and the outside environment, the more difficult it was to maintain, and this was reflected in the insulation results (Fig. 3) which are much the same for all the supports.

The relationship between skin temperatures and symptom relief is an area which requires clarification. Dixon et al. (1972) assumed that the link between the wearing of a woolly belt and relief of low back pain in some cases was due to an increase in the lumbar skin temperature. It would be preferable to repeat their type of study using objective temperature measurement to seek to establish the mechanism by which such relief is provided.

All the lumbar spinal supports tested reduced the range of movement of the lumbar spine in the normal group. Again it was possible, to some extent, to predict their effects by considering their individual construction. The plastic jacket and rigid brace were different from the corsets which had little inherent rigidity and relied more upon elastic restraint. The rigid supports therefore restricted spinal movement considerably. The fabric based supports all gave less restriction, with the long fabric support having least effect (Fig. 2). It could be that the long corset had least influence upon movements because it bore upon rib cage and pelvis and gripped neither particularly strongly. The narrower supports, however, caused a "waisting" of the subject, thereby preventing movements by the impingement of the support's edges upon pelvis and rib cage. The influence of strategic stiffening of the corset can be seen by comparing the lateral and anteroposterior movements allowed by the elasticated and narrow fabric supports. The latter had stiffening members which were effective in reducing lateral movements while the former had a rigid frontal pad which restricted anteroposterior movement.

All the spinal supports raised the resting level of intra-abdominal pressure in all postures. Individually these increases were only significant in four instances; the elasticated support when walking, and the plastic jacket, long fabric and rigid brace when sitting (Fig. 4). However, there were no significant inter-support differences. The three supports which increased the pressure on sitting all extend between thorax and pelvis, while the one which increased pressure on walking was the only elasticated one tested. Taken as a whole, the supports had a significant effect in raising intra-abdominal pressure in the two postures of walking and sitting. Under normal circumstances without the presence of a spinal support, intra-abdominal pressure is generated by the compression of the abdominal cavity involving the diaphragm, pelvic floor, and muscles of the abdominal wall. Of these latter, transversus and the obliques are most active. By their very nature the action of the oblique muscles produces a disadvantageous mechanical moment which in part offsets that generated by the raised intra-abdominal pressure. Morris et al (1961) showed that a spinal support tends to raise resting pressure levels while reducing EMG activity in the abdominal wall. Thus the pressure increases measured do not act against a disadvantageous muscle activity. This makes them more effective at spinal load relief than those generated normally by muscle effort.

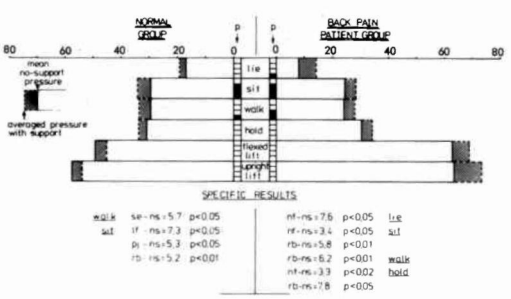

Fig. 4. Mean intra-abdominal pressure results.

During exercises when the intra-abdominal pressure reflex is excited, spinal supports had no significant effect upon peak pressure levels; however both individually and as a whole their tendency was to reduce the pressures. This may be as a result of some of the axial load being transmitted from pelvis to rib cage directly through the supports, thereby reducing the lumbar spinal load which is probably the stimulus for the reflex. It is surprising that the supports had such a uniform effect upon the intra-abdominal pressure considering the variety of types which were tested.

Differences between patients and controls

Low back pain causes reduced spinal mobility both as a direct result of discomfort and because of apprehension. The response of a low back pain patient to a spinal support is likely to differ widely from that of a normal person. Some subjects moved no more without their familiar support than when wearing it. Presumably the further a support keeps someone away from areas of painful movement the less pain and apprehension will affect the way they move. The subjects with pain had varying levels and types of painful movement so the results cannot be analyzed closely. Nevertheless, the mean effect of the long fabric support (PH) which proved least effective on the normal group was to have no influence on the mobility of the low back pain patients who wore it; pain still dominated their movement pattern (Table 4). The narrow fabric support (S) had some effect but the mean ranges of movement were still less than for the normal group, indicating a combined effect of corset and

pain. The rigid brace (G) had the most effect and the mean ranges of movement were similar to those for the normal group, indicating that the effect of the support was the dominating feature. Therefore, while the supports influenced the spinal movements of the patient group in a manner similar to the normal group, the effects were modified by the added influence of pain.

The intra-abdominal pressure results showed some less predictable differences. Considering first the mean pressure levels, it became apparent that the mean level when lying supine was 9 mm Hg higher in the normal group (Fig. 4). Since the pressure in this posture is largely a result of muscle tone in the abdominal wall, this observation indicated that the patient group had generally less active muscles in this region. The patients had all been wearing spinal supports regularly as part of the treatment for their low back pain. Morris et al, (1961) showed that a reduced abdominal muscle activity results when a corset is worn. It seems likely that, despite exercises, the long term effect upon the patient group of wearing corsets had been a loss of tone in the abdominal muscles. One patient only had a resting intra-abdominal pressure comparable with the normal group. He still maintained a physically demanding job and when questioned appeared well educated about how to use his spine in lifting, etc., and was complying with physiotherapist advice.

The spinal supports raised mean intra-abdominal pressures by similar amounts in the patient group when compared to the normal group. However, because of the lower pressures recorded when no support was worn, the effect of the supports was to raise the mean pressures only to those of the normal group without supports.

It was observed in this study, as elsewhere (Fairbank et al, 1980), that patients suffering from low back pain develop much higher reflex intra-abdominal pressures than pain-free controls. This is believed to be a response to back pain which attempts to protect the spine still further from load by increasing the load bearing role of the anterior compartments.

When wearing a spinal support the patient group tended to develop still higher pressures (7 mm Hg higher on average) compared with the normal group which tended to produce lower pressures (4 mm Hg on average). This indicated that the patient group was using the greater support an orthosis provides to the abdominal wall in order to increase the effect of the intra-abdominal pressure reflex and reduce further the mechanical stresses on the spine.

Conclusions

1. This study confirms that spinal supports influence the movement, intra-abdominal pressure and skin temperature of the wearer. Considering the wide variety of supports tested, the differences between the effects of each support are few. However, some characteristic patterns of effect, particularly in the spinal movement restriction, were found.

In order to reduce spinal movements by an appreciable amount, a rigid form of bracing is required, although a well fitting brace (RB) is better than a plastic shell (PJ) in this respect. Fabric and elasticated corsets provide little restriction of movement although the location of strengthening can enable specific painful movements to be influenced above others. The shorter corsets performed better than the longer in respect of movement restraint.

Where low back pain is temperature sensitive, the presence of thicker or padded material over the lumbar skin can be used to raise its temperature by almost 2°C. However, the material must be held in contact with the skin. Several subjects commented that the plastic jacket had a tendency to provide a cooling "funnel" which reduced its effectiveness in this respect.

No clear patterns emerge from the intra-abdominal pressure results, except that the longer supports provide significant increases in pressure when the wearer is seated, and the elasticated support increases the pressure significantly when the wearer is walking.

2. The patient group responded in a predictably different manner to the normal group in respect of spinal movement, but not of intra-abdominal pressure. The results suggest that over the period of treatment a patient becomes accustomed to his orthosis and subconsciously adopts it as part of his spinal support mechanism. Thus, under activities where the spine is lightly stressed, the presence of the support reduces the need for activity of the muscles of the abdominal wall. Under more stressful activity the orthosis strengthens the wall and enables the wearer to enhance the pressures developed during the intra-abdominal pressure

reflex. This indicates a need to plan the use of a spinal support in the context of other treatments, such as exercise regimes, especially when the patient ceases to wear his corset.

3. The study highlights the need to establish more clearly the mechanisms by which a spinal support acts upon the wearer and how the physical effects it induces are effective in providing symptom relief and a healing environment.

REFERENCES

ADNO, J. (1956). Some aspects of the mechanics of the abdomen. *South African Med. J.*, **30,** 535–539.

AHLGREN, S. A., and HANSEN, T. (1978). The use of lumbosacral corsets prescribed for low back pain. *Pros. Orth. Int.*, **2,** 101–104.

BARTELINK, D. L. (1957). The role of abdominal pressure in relieving the pressure on the lumbar intervertebral discs. *J. Bone Jt. Surg.*, **39B,** 718–725.

DAVIS, P. R., and TROUP, J. D. G. (1964). Pressures in the trunk cavities when pulling, pushing and lifting. *Erg.*, **7,** 465–474.

DIXON, A. St. John, OWEN-SMITH, B. D., HARRISON, R. A. (1972). Cold-sensitive, non specific, low back pain (a comparative trial of treatment). *Clin. Trials J.*, **9,** 16–21.

EIE, N., and WEHN, P. (1962). Measurements of the intra-abdominal pressure in relation to weight bearing of the lumbosacral spine. *J. Oslo City Hosp.* **12,** 205–217.

FAIRBANK, J. C. T., O'BRIEN, J. P., DAVIS, P. R. (1980). Intra-abdominal pressure rise during weight lifting as an objective measure of low back pain. *Spine,* **5,** 179–184.

GREW, N. D., and HARRIS, J. D. (1979). A method of measuring human body shape and movement—the "vector stereograph". *Eng. Med.*, **8,** 115–119.

GREW, N. D. (1980). Intra-abdominal pressure response to loads applied to the torso in normal subjects. *Spine,* **5,** 149–154.

KUMAR, S., and DAVIS, P. R. (1973). Lumbar vertebral innervation and intra-abdominal pressure. *J. Anat.*, **114,** 47–53.

MORRIS, J. R. W., and HARRIS, J. D. (1976). Three dimensional shapes (Letter). *Lancet,* 1189–1190.

MORRIS, J. M., LUCAS, D. B., BRESLER, B. (1961). Role of the trunk in stability of the spine. *J. Bone Jt. Surg.*, **43A,** 327–351

NORTON, P. L., and BROWN, T. (1957). The immobilising efficiency of back braces. *J. Bone Jt. Surg.*, **39A,** 111–139.

PERRY, J. (1970). The use of external support in the treatment of low back pain. *J. Bone Jt. Surg.*, **52A,** 1440–1442.

RUSHMER, R. F. (1946). The nature of intraperitoneal and intra-rectal pressure. *Amer. J. Physiol.*, **146,** 242–249.

WATERS, R. L., and MORRIS, J. M. (1970). Effect of spinal supports on the electrical activity of muscles of the trunk. *J. Bone Jt. Surg.*, **52A,** 51–60.

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Frame type socket for lower limb prostheses

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Abstract

The technique presented uses a frame-type socket for fixation of the prosthesis on the stump. Apart from rigid areas for stabilization and control, the soft tissue of the stump is enclosed by flexible material. This allows self-adjustment of the socket to changes in stump circumference and volume, while maintaining good socket fit.

Introduction

Prosthetic care for lower limb amputees presents a highly complicated problem particularly in the period immediately following the amputation. While still being shocked by the very fact of amputation and having to accept impairment of his visual appearance and the loss of essential body functions as a new fact of life, the patient develops individual expectations with regard to the future handling of his problem. To meet these expectations the prosthetist has to provide a prosthesis which not only re-establishes the lost body functions but also provides a visual and tactile appearance similar to the amputated limb. Much progress has been achieved in this context during the last years by the modern functional and cosmetic technology of modular prostheses. There still remains, however, a considerable deficiency in stump accommodation techniques.

Early prosthetic care is in the interest of all participants for functional rehabilitation. To this end there should be a controllable, undisturbed wound healing process and enclosure of the amputation stump into the most favourable mould which takes account of its anatomical and pathological circumstances. This latter requirement, in particular, is difficult to fulfil in the case of first or early prosthetic fitting, as postoperative changes in volume and circumference of the stump with enclosure or pressure dressings cannot be avoided (Fig. 1).

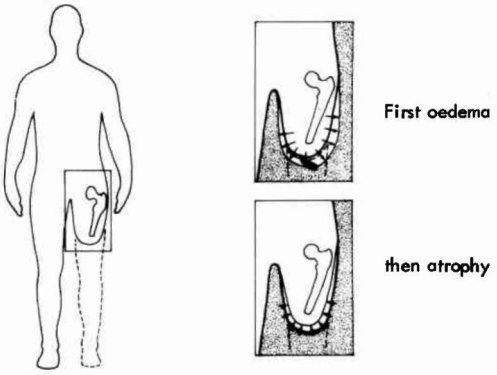

Fig. 1. Variation of volume in the amputation stump.

Consequently, the changes arising in the form and volume of the stump are not taken into account by the socket in conventional techniques. The functional value of a prosthesis being predominantly determined by the accurate matching of the socket to the stump, this problem is of particular importance.

In the techniques of suspension that are employed to date using suction or total contact sockets, the best, first or early prosthetic fitting can only be achieved by permanent correction (widening or narrowing) of the prosthetic socket (Fig. 2).

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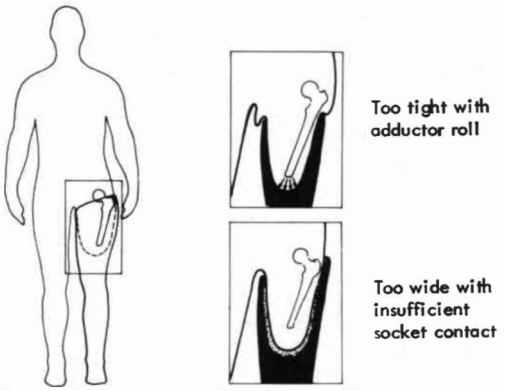

Fig. 2. Socket fitting deficiencies.

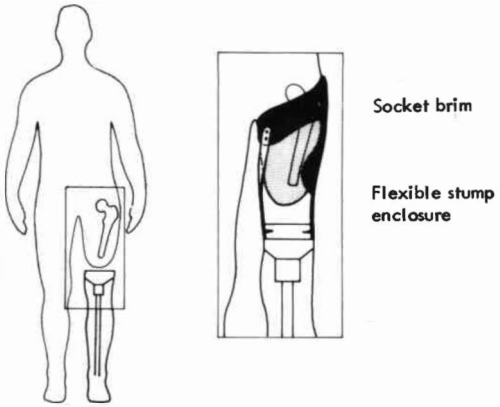

Fig. 3. Socket brim, femoral clasp and connection to prosthesis.

Some discrepancies can certainly be handled by belt systems on a short term basis. This, however, implies for the patient a continuous dependence on the orthopaedic technician up to the time when the amputation stump has attained its permanent shape.

Various efforts had been made in order to avoid this disadvantageous situation. Sockets have been developed, for instance, which compensate for stump volume changes by the use of inflatable air cushions (Kuhn, 1959; Biedermann, 1979).

This study reports an attempt to enclose the stump in a specially designed frame socket; the technique is described in some detail below.

Frame socket accommodation of above-knee and hip disarticulation stumps

Using the fact that the proximal part of the amputation stump exhibits the smallest variations in volume, in most cases a precise enclosure of the upper part can be attained by a closed proximal socket brim (Fig. 3). From this brim the load-bearing framework with its struts of elastic material extends distally to form the connection with the knee part of the prosthesis. The soft tissue, with its disposition to circumferential variations, is largely bypassed by this technique.

Several modifications of frame construction have been tested. A three-sided soft tissue release towards the ventral, medial, and dorsal direction has been preferred as it offers advantages both for the patient's comfort and the prosthetic fitting (Fig. 4). Moreover, criteria were considered which, at the present time, are estimated to be most advantageous for the accommodation of the above-knee stump and which have been excellently realised in the total contact socket (Hepp, 1948; Kuhn, 1963–68) with terminal contact, femoral clasp, and proximal brim (Fig. 5).

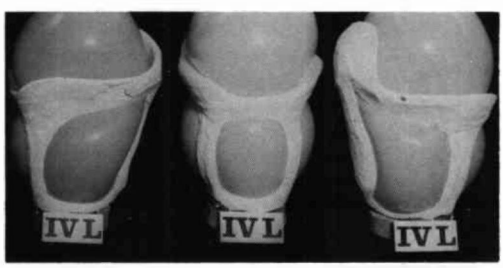

Fig. 4. Plaster cast frame socket, ventral, medial and dorsal views.

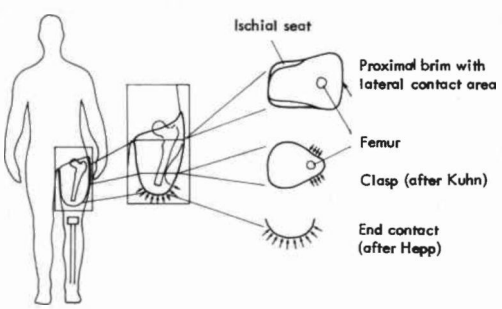

Fig. 5. Special properties of the total contact socket.

Taking into account these particular accommodation properties, there is in the frame socket, in the region of the fascia lata, a wide strut with the femoral clasp extending distally to the knee part of the prosthesis. Depending on the properties of the stump, the desired terminal contact is obtained by elastic fabric, a foam material cushion, or solid materials. The soft tissue (which is particularly subject to volume variations) protruding from the ventral, medial, and dorsal frame openings is accommodated in two different ways by elastic compression (Fig. 6). The elastic enclosure of these parts of the stump, which are not compressed by solid material, is necessary in order to avoid oedematous swellings.

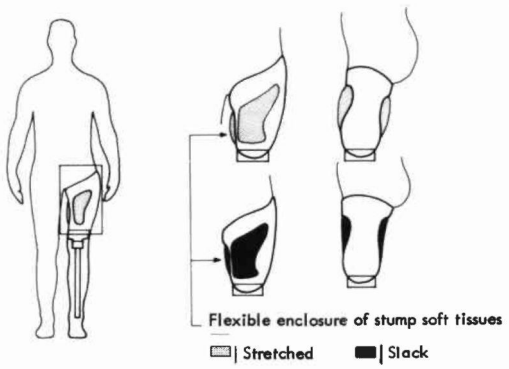

Fig. 6. Frame type socket demonstrating flexible enclosure of stump soft tissues.

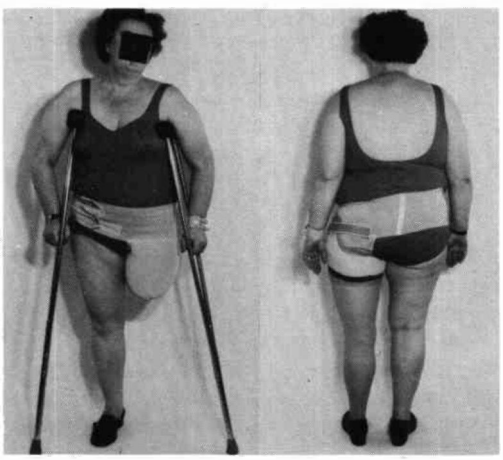

Fig. 7. Left, patient with stump compression bandage. Right, same patient wearing prosthesis with frame socket.

The first procedure comprises the compression of the stump by an elastic stocking which can be put on comparatively easily with the aid of a lateral zip-fastener (Fig. 7). Donning of this compression stocking is facilitated by an understocking. A lateral part which runs along the thigh and is fixed to the pelvic region guarantees a perfect fitting even after prolonged wear and operation. The prosthesis with the frame socket is put on in the conventional way over this compression stocking which completely sheathes the amputation stump (Fig. 8).

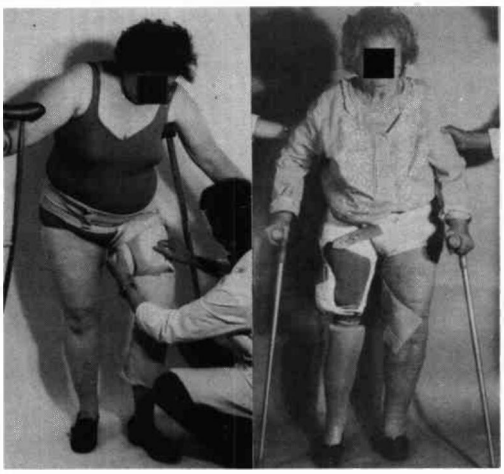

Fig. 8. Patients with frame socket prostheses. Left, resin cast, right, plaster cast.

In the second approach the system is modified so as to first introduce the stump into the frame socket as usual. The soft parts of the stump which protrude from the openings are elastically enclosed by a double traction stocking which is fixed at the upper socket rim and is drawn downward over the outside of the socket (Fig. 9). This technique can only be employed in frame sockets with thin walls, however, as there would otherwise be a discontinuity in the uniform enclosure. The prosthetic cosmesis which is pulled up after donning the prosthesis is fixed by a Velcro fastener to the distal region of the frame socket.

The two procedures outlined above enable the sheathing of most stump volumes and a consequent self regulation of variations in volume and circumference by the elastic framing and enclosure of the soft tissue of the stump.

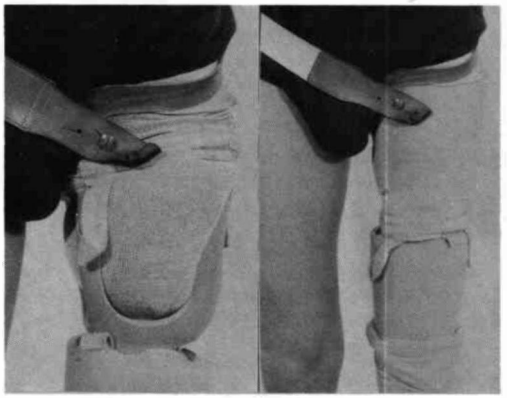

Fig. 9. Frame socket enclosure of an above-knee stump by an outer compression bandage.

Moreover, the patient is able to put on his prosthesis without problems as the openings in the frame socket allow a manual and tactile contact with the stump surface and hence accommodation of the stump into the socket without strain.

As a result of favourable experiences in the care of above-knee amputees this techniques was used also for the construction of hip disarticulation prostheses. To this end the prosthesis is fixed at the pelvic stump by a miniature pelvic basket (Fig. 10), while the voluminous soft parts are enclosed by a slightly or moderately compressing TIGGES* bandage without however using a back pad (Fig. 11). Apart from guaranteeing good functioning of the prosthesis it is more comfortable for the patient as a result of the elastic control of the device.

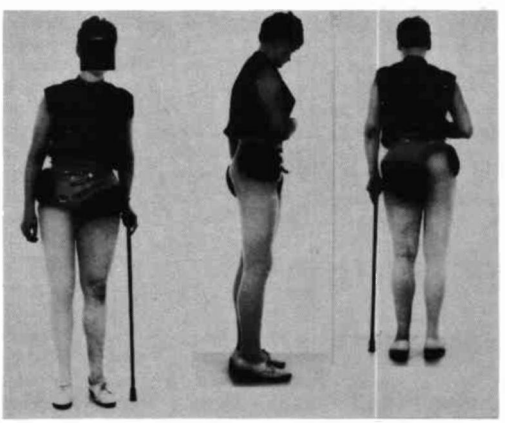

Fig. 10. Hip disarticulation prosthesis with miniature pelvic basket.

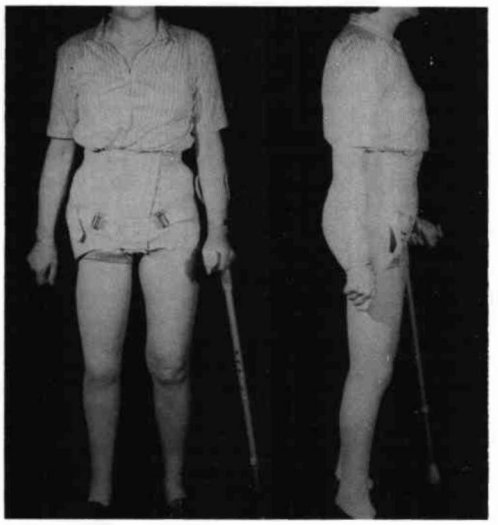

Fig. 11. Hip disarticulation prosthesis with miniature pelvic basket and TIGGES bandage applied.

There are no essentially new guidelines for the experienced orthopaedic technician to be considered in the construction of this type of prosthesis, as the frame socket is fixed to the prosthesis modules in the same way as conventional sockets.

The frame socket itself is designed on the basis of an individual plaster mould of the amputation stump that is converted to a plaster positive. This positive serves for the construction of a plastic socket, which is then provided with the necessary openings to yield the desired frame pattern. The completed frame socket is fixed in an appropriate way to the prosthesis modules.

In the course of production of the frame socket prosthesis the adaptation of the frame socket to the prosthesis by means of adjustment devices is an optional possibility for optimization of the system.

Results

A total of 23 patients were fitted with the new frame socket technique (14 above-knee amputees and 9 hip disarticulation patients). The patients unanimously rated the tactile contact between the amputation stump and the prosthesis and the visual appearance of the system as very agreeable. They reported no difficulties in the handling of the device. The

*Product of Otto Zours KG,
Postfach 480,
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load bearing sensation in the frame socket due to its elastically supporting apertures, was unanimously reported to be most comfortable and free of any problems. The static and dynamic stability of the patients was in no case estimated as being inferior to that of patients provided with conventional devices. The new frame socket technique has also been applied to patients who had complained about their previous closed socket prostheses; these patients are now free of complaints.

Discussion

The technique of amputation stump accommodation described in this paper is a practicable alternative in the early care of amputees which offers the possibility of compensating for variations in stump circumference and volume. This is particularly necessary in the care of patients after tumour-indicated amputation as considerable variations in body weight and hence in stump volume are to be expected during the prolonged period of chemotherapy which in most cases follows the operation. Moreover, the elastic enclosure of the soft stump parts yields space for muscle contractions, avoids unnecessary perspiration and provides for ventilation of the amputation stump which adds to the patient's comfort. The large range of possible variations in the stump enclosure given by this technique enables early independence from the orthopaedic technician coupled with an optimum prosthetic fitting and adaptation. The correct donning and use of the prosthesis is simplified for the patient making a fast functional rehabilitation possible.

REFERENCES

BIEDERMANN, W. G. von, and HOLTKAMP, H. (1979). Erfahrungen mit einem pneumatischen Einbettungssystem für die technische Orthopädie. *Orthopädie-Technik,* **7(79),** 109–112.

HEPP, O. (1948). Haftprothesen. *Zeitschrift für Orthopädie.* **77,** 219–279.

KUHN, G. (1959). Personal communication.

KUHN, G. (1963–1968). Personal communication.

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Variation of mechanical energy levels for normal and prosthetic gait

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Abstract

Mechanical energy levels were investigated for normals and for below-knee amputees during level walking. The weight of the prostheses was varied by attaching 0·5 kg extra weight to the prostheses.

The measurements and analyses were made with the ENOCH system consisting of a minicomputer (HP 21 MX), an optoelectronic device for displacement data measurement (Selspot) and a force plate (Kistler) for measurement of ground reaction forces.

Results by Winter et al (1976) on the energy changes during normal walking obtained from displacement data on one leg only were verified using data from both legs and the trunk.

For the amputees it was concluded that the energy changes increased for the prosthetic shank when the weight increased. For the other body segments and for the body total no significant differences were found.

Introduction

The energy expenditure during walking is an important parameter for the evaluation of human gait. Ralston and Lukin (1969) found a fairly constant ratio between the metabolic expenditure and the positive work for subjects during normal walking and walking with extra trunk and foot loading. Therefore the mechanical energy changes can be used to get an indication of the energy requirements during walking. They defined the positive work from the changes in total mechanical (kinetic plus potential) energy of the body.

In recent years the mechanical energy levels during walking have been studied by Winter et al (1976) and Cappozzo et al (1976) among others. In both these investigations displacement data for one leg were used to describe the motion of the whole body.

For the construction of lower extremity prostheses the weight is an important design parameter. The optimum weight of the prostheses have been a subject of debate.

Inman (1967) claimed that a prosthesis should not be made too light because with a lighter prosthesis the amputee develops less kinetic energy at the end of swing phase to be fed back into the body to maintain his forward velocity.

Quigley et al (1977) reported on the oxygen consumption during walking with ultra-lightweight and standard BK prostheses. The trend was toward a higher oxygen consumption per metre and kilogram body weight with the heavier type of prosthesis.

The investigation presented in this paper was undertaken to test the effect of a small increase in the weight of BK prostheses on the mechanical energy levels during walking. Gait data from both legs and the HAT (head, arms, trunk) for normals were also analyzed to test the results of Winter et al (1976) and Cappozzo et al (1976).

Patients and methods

A minicomputer based system—called ENOCH—was used for the measurements and analyses. In this system, described by Gustafsson and Lanshammar (1977), an optoelectronic device, Selspot, is used for kinematic data collection and ground reaction data are obtained

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from a Kistler force plate. Output of result diagrams are made on a graphic computer terminal with a hardcopy unit or in tabular form on a line printer.

Two Selspot cameras were used to obtain kinematic data for both legs. Landmarks (light emitting diodes) were placed on the shoulder, hip joint, knee joint, ankle joint, heel and toe base for both sides. The measurement area was approximately 3 × 3 metre allowing for the registration of 3 steps in each measurement. Data was collected at the rate of 158 Hz. The standard deviation of the measurement noise was 0·002 m and the systematic coordinate error was estimated to be less than 0·02 m after correction of errors related to floor reflections, lens distortion and detector characteristics (Gustafsson & Lanshammar, 1977).

The displacements of the centre of mass for the different body segments in the model, HAT, thighs and shanks, were calculated from the measured coordinate data. The required body segment parameters were obtained according to the method described in Gustafsson and Lanshammar (1977) which is based on data from Drillis and Contini (1966), Contini (1970), Contini (1972) and Chandler et al. (1972). The velocities of the different body segments were calculated by numerical differentiation of the displacement data with a method described in Gustafsson and Lanshammar (1977).

Angular velocities for the body segments were obtained from the linear displacement and velocities by straightforward application of trigonometric relations.

The energy levels for the body segments were calculated as

$$ET_i = 0{\cdot}5 \cdot m_i \cdot v_i^2 + 0{\cdot}5 \cdot j_i \cdot w_i^2 + m_i \cdot g \cdot h_i$$

where m_i is the segment mass, j_i is the rotational moment of inertia about the centre of mass, v_i is the linear velocity, w_i is the angular velocity and h_i is the vertical coordinate for the mass centre of the ith segment.

It should be noted that only planar motion was included in the analysis. Of course this restricts the validity of the results, especially for pathological gait. Further the shank and foot was treated as one rigid body. The total body energy was calculated as the sum of the energy levels of the five body segments.

Measurements were made on three male persons with below-knee prostheses and on two male normals. Data on the subjects are given in Table 1. All the amputees were using TPJ prostheses. The TPJ (Torsten Pettersson Jigg) system is a Swedish method for fabrication of light-weight BK prostheses.

For the amputees gait measurements were made with their normal prostheses but also when 0·5 kg extra weight had been applied to the prostheses. The extra weight was in the form of lead plates attached with a belt around the centre of mass of the prosthesis.

Table 1. Subject data

SUBJECT	SEX	AGE	WEIGHT (kg)	HEIGHT (M)	PROSTHESIS	NO. OF MEASUREMENTS	
						WITHOUT EXTRA WEIGHT	0.5 kg EXTRA WEIGHT ON THE PROSTHESIS
JK	M	46	55.5	1.68	TPJ (1.52 kg)	7	8
EL	M	41	74.5	1.77	TPJ (1.45 kg)	8	8
DR	M	26	87.5	1.77	TPJ (1.38 kg)	6	2
HL	M	29	72.5	1.91	NORMAL	1	
KÖ	M	34	73.1	1.84	NORMAL	1	

Before the first measurement the subjects were allowed approximately 15 minutes of gait training to get used to the new prosthetic weight.

The subjects were allowed to choose their own comfortable step rate. It was approximately 75 steps/min for JK, 105 steps/min for EL, 90 steps/min for DR, 90 steps/min for HL and 95 steps/min for KÖ. A metronome was used to help the subjects keep a constant step rate during all the measurements. The stride lengths were approximately 1·30 m for JK, 1·50 m for EL, 1·25 m for DR, 1·50 m for HL and 1·65 m for KÖ.

Errors

The most important error contribution to the calculated energy levels is the segment masses. According to Gustafsson and Lanshammar (1977), the standard deviation of these errors is approximately 10 per cent. Compared to this figure the errors in the kinematic variables are very small during most of the gait cycle. The standard deviation of the stochastic error in the linear velocities was calculated to be less than 0·01 m/s (Gustafsson & Lanshammar, 1977), which is below 1 per cent of the mean velocity. The systematic error in linear velocities due to the limited bandwidth of the differentiating filter is also very small during most of the gait cycle.

However, close to heel strike the coordinates have high frequency components which can result in substantial errors in calculated derivatives. This can be the reason for the large errors reported in Cappozzo et al (1976) concerning a comparison between the muscular work and the total body energy variations during the heel strike phase.

The rotational component of the energy is much smaller than the translational and therefore the error in this term is not so important. In the term for the potential energy finally, the error in the vertical coordinates of the centres of mass are at most a few centimetres, which again gives much smaller error contribution as compared to the 10 per cent expected error in the segment masses. Note especially that for the HAT, an error in the location of the mass centre will not effect the changes of its potential energy values very much. The only effect of such an error is that the level around which the changes take place will be increased or decreased.

To summarize, the expected error in the calculated energy levels is approximately 10 per cent except during the heel strike phase where the error can be larger.

When energy changes are calculated by subtraction of energy levels the situation is changed, especially for the potential energy of the trunk. Since the vertical displacement of the centre of mass for the HAT is only about 0·05 m, a coordinate error of 0·01 m due to skin movements for example, can result in a 20 per cent error in the potential energy change. For the kinetic energy of the HAT and for both kinetic and potential energy of the other body segments the relative energy changes are much larger and the results for energy levels are essentially applicable also for the energy changes.

Results

In Figures 1–4 statistics for the resulting energy changes are presented. The values for the shank and thigh represent the difference between the maximum energy level during swing phase and the minimal energy level found for the following stance phase.

For the HAT and the total energies, the values are the difference between the maximum energy value during the first measured swing phase and the minimum value that is obtained close to the next heel strike, plus the same difference for the following step.

All energy values have been normalized by dividing with the current stride length and the subject's body weight. The middle line of the three horizontal lines on top of each bar represents mean value and the two others give the standard deviation for all measurements on the subject. For comparison the corresponding results for two male normals are also given in the figures together with results taken from Winter et al (1976). In the latter case the subject was a 20 year old female (subject 1 in Winter et al, 1976). From the Figures 1–4 it is clear that there is a good correspondence in the energy changes for all segments between the normal data presented here and the results obtained by Winter et al (1976). The results presented in Cappozzo et al (1976) can be compared only for the HAT and total body energy changes, because shank and thigh energy values are not presented separately. For the HAT the energy changes for one normal subject (U.D. in Cappozzo et al, 1976) corresponded to 0·1 J/m/kg, while it was approximately 0·2 J/m/kg for the data presented in Figure 3. The total body energy changes in Cappozzo et al (1976) for the subject U.D. was 0·4 J/m/kg which concords well with the results in Figure 4.

The energy changes reported in Winter et al (1976) and in Cappozzo et al (1976) were obtained using some simplifying assumptions. Data was collected for one leg only while the trajectory of the other leg was assumed to be the same displaced in time by half a stride period. Further the trajectory of the centre of mass of the trunk was assumed to be the same as the average of the trajectories of the greater trochanter marker for the left and right leg. Rotation of the trunk was not included in the analysis.

Especially the first of these simplifications could be expected to give considerable errors in the total body energy variations. This is because the shank energy is rapidly decreasing prior to heel strike while it is rapidly increasing shortly thereafter for the other leg. Therefore the minimum value of the total body energy, which occurs shortly before heel strike, can be heavily affected if the assumed time displacement in the data for the two legs are erroneous.

In the present investigation none of the mentioned simplifications were made.

Shank

The energy changes found for the shank on the normal leg of the amputees were comparable to those obtained for the normals (Figure 1). For the prosthetic side however the energy changes are considerably lower for the amputees. The reason for this difference is that the weight of the prosthesis is much lower than the weight of the normal leg. Since the displacements of the prosthetic shank were approximately the same as those for the normal side this resulted in a smaller peak energy value during swing and hence a smaller energy change was obtained.

With 0·5 kg extra weight on the prosthesis the energy changes on the prosthetic shank were increasing significantly for all subjects. This increase was approximately proportional to the weight increase, which is what to expect when the gait pattern is not changing. For the normal leg no significant change was found.

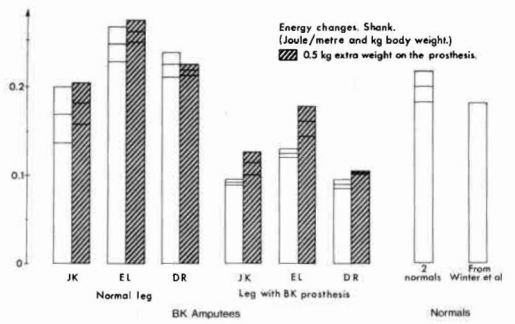

Fig. 1. Energy changes for the shank. The horizontal lines on each bar indicate mean value and standard deviation. The number of measurements involved is given in Table 1.

Thigh

The magnitude of the energy changes for the thighs were comparable between the amputees and the normals (Figure 2). This applies for both legs of the amputees.

When 0·5 kg extra weight was applied on the prostheses no statistically significant changes were found. For two of the subjects the values decreased slightly while they increased for one subject.

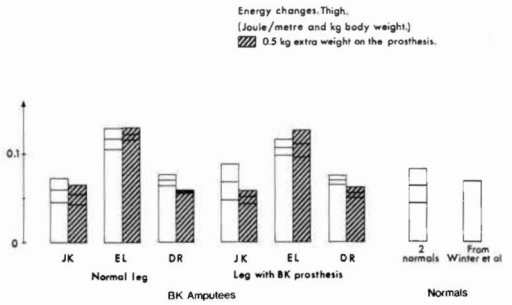

Fig. 2. Energy changes for the thigh.

HAT

Also for the HAT the energy changes for the amputees were close to those obtained for normals (Figure 3). The values did not change significantly when 0·5 kg extra weight was added to the prostheses.

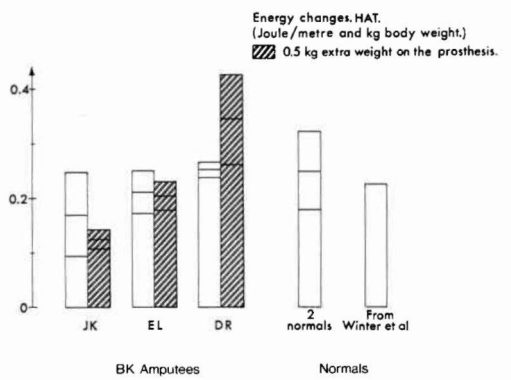

Fig. 3. Energy changes for the HAT (head, arms, trunk).

Total energy changes

The total energy changes are illustrated by Figure 4. Since the maximum values of the energy levels for the different body segments do not coincide in time the changes in the total energy level were always less than the sum of the segments energy changes.

In the total energy changes no significant difference was found when 0·5 kg extra weight was added. The values for the normals were also comparable to those for the amputees.

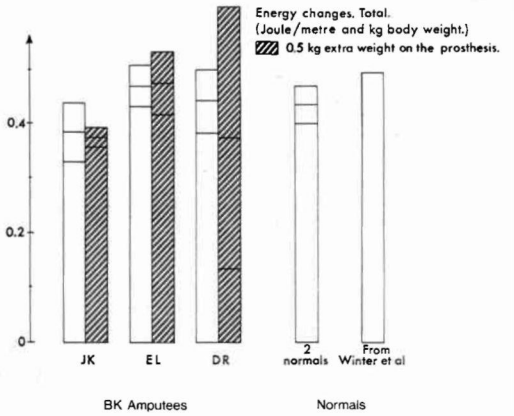

Fig. 4. Energy changes for the whole body.

Discussion

When 0·5 kg extra weight was added to the prostheses it was found that the energy changes for the prosthetic shank increased significantly for all subjects. For the other segments and for the total energy changes no significant effect of the extra weight was found.

Suppose that the energy changes of the prosthetic shank increase by the amount we have observed as a response to the extra weight, while the energy changes of the other segments are not influenced at all. Then there would be an increase in the total body energy changes, even if it would not be large enough to be statistically certified for the limited number of trials in this investigation.

However from our experiments with varying prosthetic weight there is not even a slight trend in the results on total body energy changes. It remains to verify if this result is purely coincidental, an effect of the stochastic nature of the human gait, or if there are any compensatory reactions to the increased prosthetic weight which make the total body energy changes rather insensitive to such weight changes.

Obviously the results presented here on the total body energy changes are not consistent with the results of Quigley et al (1977) who reported a higher oxygen consumption with a heavier type of prosthesis. Further research will be required to decide whether the above mentioned results of Ralston and Lukin (1969) do not apply to the present case, or if there is some other explanation of this discrepancy.

Conclusion

The results presented in Winter et al (1976) on energy changes during normal locomotion were verified using gait data for both legs and for the HAT (head, arm, trunk). In contrast to this Winter et al (1976) as well Cappozzo et al (1976) used displacement data for one leg only.

The results from Cappozzo et al (1976) were verified for the total body energy changes, while the values given for the HAT in Cappozzo et al (1976) were about half of those in the present study. For the shank and thigh no data was given in Cappozzo et al (1976).

Energy changes for BK amputees were also studied. It was concluded that for all body segments except the prosthetic shank, the results were quite similar to those obtained for normals. For the prosthetic shank the energy changes were much smaller than for a normal shank. This is explained by the light weight of the BK prosthesis as compared to a normal leg while the trajectories of the prosthesis were similar to those of the normal shank.

When the weight of the prostheses was increased by 0·5 kg the energy changes for the prosthetic shanks increased significantly. For the other body segments and for the total body no significant changes were noted.

Acknowledgements

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REFERENCES

CAPPOZZO, A., FIGURA, F., MARCHETTI, M. (1976). The interplay of muscular and external forces in human ambulation. *J. Biomechanics,* **9(1),** 35–43.

CHANDLER, R. F., CLAUSER, C. E., MCCONVILLE, J. T., REYNOLDS, H. M., YOUNG, J. W. (1975). Investigation of inertial properties of the human body. DOT HS-801 430.

CONTINI, R. (1970). Body segment parameters (Pathological). Technical Report No 1584.03. School of Engineering and Science, University of New York.

CONTINI, R. (1972). Body segment parameters, Part II. *Art. Limbs,* **16(1),** 1–19.

DRILLIS, R., CONTINI, R. (1966). Body segment parameters. Technical Report 1166.03. School of Engineering and Science, University of New York.

GUSTAFSSON, L., LANSHAMMAR, H. (1977). ENOCH—An integrated system for measurement and analysis of human gait. UPTEC 7723R. Institute of Technology. Uppsala University, Uppsala, Sweden.

INMAN, V. T. (1967). Conservation of energy in ambulation. *Arch. Phys. Med. Rehab.* **48(9),** 484–488.

QUIGLEY, M. J., IRONS, G. P., DONALDSON, N. R. (1977). The Rancho ultralight below-knee prosthesis. Rehabilitation Engineering Center at Rancho Los Amigos Hospital, Downey, California.

RALSTON, H. J., LUKIN, L. (1969). Energy levels of human body segments during level walking. *Ergonomics,* **12(1),** 39–46.

WINTER, D. A., QUANBURY, A. O., REIMER, G. D. (1976). Analysis of instantaneous energy of normal gait. *J. Biomechanics,* **9(4),** 253–257.

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Technical note—auditory feedback of knee angle for amputees

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Abstract

A new gait training device has been developed to provide auditory feedback of knee angle information to above-knee and hip disarticulation amputees. Traditionally, new amputees have relied on visual feedback of knee position during gait training (van Griethuysen, 1979). This auditory feedback system eliminates the need for visual feedback by providing a frequency encoded tone corresponding to knee angle.

Operation

The system is designed to produce a tone which increases in frequency as the knee flexes. When the knee is fully extended there is an absence of sound. As the knee is slightly flexed (less than 2°) a tone of approximately 60 Hz is generated. This frequency increases linearly with knee flexion up to 600 Hz which corresponds to the knee being flexed to 70°. With flexion over 70° the tone is absent so as to allow the patient to sit without having to turn off the unit.

During the swing phase of gait there is constant auditory feedback which can be beneficial in establishing an even, repeatable cadence. In addition, the absence of sound at full extension indicates that it is safe for the amputee to place the heel down.

The electronic operation of the device is quite simple and is outlined in Figure 1. The voltage from the potentiometer is amplified and then inputted to a voltage comparator and also to a voltage controlled oscillator. If the voltage corresponds to an angle between 2° and 70° the voltage controlled oscillator is enabled and produces a tone.

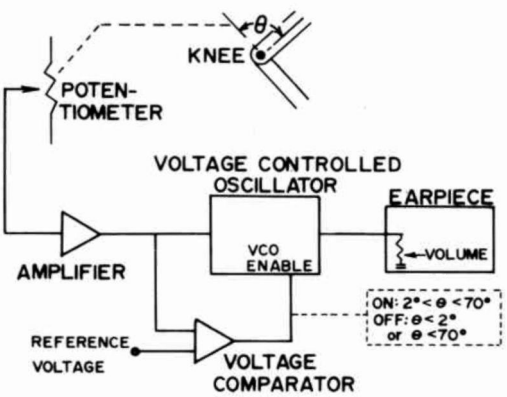

Fig. 1. The electronics of the angular feedback device.

Description

The knee angle feedback system consists of a single turn potentiometer with a miniature cable connecting it to a small belt box containing the electronics and a loudspeaker. As an alternative to the loudspeaker, a modified hearing aid earpiece can be used.

The potentiometer can be mounted either directly at the pivot point on a prosthesis (Fig. 2) or between two aluminium brackets which strap above and below the prosthetic knee (Fig. 3). The attachment method is used only on prostheses that have been fitted with cosmetic covers.

Discussion

This angular feedback system was originally designed to be used as a permanent system with a

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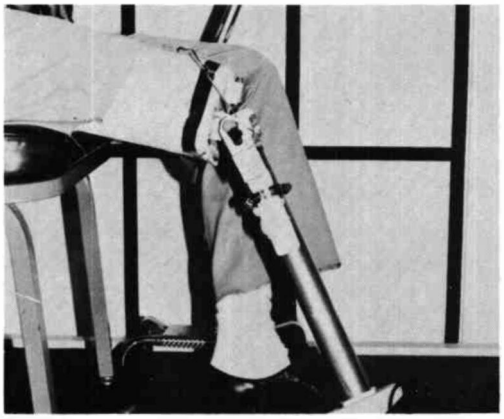

Fig. 2. Angular feedback device mounted on a temporary AK prosthesis.

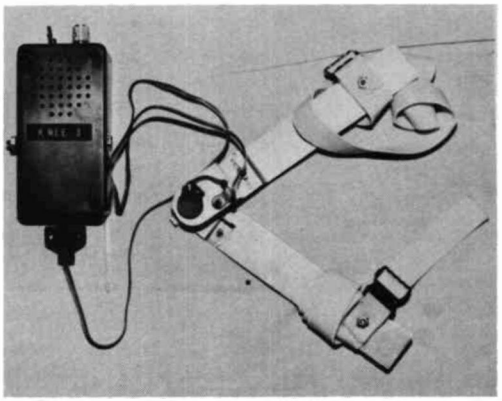

Fig. 3. Potentiometer mounted between two brackets for strapping to the prosthesis. Also shown is the belt box containing the electronics and loudspeaker.

small earpiece to deliver the auditory feedback to the patient's ear. The initial trial was on a 10 year old male. During the first week of gait training the patient elected to give up the device. The reasons for ending participation were that the patient did not like the additional attention brought to him by the device and the appearance of the earpiece was unacceptable. It was decided that these objections were not valid for a temporary training device. Also, experience with other auditory feedback systems for amputees indicates that they are most effective during the first weeks of training. Therefore, the system as described here for gait training evolved.

The research concerning the angular feedback system is still in the preliminary stages. Two additional patients have used the device after its conversion to the present training system. The first, a 31 year old AK amputee who used the system during a gait retraining period, indicated that it helped in knowing the period of swing of the prosthesis as related to the rhythm of her steps. The second patient was a 69 year old AK amputee who wore the device in his initial gait training. He reported that the device assisted him in knowing when his leg was straight. These favourable comments are supported by the positive reaction from the physical therapists involved in the gait training.

In conclusion, the angular feedback device provides a relatively inexpensive and simple means of assisting above-knee and hip disarticulation amputees in developing confidence in their gait during the training and retraining periods. From the patients' impressions it is concluded that this form of feedback can be readily learned and accepted. Expansion of the patient population and further evaluation of patient acceptability will be pursued.

Acknowledgement

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REFERENCE

VAN GRIETHUYSEN, C. (1979). Gait training for the below-knee amputee. *Pros. Orth. Int.* **3,** 163–165.

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Wound healing complications following major amputations of the lower limb

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Abstract

A series of 320 amputations was analyzed with regard to wound healing complications and re-amputation rates.

Among 111 AK amputations complications in wound healing were encountered in 14 per cent (15/111) of the cases, leading to re-amputation in 2 per cent (2/111).

TK amputations were followed by wound healing problems in 30 per cent (20/66) of the cases with re-amputation in 20 per cent (13/66) at AK level, as compared to 40 per cent (57/143) with wound healing complications and 20 per cent (28/143) re-amputations in BK-amputees.

As failure of BK amputation leads to re-amputation at AK level it is recommended that the TK level be selected in doubtful cases.

Introduction

Most amputation series reviewed in the literature consider above-knee and below-knee amputations. The re-amputation rate following below-knee amputations has been reported as 12–46 per cent (Baur et al. 1978; Chilvers et al. 1971; Christensen, 1976; Hierton & James, 1973; Lindholm, 1964; Robinson, 1976; Sarmiento et al. 1970; Warren & Kihn, 1968).

In this clinic major amputations of the lower limb are performed at the most distal level according to pre-operative measurements of the skin perfusion blood pressure (Holstein & Lassen 1973; Holstein et al. 1979 a, b, Holstein, 1980). The present study describes the results with respect to the re-amputation rates.

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Patients and methods

A thorough description of the series has been given in a previous report (Mandrup-Poulsen and Jensen, 1982). Amputations of 320 limbs were performed because of gangrene on 310 patients with a mean age of 70 years (range 40–94). In 208 limbs (65 per cent) chronic arteriosclerosis was the cause of amputation, whereas acute arterial thrombosis accounted for 32 cases (10 per cent) and vascular disease combined with diabetes mellitus for the remaining 80 cases (25 per cent).

The level of amputation was grossly determined by measurement of the skin perfusion blood pressure. The pressure was recorded at ankle level and 10 cm above and below the knee joint. The sufficient blood pressure was considered to be at least 30–40 mm Hg, although a few amputations were performed at pressure levels of 20–30 mm Hg in order to gain length of the limb in patients with an ischaemic contralateral limb.

In the retrospective study the patient files were examined, recording the primary level of amputation, the number of wound healing complications and the final level of amputation.

Results

Above-knee (AK) amputations were primarily performed in 111 patients with a mean age of 74 years (42–94), through-knee (TK) in 66 patients with a mean age of 72 years (52–94) and below-knee (BK) in 143 patients with a mean age of 67 years (40–89). A multivariate contingency table analysis demonstrated the level of amputation to be related to the age ($P < 0{\cdot}03$) and to the cause ($P < 0{\cdot}007$). This means that the number of BK amputations decreases and AK amputations increases with age and that more BK amputations were performed in cases with diabetes mellitus, as shown in Figure 1.

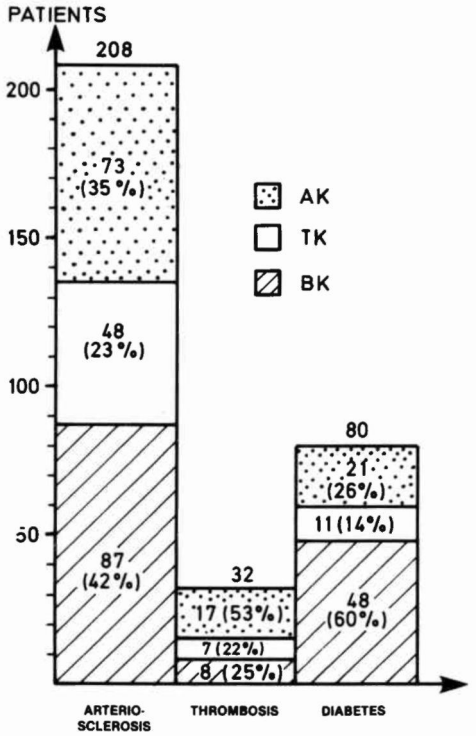

Fig. 1. Level of amputation related to cause of amputation.

As seen from Table 1, the number of wound healing complications were significantly related to the level of amputation ($P < 0{\cdot}0005$), as the majority of complications encountered arose after BK and TK amputations. There was no significant correlation to the cause of amputation, as 31 per cent (74/240) of complications were encountered in vascular cases and 23 per cent (18/80) in diabetics. All stump complications recorded in the present series were local infections or secondary skin necrosis.

Conservative treatment of the stump complications was followed by healing of the stump in 13 per cent (42/320) of cases, whereas local operations were successful in only 2 per cent (7/320) of cases (Table 2). Re-amputation at a higher level was necessary in 20 per cent (41/209) of TK and BK amputations. The number of re-amputations was significantly correlated to the cause of amputation ($P < 0{\cdot}025$, Chi-square test), as vascular cases accounted for 16 per cent (39/240), as compared to 5 per cent (4/80) of diabetics.

Table 1. Wound healing complications in relation to level of amputation.

AK	15/111	(14%)
TK	20/66	(30%)
BK	57/143	(40%)
Total	92/320	(29%)

The final level of amputation is shown in Table 3. It is seen that failure of TK amputations was followed by amputation at AK level. It is also noticed that failure of BK amputations led to re-amputation at AK level in the majority of cases, as only 3 out of 7 re-amputations at TK level resulted in successful healing of the stump.

Mortality following re-amputation was 7 per cent (6/92) in patients with wound healing complications, as compared to 2 per cent (4/167) in totally uncomplicated cases.

Table 2. Treatment of wound healing complications in relation to level of amputation.

	Conservative treatment	Operative revision	Re-amputation
AK	12/111 (11%)	1/111 (1%)	2/111 (2%)
TK	7/66 (11%)	—	13/66 (20%)
BK	23/143 (16%)	6/143 (4%)	28/143 (20%)
Total	42/320 (13%)	7/320 (2%)	43/320 (13%)

Table 3. Correlation between primary and final level of amputation.

BK to TK	3/143	(2%)
BK to AK	25/143	(18%)
TK to AK	13/66	(20%)

Discussion

In the fifties and sixties amputation for arteriosclerotic gangrene of the lower limb was synonymous with amputation at AK level. It must, however, be realized that successful prosthetic fitting following AK amputations was achieved in less than two thirds of geriatric patients in the best series (Chapman et al. 1959) and less than every quarter in others (Hierton & James 1973; Warren & Kihn, 1968).

It is true that AK amputations rarely necessitate further operations (Baur et al, 1978; Warren & Kihn 1968), but the risk of death was not proven to increase appreciably with the appearance of wound healing complications in the present series.

The goal of prosthetic treatment must be to return the geriatric patient home with preserved gait function. This means that the most distal level of amputation should be selected. It is, however, unacceptable always to choose the BK level, if the re-amputation rate is 46 per cent, as reported by Sarmiento et al, (1970). The level of amputation was determined by the skin perfusion blood pressure in the present series (Holstein et al, 1979 a, b; Holstein, 1980), although a few amputations were performed in spite of unacceptably low pressures in bilaterally threatened limbs and a number of amputations performed as life saving procedures without further investigations. The re-amputation rate following TK or BK amputations was fairly consistent with recent reports (Christensen 1976, Hopkins & Harris 1965, Howard et al. 1969, Newcombe & Marcuson 1972, Warren & Kihn 1968), but somewhat lower than others (Chilvers et al, 1971, Hierton & James 1973, Lindholm 1964). It is, however, most interesting to experience that failure after BK amputations in the majority of cases leads to re-amputation at AK level. The TK level is thus lost, which is decisive considering the high success rate of prosthetic fitting following TK amputations (Chilvers et al, 1971; Hopkins & Harris, 1965; Howard et al, 1969; Newcombe & Marcuson 1972).

In our opinion the limb should be carefully evaluated prior to selection of the level of amputation, including measurement of the skin perfusion blood pressure. In cases with borderline skin perfusion blood pressure or doubtful clinical condition of the skin or muscles, we advocate selection of the TK amputation level, as this level of amputation is followed by as successful a prosthetic fitting as after BK amputations.

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REFERENCES

BAUR, G. M., PORTER, J. M., AXTHELM, S., EIDEMILLER, L. R., HOOD, R. (1978). Lower extremity amputation for ischaemia. *Amer. Surg.* **44,** 472–477.

CHAPMAN, C. E., PALMER, H. F., BELL, D. M., BUCKLEY, A. A. (1959). Follow-up study on a group of older amputee patients. *J Amer. Med. Assoc.* **170,** 1396–1402.

CHILVERS, A. S., BRIGGS, J., BROWSE, N. L., KINMONTH, J. B. (1971). Below and through-knee amputations in ischaemic disease. *Br. J. Surg.* **58,** 824–826.

CHRISTENSEN, S. (1976). Lower extremity amputations in the county of Aalborg 1961–1971. Population study and follow-up. *Acta Orthop. Scand.* **47,** 329–334.

HIERTON, T., JAMES, U. (1973). Lower extremity amputation in Uppsala county 1947–1969. Incidence and prosthetic rehabilitation. *Acta Orthop. Scand.* **44,** 573–582.

HOLSTEIN, P., LASSEN, N. A. (1973). Radioisotope clearance technique for measurement of distal blood pressure in skin and muscles. *Scand. J. Clin. Lab. Invest.*, **31, suppl. 128,** 143–147.

HOLSTEIN, P., DOVEY, H., LASSEN, N. A. (1979a). Wound healing in above-knee amputations in relation to skin perfusion pressure. *Acta Orthop. Scand.* **50,** 59–66.

HOLSTEIN, P., SAGER, P., LASSEN, N. A. (1979b). Wound healing in below-knee amputations in relation to skin perfusion pressure. *Acta Orthop. Scand.* **50,** 49–58.

HOLSTEIN, P. (1980). Ischaemic wound complications in above-knee amputations in relation to the skin perfusion pressure. *Pros. Orth. Int.* **4,** 81–86.

HOPKINS, J. O. P., HARRIS, E. E. (1965). Knee disarticulation in arteriosclerotic limb. *Pros. Int.* **2,** 4–7.

HOWARD, R. R. S., CHAMBERLAIN, J., MACPHERSON, A. I. S. (1969). Through-knee amputation in peripheral vascular disease. *Lancet* **2,** 240–242.

MANDRUP-POULSEN, T., JENSEN, J. S. (1982). Incidence of major amputations following gangrene of the lower limb. *Pros. Orth. Int.* **6,** 35–37.

LINDHOLM, R. (1964). Features of amputation surgery among civilians during the period 1930–1960. *Acta Orthop. Scand.* **35,** 74–89.

NEWCOMBE, J. F., MARCUSON, R. W. (1972). Through-knee amputation. *Br. J. Surg.* **59,** 260–266.

ROBINSON, K. P. (1976). Long posterior flap amputation in geriatric patients with ischaemic disease. *Ann. R. Coll. Surg. Engl.* **58,** 440–451.

SARMIENTO, A., MAY, B. J., SINCLAIR, W. F., MCCOLLOUGH, N. C., WILLIAMS, E. M. (1970). Lower extremity amputation. The impact of immediate postsurgical prosthetic fitting. *Clin. Orthop.* **68,** 22–31.

WARREN, R., KIHN, R. B. (1968). A survey of lower extremity amputations for ischaemia. *Surgery* **63,** 107–120.

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Shoe inserts for small deformed feet

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Abstract

Modern materials and a better understanding of the biomechanical requirements enable adaptations to shoes to be made quickly and easily in cases where the deformed foot is small enough to fit satisfactorily into standard shop-bought or standard deep footwear. A flexible self-generating polyurethane foam is used inside the shoe. It expands to the internal shape of the shoe and the external shape of the foot. It can be used either against the patient's own foot or against a positive cast of the foot. The technique has been used for 75 patients and has proved successful. The insert so made is durable and economical.

Introduction

With the present shortage of surgical shoe makers and the high cost of bespoke footwear, (now at least £150 a pair in UK), it is necessary to maximise the use of standard footwear and provide adaptations whenever possible. Patients are pleased to wear standard shoes which provide the opportunity for better looking footwear with greater variety, even when they have to pay for them—and their repair and replacement. It is clearly more economical for a National Health Service.

Adaptations to standard footwear to accommodate misshapen feet have been made for centuries. The traditional methods still used for making these adaptations are often centuries old. Shoe stiffening is often achieved with steel plates built into the sole and "build-ups" are made in layered cork.

Modern materials, such as polyurethane foams and high molecular weight polyethylene can be used instead, with advantage, for making these shoe inserts.

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Method

Choice of using the foot or a cast of it

The technique for filling spaces between the small foot and the shoe which is described below can be performed using the foot itself (suitably covered) or a cast of it. The occasions when it is more satisfactory to use a cast are:

(a) When the attitude of the foot (or part of it e.g. forefoot) requires passive correction and this cannot be achieved when the foot is in the shoe whether weight bearing or not.

(b) When accentuation of build-up or relieved areas is required.

(c) When the foot tends to slide forward in the shoe during foaming and must be held back into the heel.

(d) When a thin layer of foam is required over the dorsum of the foot. (It is better to add a material like Pelite—which is stiff enough to stand on its own—to the cast.)

(e) When the patient has difficulty, e.g. with a flail foot, in inserting the foot into the shoe quickly.

The principle advantage of using the foot directly is, of course, that of speed. The production can proceed on a 'while-u-wait' basis. However, for this to be possible the patient has to attend with the correct shoes and an insert base to fit those shoes has to be ready. In most cases the patient will first present without shoes, especially if they are being transferred from surgical or other non-standard footwear. Thus in these cases it is easier to take a cast of the foot at the first visit and the insert (and subsequent ones) can be made up to the cast, with the patient attending on the second occasion for fitting and supply.

The insert base

Footwear containing a foot much smaller than that for which it was designed will usually need to be stiffened. The stiffening is very frequently satisfactorily achieved with an insole of 4 mm or 6 mm thick Ortholen® (high molecular weight polyethylene) shaped to the last for the shoe—or a similar last. The optimum amount of stiffening in each case has to be judged with skill. It is a biomechanical compromise to achieve flexing of the forepart of the shoe compatible with the movements of the foot. It will be influenced by the ranges of movement and muscle power at the knee, ankle, subtalar joint and toes. A wedge style shoe is sometimes stiff enough and does not require additional stiffening. Stiffeners sometimes need to be specially weakened across the region required to flex.

The foam filling technique

(a) The fit of the foot (or cast) in the shoe is checked, particular attention being paid to the heel area.

(b) The fit of the insole base (stiffener, if necessary) in the shoe and on the foot is checked. Any surface contouring (e.g. metatarsal pads, etc.) which is required is added.

(c) A shoe-shaped polythene bag (280–300 gauge) is cut, sealed and heat shrunk onto a last of approximately appropriate size for the shoe. This acts to protect the shoe and release the insert, when it is formed, from it.

(d) The insole base is placed in the polythene shoe bag and both are placed in the shoe.

(e) i. When using the patient's foot directly: The foot is clothed in stockinette and the material is bundled with extra thickness in areas such as beyond the toes and over tender prominences, where freedom is required (Fig. 1, left). The extra thickness of materials is held in place with thin adhesive tape. The foot is then covered with a shaped sock (cut and sealed with an electric hot knife (Fig. 1, right), made of Ambla P072 (polyurethane film outermost) which is then painted with a release agent, e.g. perchlorethylenum (Fig. 2, left) where adhesion to the foam is not required. The Ambla material serves to protect the foot as well as provide a lining interface to the foam.

ii. When using a cast: The cast is modified/rectified to achieve the necessary corrections of attitude or relief of pressure problem areas. The cast is then covered with a thin latex sheath. This does not require a release agent.

Fig. 1. Left, foot clothed in stockinette. Right, cutting of Ambla P072 shaped sock.

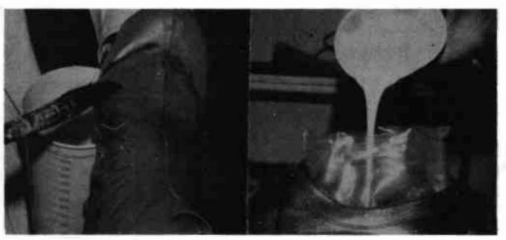

Fig. 2. Left, Release agent being applied. Right, flexible polyurethane foam being poured into polythene bag containing insole base.

(f) A suitable quantity of the components of a flexible self-generating polyurethane foam (e.g. Otto Bock Pedilen W150) are mixed and quickly poured into the polythene bag in the shoe (Fig. 2, right).

(g) The foot or the cast is placed in the shoe and the correct attitude (the patient should bear weight on the foot) is maintained for a further 2 minutes (Fig. 3, left). It should be held still for a further 3 minutes for completion of

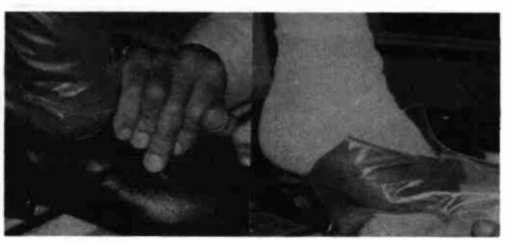

Fig. 3. Left, foot placed in shoe and correct attitude maintained for appropriate time (see text). Right, when polymerization is complete the foot is removed from the shoe together with the bag and insert.

polymerization. The foam expands to fill the space in the shoe and it sets to the internal shape of the shoe and the external shape of the foot.

(h) The foot or cast is removed from the shoe together with the polythene bag and insert (Fig. 3, right). The foam can be peeled away from the areas not adhering and trimmed and machined as necessary. In the case of the Ambla lining, it is trimmed and cemented around its borders.

(j) The insert base, with the superimposed flexible foam filling is then completed by a layer of preformed polyurethane foam (e.g. Poron) which gives a durable and comfortable finish (Fig. 4, left).

The insert is best left a further 24 hours before being taken into use to ensure stabilization of the polymer.

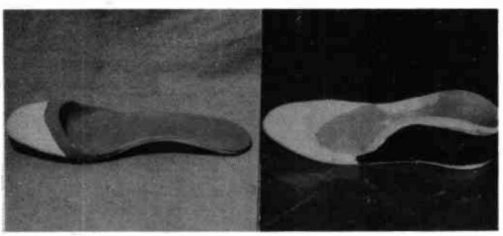

Fig. 4. Left, the finished insert. Right, insert with rigid foam under the raised heel area for patients with fixed equinus deformity.

The selection of footwear

Many factors are involved in the selection of footwear. There are certain obvious criteria such as that the foot must be capable of being contained in it. (Sometimes a pair of oversize shoes or boots can be used with a standard insole for the normal foot and a special insert for the odd one.) Where there is a degree of fixed equinus it is often possible to use standard shop-bought boots and an insert with rigid foam (e.g. Pedilen W300) under the raised heel area (Fig. 4, right).

Where the foot is too deep for ordinary shoes, standard 'Deep Shoes' may be used with the necessary insert. Mocassins are almost invariably unsuitable where there is any tendency to lateral instability. Problems involving retention of the shoe on the foot usually need boots or shoes with high fronts (preferably laced). Narrow and high-heeled ladies fashion shoes are seldom possible.

Patient selection

The basic criteria for patient selection should be:

(a) A desire to wear standard or normal-looking footwear.

(b) Feet or foot the dimensions of which (in the walking attitude) do not exceed the internal dimensions of the proposed footwear, including any necessary depth for plantar surface contouring. (The question of alteration of shape/dimension with dynamic loading has to be interpreted in the light of experience.)

(c) Sufficient forefoot and normal or near normal heel shape for the footwear to stay on the foot adequately. (Boots or bootees can overcome some of these problems.)

Results

This technique has been used to make more than 200 inserts for 75 patients. In all cases but 3 (early cases) the insert has outlasted the footwear. The average duration of patient trial was 13·6 months. Since the problems being treated should be regarded in terms of their anatomical and biomechanical abnormalities the disease conditions involved are not so important. They included 53% congenital malformations, 5% trauma sequelae and the remaining 42% from such conditions as anterior poliomyelitis, arthrogryposis, muscular dystrophy, amputation following malignancy and spina bifida.

In all but one case patients were pleased (often delighted) to convert to shop-bought or standard shoes even though they had to pay for them themselves.

Conclusions

A number of patients can be satisfactorily converted from bespoke surgical footwear to ordinary shop-bought or standard 'deep shoes' with consequent benefits to the patient in terms of appearance and to the economy in terms of Health Service savings.

The technique to make removable inserts as described and using if necessary, a flexible polyurethane filler is not difficult to learn, is relatively inexpensive and safe. It is, however, a technique for the surgical shoemaker rather than the orthotist, chiropodist or physiotherapist.

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Clinical evaluation of a knee-ankle-foot-orthosis for hemiplegic patients

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Abstract

A knee-ankle-foot-orthosis has been developed that incorporates a genucentric knee joint and a similarly designed ankle joint. Its design is discussed and a clinical evaluation of its use on twenty five hemiplegic patients is presented.

Introduction

At present, various orthoses for hemiplegics are designed to correct deformities and allow effective ambulation. A knee-ankle-foot-orthosis (KAFO) has been newly developed with a genucentric knee joint (Foster & Milani, 1979) and a similarly designed ankle joint connected to an arch foot support with a specially designed heel trim. Over the past three years, more than thirty patients have been prescribed these orthoses with very satisfactory results.

Materials and design

The orthosis is a long leg shelled type, made of a flexible plastic laminate, "Subortholen". The thigh part of the orthosis consists of an anterior wall and posterior cloth cuff. It is connected with the lower leg shell by the polycentric (genucentric) knee joint which is made by overlapping the anterior thigh shell and the lower leg shell (Fig. 1).

This allows a range of movement from 10 to 130 controlled by a small plastic stopper attached to the upper edge of the lower leg shell (Fig. 2).

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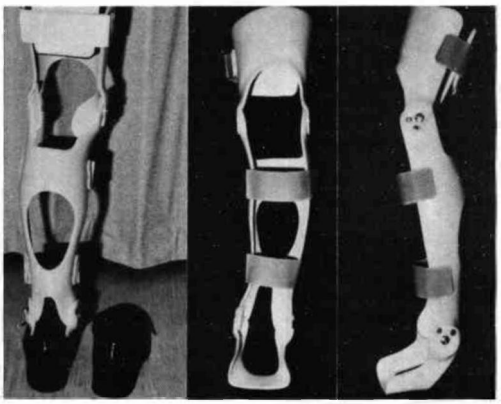

Fig. 1. The knee-ankle-foot-orthosis.

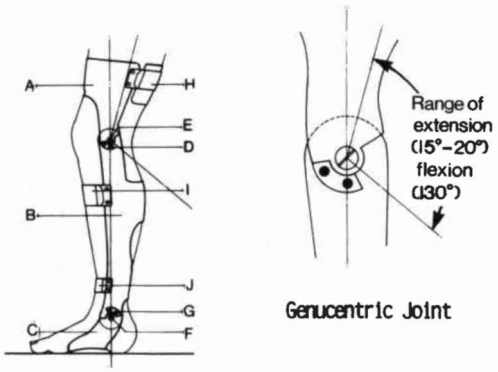

Fig. 2. (A) Anterior thigh shell
(B) Lower leg shell
(C) Arch support
(D) Plastic stopper
(E) Genucentric knee joint
(F) Polycentric ankle joint
(G) Plastic stopper
(H) Posterior cuff
(I) Upper anterior cuff
(J) Lower anterior cuff

When the knee joint of a subject is unstable, the knee joint of the orthosis can be firmly locked by means of the small bars attached posteriorly.

The orthosis is light in weight, weighing about 500 gm. The ankle joint of the brace is also a polycentric joint made in the same way as the knee joint. The ankle joint allows a range of movement from 0 to 30 in dorsiflexion. Holding the knee joint in a slightly flexed position in the orthosis, a hemiplegic can swing his affected leg forward easily. If the knee joint of a subject is unstable it can be fixed in the extended position by locking the small bars (Fig. 3); these can be unlocked after gait training, when the subject becomes able to achieve smooth ambulation with a free knee joint.

The heel trim of the orthosis is shaped so that the subject's heel does not slip out of the orthosis. The foot part of the orthosis consists of the arch support connected to the ankle joint; it does not cover the forefoot. By this device, the subject can carry out the toe-off phase of ambulation more easily (Fig. 4). This orthosis has moderate flexibility and consequently it can prevent confinement or immobilization of the affected leg in the brace. The patient can put on or remove the orthosis easily and insert it into a shoe.

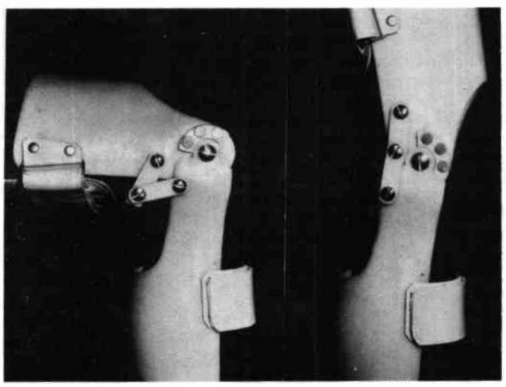

Fig. 3. Knee joint locking mechanism.

Results

The orthosis has been fitted to thirty six hemiplegic patients over the past three years, October 1978 to March 1981. These subjects were graded as stage 3–3·5 in Brunnstrom's functional classification of hemiplegia. (Brunnstrom, 1966). Twenty five of the thirty six cases were observed to investigate the results of the fittings (Table 1).

Fig. 4. Effective heel strike and toe off.

All twenty five cases were able to ambulate smoothly after fitting and all have continued to wear the orthosis without rejection. The length of time it has been fitted was 14·6 months on average, the longest period being thirty five months and the shortest one month.

Cases 6 and 23 could remove the orthosis after 12–15 months, and can now carry out independent ambulation. Another two cases, cases 4 and 25, prefer to wear the orthosis, however they seem to be able to ambulate independently without them.

Three cases were fitted with a KAFO after being fitted with an AFO due to instability of the knee joint (case 8), muscle weakness (case 11), and severe foot deformity (case 22). All could ambulate effectively in the KAFO.

Case 8 can now walk in the KAFO for long distances, 6 kilometres daily, Case 5 is now working as a member of the municipal assembly, the knee joint of the KAFO in his case was locked at the onset of application.

Case 14 with psychological disturbances, and cases 15 and 24 with weakened leg muscles can walk in the KAFO under supervision. The following investigations were performed.

Table 1. Hemiplegics supplied with KAFO

Case number	Sex	Age	Attack of stroke	Onset of bracing	Brunnstrom's classification		Duration of bracing (months)	Present stage		Comments
					UE	LE		UE	LE	
1	M	71	Jun. '78	Oct. '78	3	Pre 3	35	3·5	3·5	(1) A recurrence of stroke in Feb. 1981.
2	M	54	Jun. '78	Oct. '78	2	Pre 3	35	3	3	(4) Marked preference for KAFO.
3	F	57	Mar. '79	Apr. '79	2	Pre 3	29	3	3	(5) Engaged as a member of the municipal assembly at present. Application of KAFO with locked knee joints at the onset.
4	M	64	Mar. '78	Jul. '79	3	3	26	3	3·5	(6) Removal of KAFO at present after application for about 1 year.
5	M	65	Apr. '79	Sep. '79	2	2	24	3	3	(8) Maintenance of ability to ambulate for a long distance about, 6 km daily, without breakage of KAFO.
6	F	55	Jul. '78	Jun. '79	3	Pre 3	12	4	3·5	(9) Maintenance of ability to ambulate for a distance about 2–3 km daily in home.
7	F	67	Jul. '76	Oct. '79	3	3	23	3	3	(11) Exchange for KAFO due to instability of the knee joint in AFO. Preference for KAFO.
8	M	53	Feb. '72	Jun. '79	3	3	27	3	3	(12) Acquirement of stability of the knee joint in KAFO.
9	M	62	Jun. '80	Aug. '80	2	Pre 3	13	3	3	(14) Acquirement of the ability to ambulate under observation, psychological disturbance and distinct synergy movement.
10	M	27	Mar. '80	Sep. '80	3	3	12	4	3·5	(15) Acquirement of the ability to ambulate fairly quickly in spite of muscle weakness at hip and knee joints.
11	M	53	Dec. '79	Nov. '80	3	3	10	3	3	(16) Acquirement of the ability to ambulate in spite of flaccid and weakened muscles at hip and knee joints.
12	F	62	Oct. '80	Dec. '80	2	Pre 3	9	3	3·5	(19) In addition to a stable and not so slow ambulation, exchange for AFO in the near future, due to obtaining stable knee joint in KAFO.
13	F	70	Feb. '79	Apr. '79	2	Pre 3	29	3	3	(20) Continuous performance of gait training in KAFO at present.
14	F	62	Dec. '80	Apr. '81	1	Pre 3	5	2	3	(21) Exchange for KAFO due to incomplete correction of severe foot deformity in AFO, fair improvement of foot deformity in KAFO.
15	M	58	Feb. '81	Apr. '81	1	2	5	3	3	(22) Removal of KAFO after application for 15 months maintenance of the ability of independent ambulation at home at present.
16	M	42	Nov. '80	Apr. '81	1	2	5	3	3	(24) Maintenance of the ability to ambulate under observation in spite of marked muscle weakness and low graded functional stage.
17	F	45	Dec. '80	Feb. '81	2	Pre 3	7	3	3	(25) Exchange for AFO after application of KAFO for 6 month's stable ambulation.
18	F	61	Feb. '81	Apr. '81	2	Pre 3	5	3	3·5	
19	M	40	Feb. '81	May. '81	2	3	4	3	3·5	
20	M	51	Mar. '81	Aug. '81	1	2	1	2	3	
21	M	67	Feb. '81	Apr. '81	1	2	5	2	3	
22	M	51	Mar. '81	Jul. '81	3	3	2	3	3	
23	M	48	Oct. '78	Jun. '80	3	3	15	4	3·5	
24	F	65	Aug. '77	Feb. '80	2	Pre 3	19	2	3	
25	M	58	Dec. '80	Jan. '81	3	3	8	4	4	
Average		56	UE = Upper Extremity LE = Lower Extremity				14·6			

Cases – functional stage between 3-3·5 based on Brunnstrom's classification

	KAFO (Group A)		AFO (Group B)	
Sex	Male	Female	Male	Female
Numbers	16	9	33	17
Average age in years	56		59	
Total number	25		50	
Average time after stroke	20 months		40 months	

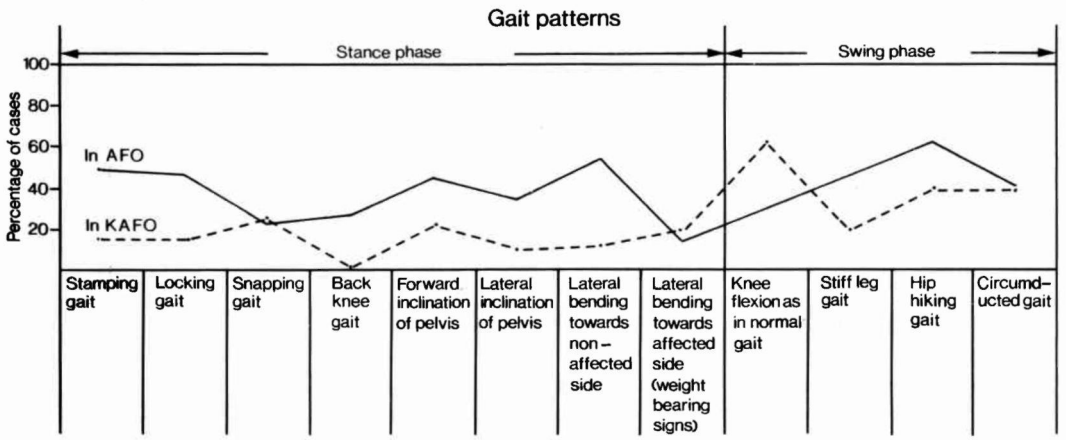

Fig. 5. Improvement of gait pattern in KAFO compared with AFO.

(1) Twenty five cases fitted with KAFO's (Group A) were compared with fifty cases fitted with AFO's (Group B) by evaluating twelve characteristics of gait (Fig. 5).

These comprised ten unfavourable signs (stamping gait – foot stamping on the ground; locking gait – knee fixed in full extension by spasticity; snapping gait—the knee snapping into full extension; back-knee gait – hyperextended knee; forward inclination of pelvis; lateral inclination of pelvis; lateral bending towards non-affected side; stiff leg gait; hip hiking gait; circumducted gait) and two favourable signs (weight-bearing by the affected leg in the orthosis and knee flexion during swing phase). All cases were graded as stage 3–3·5 in Brunnstrom's functional classification. Group B showed many more unfavourable signs than Group A, for example, stamping gait, locking gait, back-knee gait, forward inclination of pelvis, lateral bending towards non-affected side, stiff leg gait, hip hiking gait. Knee flexion was better in Group B compared to Group A.

(2) The necessary time to walk for a distance of 5–10 m was investigated. The above-mentioned two groups and a third control group (C), which comprised thirty normal adult males, were studied. Three estimations were performed, namely, the time necessary for walking 10 m in a straight line drawn on the floor, walking and returning a distance of 10 m in an L-shaped line (a total of 20 m), and walking and returning for a distance of 5 m in an S-shaped line to (a total of 10 m) (Fig. 6).

Group A showed about half to one third the times as compared with group B. The times of group A were only slightly longer than those of group C. Almost all of group A seemed able to turn easily and ambulate on the S-shaped line smoothly.

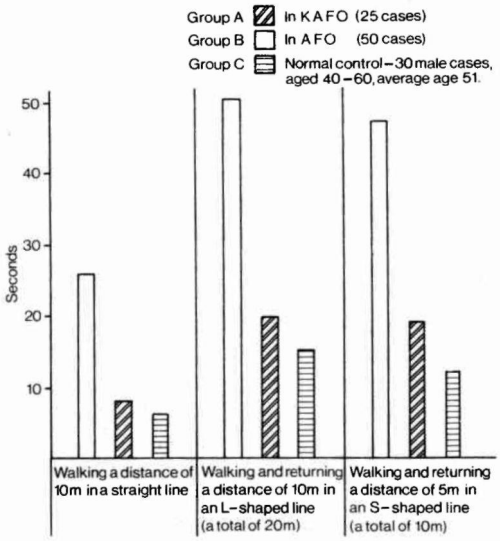

Fig. 6. Walking times, KAFO (Group A) and AFO (Group B) compared. Group C normal controls.

(3) EMG investigations were carried out during walking to compare the KAFO user with the AFO user.

Figure 7 shows typical records of the EMG obtained from the main leg muscles of the affected side by means of surface electrodes. The KAFO user exhibits periodic patterns typical of normal gait.

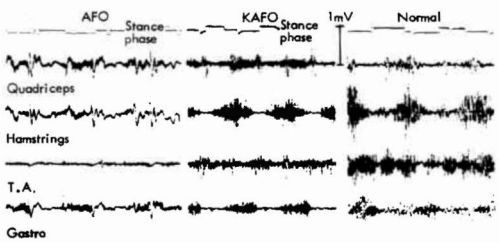

Fig. 7. EMG of a typical subject walking in AFO and KAFO (see text).

Indications

This KAFO is principally indicated for hemiplegics graded as stage 3–3·5 in Brunnstrom's functional classification aimed at ambulation in braces.

The above-mentioned hemiplegics can ambulate in the KAFO, showing significant improvement of unfavourable signs, such as stamping gait, locking gait, back-knee gait, forward inclination of pelvis, lateral inclination of pelvis, lateral bending towards non-affected side, stiff leg gait, hip hiking gait, circumduction. These unfavourable signs are not significantly improved upon in hemiplegics by using an AFO. As the results of the application show, this KAFO has a wider range of indication, in that it can be applied to hemiplegics with leg muscle weakness, or instability of knee or ankle joints (or both joints) and also with less severe deformities.

Summary

The KAFO described provides hemiplegics with effective and dynamic ambulation, because of its light weight, easy application, reasonably located genucentric knee and ankle joints, together with the flexibility of thigh and lower leg cuffs and arch support. The flexibility of this orthosis permits proper torsion of thigh and lower leg cuff.

After application of the KAFO, hemiplegics become able to extend or flex their hip or knee joints in a wide range of motion.

As the result of these characteristics, hemiplegics can ambulate smoothly and effectively in the KAFO as described in the results and practical investigations.

REFERENCES

BRUNNSTROM, S. (1966). Motor testing procedures in hemiplegia based on sequential recovery stages. *Phys. Ther.* **46,** 357–375.

FOSTER, R., MILANI, J. (1979). The genucentric knee orthosis; a new concept. *Orth. Pros.* **33 (2),** 31–44.

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April 1982, Vol. 6, No. 1

Deutsch

Die funktionelle Behandlung von Knoechelfrakturen mit Orthesen vorlaeufiger Bericht
S-A. Ahlgren, J. Hultin, J. Nilsson und L. Westman
Pros. Orth. Int. **6:1,** *24–26*

Zusammenfassung
Der Bericht untersucht die Behandlung von Knöchelfrakturen mit Orthesen. Zwei im oberen Sprunggelenk bewegliche Orthesen und die Behandlungsergebnisse an 10 Patienten werden vorgestellt.

Untersuchung der elastizitaet Geschaeumter Kunststoffe und Moosgummi fuer Schuheinlagen
G. Campbell, E. Newell und M. McLure
Pros. Orth Int. **6:1,** *48–52*

Zusammenfassung
Das Verhalten von 31 verschiedenen Schaumstoffen unter Druck wurde untersucht inbezug auf deren Eignung in Schuheinlagen. Die Werkstoffe konnten in drei Gruppen eingeteilt werden: Wenig, mässig und stark deformierbar. Die mässig deformierbaren Materialien scheinen für die klinische Anwendung am besten geeignet zu sein.

Ganguntersuchungen an Spina-Bifida-Kindern
N. C. Carroll, D. Jones, W. Maschuich, M. Milner und C. White
Pros. Orth. Int. **6:1,** *27–34*

Zusammenfassung
Am Ontario Crippled Children's Center (OCCC) werden zahlreiche Kinder mit Spina bifida untersucht, die langwieriger und kostspieliger Behandlung bedürfen. Operative Behandlung und Orthesenversorgung sind die beiden wichtigsten Behandlungsmöglichkeiten zur Verhütung von Deformitäten. Die Nützlichkeit einer Orthese liess sich bisher nur durch den praktischen Versuch abklären.

Die arbeit befasst sich mit dem Einfluss der Orthesen auf das Gangbild des Kindes. Behandelt wurden 15 Kinder mit lumbaler oder sakraler Myelomeningocele. Die Untersuchung umfasste einen dreiteiligen Fragebogen, ausgefüllt durch orthopäden, Physiotherapeuten und orthopädietechniker. Im Ganglaboratorium erfolgte die klinische Untersuchung des Gangbildes, das mit dem Videorecorder aufgenommen wurde. Dabei wurde versucht, objektive und subjektive Daten auseinanderzuhalten.

Wundheilungsprobleme bei Amputierten Gefaesspatienten
G. Horne und J. Abramowicz
Pros. Orth. Int. **6:1,** *38–40*

Zusammenfassung
Die Arbeit befasst sich mit der Behandlung des amputierten Gefässpatienten, insbesondere mit den operativen Methoden und der Nachbehandlung mit dem Ziel einer möglichst ungestörten Wundheilung. Besprochen werden schliesslich die Behandlungs—methoden bei Wundheilungsstörungen.

Vorschlag zur Technischen Pruefung Myoelektrischer Handprothesen mit Einfacher Funktion
B. Ingvarsson, I. Karlsson, L-G. Ottosson und M. Thyberg
Pros. Orth. Int. **6:1,** *41–42*

Zusammenfassung
Auf Verlangen des Swedish Institute for the Handicapped hat das orthopädietechnische Versuchslaboratorium der Universitätsklinik von Linköping in Schweden Richtlinien ausgearbeitet für die technische Prüfung monofunktionaler myoelektrischer Handprothesen.

Die Haeufigkeit von Beinamputationen Infolge Gangraen
T. Mandrup-Poulsen und J. Steen Jensen
Pros. Orth. Int. **6:1,** *35–37*

Zusammenfassung
Die Haeufigkeit von Beinamputationen nach Gangrän in den Jahren 1971 bis 1979 im Bezirk von Kopenhagen wurde errechnet. Sie betrug 0,3%o aller Einwohner über 40 Jahre. Männer sind doppelt so häufig betroffen wie Frauen. Das Risiko steigt exponential mit zunehmendem Alter. Die Häufigkeit hat sich in den untersuchten 8 Jahren nicht geändert.

Nachkontrolle von 14 400 Amputierten Waehrend 25 Jahren
I. C. Narang und V. S. Jape
Pros. Orth. Int. **6:1,** *10–16*

Zusammenfassung
Der Bericht befasst sich mit 14 400 behinderten Zivilpersonen, die über 25 Jahre am Prothesenzentrum Pune, Indien, behandelt wurden. Die Untersuchung erstreckt sich auf Geschlechts- und Altersverteilung, Ursache und Höhe der Amputation, Kostenträger und andere für die Rehabilitation wichtige Faktoren.

Armprothesen fuer Kurze Oberarmstuempfe
J. K. Ober
Pros. Orth. Int. **6:1,** *17–20*

Zusammenfassung
Untersuchung der Wertigkeit kurzer Oberarmstümpfe im Bewegungsmuster des Patienten. Die Prothesenversorgung darf die Beweglichkeit keinesfalls einschränken. Sie hat auch auf das individuelle Bewegungsmuster eines jeden Patienten Rücksicht zu nehmen. Einige Versorgungsbeispiele werden vorgestellt.

Eine Untersuchung Kinematischer und Kinetischer Variablen fuer die Erfassung des Prothesenganges mit Hilfe des Enoch-systems
K. Oberg und H. Lanshammar
Pros. Orth. Int. **6:1,** *43–47*

Zusammenfassung
Untersuchung des Gangbildes, Gelenkwinkel, Bodenkräfte und Gelenkbewegungen bei normalen Versuchspersonen, Unter- und Oberschenkelamputierten.

Die Untersuchung zeigte, dass der Hüft-Kniewinkel und verschiedene Symmetriediagramme (z.B. Winkel des linken Knies verglichen mit der Gegenseite) geeignet sind, Abweichungen vom Gangbild zu quantifizieren. Ein System namens Enoch diente zur Messung und Untersuchung. Das System besteht aus einem Mini-computer, der mit verschiedenen Untersuchungsgeräten verbunden ist. Die Ergebnisse wurden durch einen Computer Terminal (Tektronix) dargestellt.

Neue Gelenke aus Kunststoff fuer Knuststoff-orthesen
H. Watanabe, T. Kutsuna, H. Morinaga und T. Okabe
Pros. Orth. Int. **6:1,** *21–23*

Zusammenfassung
Gelenke aus Kunststoff für Orthesen sind Metallgelenken überlegen. Sie sind leicht, geräuschlos, bequem zu tragen, rostfrei, korrodieren nicht und sind strahlendurchlässig.

Die Verfasser haben zwei verschiedene Gelenke aus Kunststoffen entwickelt, das eine für das Sprunggelenk, das andere für Knie, Ellbogen oder Hüfte. Als Werkstoff wurde Polypropylen gewählt, weil es geschmeidig und dennoch widerstandsfähig ist.

Español

Ortesis para tratamiento funcional de las fracturas de tobillo informe preliminar
S-A. Ahlgren, J. Hultin, J. Nilson y L. Westman
Pros. Orth. Int. **6:1,** *24–26*

Resumen
Este reporte analiza el tratamiento ortesico de pacientes con fractura de tobillo. Se describen dos ortesis funcionales que permiten movimiento de este. Se discuten los resuitados obtenidos al usar dichas ortesis en diez pacientes.

Pruebas de compresion de gomas y espumas plasticas para su uso en plantillas
G. Campbell, E. Newell y M. McLure
Pros. Orth. Int. **6:1,** *48–52*

Resumen
Se han experimentado treinta y un materiales por compresión para comprobar la relación entre la presión por área y la deformación del material con objeto de evaluar varias espumas plásticas y gomas como materiales para plantilla. Se ha visto que los materiales pueden clasificarse en tres categorías (muy duro, moderadamente deformable y muy deformable). Se ha seleccionado el grupo de moderadamente deformables como los mejores para su aplicación clínica.

Evaluacion efectiva de la marcha de ninos con meningomielocele
N. C. Carroll, D. Jones, W. Maschuich, M. Milner y C. White
Pros. Orth. Int. **6:1,** *27–34*

Resumen
En el Ontario Crippled Children's Centre (OCCC), se tratan muchos niños con espina bífida que necesitan un largo y costoso tratamiento de rehabilitación. Los objetivos del tratamiento ortopédico se pueden conseguir mejor por medio de una evaluación efectiva de cada niño, lo que nos dirige al tratamiento conservador y operatorio durante la infancia y la niñez.

La cirugía y la ortésica son los principales caminos para corregir o prevenir el desarrollo de deformaciones ortopédicas. Ahora el único método de evaluar una ortesis es esperar y ver si mejora la función o evita la deformidad.

Esta comunicación se refiere al trabajo inicial llevado a cabo para aclarar los factores que contribuyen a la progresión de las deformidades de la extremidad inferior. La intención fue medir estos factores con el mínimo de coste y de forma no invasiva y utilizar la información que se ha obtenido en el desarrollo de la ortésica en los niños.

El objeto inicial de este estudio fue cuantificar el efecto de una determinada ortesis en la marcha de cada niño. Se han examinado 15 niños con meningomielocele a nivel lumbar y sacro en un total de 59 pruebas. Los datos son a tres niveles: del técnico ortopédico, del fisioterapeuta y del ingeniero. En el laboratorio de marcha del OCCC se han realizado un examen clínico, un video de la marcha y un control con instrumentos. El criterio para la evaluación se hizo de los datos que se obtuvieron. Este criterio fue después de examinar la frecuencia de algunas características de la marcha en grupo de estudio.

El examen de la influencia de una ortesis en la marcha del niño se concretó en las que no se pudo indicar "lo mejor" o "mejor". El valor de estos juicios ha sido indicar los relativos méritos de la ortesis seleccionada. Combinando la objetividad de los datos obtenidos con instrumentos del análisis de la marcha con la capacidad subjetiva pero tangible del observador se ha conseguido una mejora de la evaluación.

Tratamiento de los problemas de cicatrizacion en los amputados con problemas vasculares
G. Horne y J. Abramowicz
Pros. Orth. Int. **6:1,** *38–40*

Resumen
Describe las técnicas quirúrgicas y el cuidado postoperatorio para evitar los problemas de cicatrización. También se estudia el tratamiento de los mismos.

Instrucciones tecnicas para comprobar las manos protesicas mono-funcionales controladas mioelectricamente.
B. Ingvarsson, I. Karlsson, L-G Ottosson y M. Thyberg
Pros. Orth. Int. **6:1,** *41–42*

Resumen
Para conseguir un buen standard de protésis y ortésis en Suecia el Swedish Institute for the Handicapped está probando este grupo de aparatos. Una parte importante del trabajo es redactar las instrucciones.

A petición del Swedish Institute for the Handicapped, el Laboratory of Rehabilitation Engineering en la University Hospital de Linkoping en Suecia ha propuesto unas instrucciones para la comprobación de ese tipo de protésis de mano. Estas instrucciones contienen varias pruebas que son importantes para la función de la protésis.

Incidencia de amputaciones por gangrena de extremidades inferiores
T. Mandrup-Poulsen y J. Steen Jensen
Pros. Orth. Int. **6:1,** *35–37*

Resumen
Se ha calculado la incidencia de las amputaciones por gangrena en el distrito de Copenhague entre 1971 y 1979. La incidencia es de 0.3 por mil habitantes de más de 40 años de edad y la relación de hombre a mujer ha sido de 2:1. La incidencia aumenta con la edad. Estas cifras no cambiaron durante este período de 8 años.

Estudio de 14.400 neuvos discapatizados tratados durante 25 años en un centro de miembros artificiales
I. C. Narang y V. S. Jape
Pros. Orth. Int. **6:1,** *10–16*

Resumen
Reporta estos casos tratados en el Centro de Miembros Artificiales de Pune, India. Se estudia su distribución por sexo, edad, causa, nivel de amputación, fuentes de pago y otros factores con ellos relacionados.

Protesis de miembro superior para amputaciones a alto nivel del brazo
J. K. Ober
Pros. Orth. Int. **6:1,** *17–20*

Resumen
En un análisis de la influencia del movimiento del muñón residual y de la parte alta del cuerpo en las funciones de la protésis. Se indican las técnicas protésicas que no restringen el movimiento. Se presenta una protésis con cable sencillo híbrido en diferentes casos. La estructura cinética y el control de la protésis varia en cada caso según las características de manipulación del paciente. Se presenta una técnica que tiene características modulares vistas desde el punto de vista cinématico.

Investigacion de variables cinëmaticas y cineticas para la descripcion de la marcha protesica usanso el sistema enoch
K. Oberg y H. Lanshammar
Pros. Orth. Int. **6:1,** *43–47*

Resumen
Se han investigado formas de marcha, ángulo de las articulaciones, fuerzas de reacción del suelo y momentos de las articulaciones durante la marcha en sujetos normales y en amputados.

La investigación muestra que el diagrama del ángulo cadera-rodilla, así como otros diagramas simétricos (p. ejemplo: el ángulo de la rodilla izquierda don el de la rodilla derecha) nos da un medio fácil de evaluar las anormalidades en la marcha. Y además un diagrama vector combinado del diseño de la marcha y la fuerza tiene interés en la evaluación de los momentos de la articulación.

También se incluyen las fuerzas de reacción del suelo y los momentos de los musculos en las articulaciones. Los momentos en la articulación de la rodilla son muy diferentes en los amputados por encima y debajo de la rodilla, comparados con el sujeto normal. También se han encontrado algunas ideas sobre la estabilidad de la rodilla.

Se ha usado el método Enoch. El sistema consiste en un microcomputador conectado al equipo para medir los desplazamientos (Selspot) y las fuerzas de reacción del suelo. Para la presentación de los resultados se ha usado un computador terminal gráfico.

Nuevas articulaciones de plastico para ortesis de plastico
H. Watanabe, T. Kutsuna, H. Morinaga y T. Okabe
Pros. Orth. Int. **6:1,** *21–23*

Resumen
Las articulaciones de plástico para ortésis tienen más ventajas que las de metal. Son más ligeras, silenciosas, comodas, no se oxidan y son transparentes a los Rayos X.

Los autores han desarrollado dos tipos de articulaciones, una para tobillo y otra para rodilla, codo o cadera. Se ha usado polipropileno por su flexibilidad y resistencia.

Français

Ortheses pour le traitement fonctionnel des fractures de cheville rapport preliminaire
S-A. Ahlgren, J. Hultin, J. Nilsson et L. Westman
Pros. Orth. Int. **6:1,** *24–26*

Résumé
Ce rapport traite des orthèses pour des patients souffrant de fractures de cheville. Deux orthèses fontionnelles permettant un mouvement de l'articulation tibio-calcanéénne sont décrites. La discussion porte sur les résultats chez 10 patients porteurs de ces prothèses.

Tests de compression sur des matieres plastiques et des caoutchoucs utilises pour des semelles orthopediques
G. Campbell, E. Newell et M. McLure
Pros. Orth. Int. **6:1,** *48–52*

Résumé
31 Matériaux ont été testés quant à leur compressibilité (force par unité de surface par rapport à leur déformation), dans le but de dresser la liste des mousses plastiques et caoutchoutées les plus aptes à servir pour des semelles internes. Nous avons pu les classifier en 3 groupes: "très dur", "peu déformable" et "très déformable" selon les courbes de mesures. Le groupe des "peu déformables" est celui promettant les meilleurs résultats d'applications cliniques.

Étude de la marche chez des enfants porteurs de myélo-méningocoeles
N. C. Carroll, D. Jones, W. Maschuich, M. Milner et C. White
Pros. Orth. Int. **6:1,** *27–34*

Résumé
Au centre 'Ontario Crippled Chidren's Center' (OCCC) des enfants handicapés de l'ontario, nous voyons de nombreux enfants avec spina bifida, nécessitant des traitements coûteux et prolongés. Le but thérapeutique orthopédique le plus efficace est obtenu grâce à une plan "effectif" pour chaque enfant guidant le traitement conservatif et opératoire durant les premières années de l'enfant. La chirurgie et les orthèses sont les 2 meilleurs traitements pour éviter les difformités.

Cet article concerne un travail-pilote cherchant à cerner les facteurs qui contribuent à la progression des difformités des extrêmités inférieures. L'intention était de mesurer ces facteurs tant au point de vue financier qu'à celui du traitement non-invasif et d'utiliser cette information pour des orthèses et le développement de ces enfants.

Le but premier de cette étude était de quantifier l'effet d'une orthèse sur la marche de chaque enfant. Quelque 15 enfants sur un total de 59 avec myelo-méningocoele lombaire ou sacrée ont été examinés. Le relevé des données a été fait dans 3 protocoles différents par un orthopédiste, un physiothérapeute et un technicien. Au laboratoire de l'OCCC, l'étude englobait un examen clinique global, une étude de la marche au vidéo-tape et und mesure de la marche aux instruments. Certains critères furent proposés et extraits de ces relevés; ils furent décidés en examinant la fréquence de certains caractères de la marche du groupe étudié. Un exemple est donné.

En examinant la performance et l'influence d'une orthèse sur la marche d'un enfant, nous avons reconnu les situations dans lesquelles il n'était pas possible de reconnaître le "mieux" du "parfait". La valeur de cette étude était de relever les mérites *relatifs* de certaines orthèses. En combinant les résultats objectifs avec ceux subjectifs mais concrets d'un observateur expérimenté il est possible d'obtenir un jugement significatif de la performance.

Probleme de plaie operatoire chez les amputes sur troubles vasculaires
G. Horne et J. Abramowicz
Pros. Orth. Int. **6:1,** *38–40*

Résumé
Ce travail traite des soins aux amputés sur troubles vasculaires. La technique chirurgicale et les soins postopératoires visant à prévenir les problèmes de cicatrisation sont décrits. Le traitement des patients dont les plaies cicatrisent mal est décrit.

Proposition de tests techniques concernant les protheses de mains myo-electriques
B. Ingvarsson, I. Karlsson, L-G. Ottosson et M. Thyberg
Pros. Orth. Int. **6:1,** *41–42*

Résumé
Sur la demande de l'institut suédois pour les handicapés, le laboratoire technique de rééducation de l'hôpital universitaire de Linkoping en Suède a proposé des directives concernant les tests techniques de prothèses de main myo-électriques mono-fonctionnelles. Ces

instructions contiennent divers facteurs d'instruction et de contrôle importants pour ce genre de prothèses.

Fréquence des amputations hautes faisant suite a une gangrene de membre inferieur
T. Mandrup-Poulsen et J. Steen Jensen
Pros. Orth. Int. **6:1,** *35–37*

Résumé
La fréquence des amputations hautes faisant suite à une gangrène du membre inférieur a été calculée durant la période de 1971 à 1979 dans la région de Copenhague.

Elle était de env. 0,3 p. mille habitants en-dessus de 40 ans et 2 fois plus élevée chez les hommes que chez les femmes. Cette fréquence augmente selon une courbe expérimentelle avec l'âge. Elle n'a pas varié au cours des années.

Etude retrospective de 14'400 amputés (civils) traités en plus de 25 ans
I. C. Narang et V. S. Jape
Pros. Orth. Int. **6:1,** *10–16*

Résumé
Cette étude comprend 14'400 patients civils amputés traités pendant plus de 25 ans au centre d'appareillage artificiel de Puna aux Indes. Il examine la distribution selon le sexe et l'âge, les facteurs financiers et autres concernant la rééducation.

Prothèses de membre supérieur adaptées sur des moignons courts
J. K. Ober
Pros. Orth. Int. **6:1,** *17–20*

Résumé
Ce travail recouvre l'analyse du moignon, des mouvements de la partie supérieure du tronc et du schéma corporel en mouvement lors d'un travail manuel.

Une attention particulière est portée aux prothèses qui ne doivent en aucun cas réduire les mouvements résiduels. Différents cas cliniques sont présentés, porteurs d'une prothèse à contrôle hybride.

La structure cinéthique et le contrôle de la prothèse varie de cas en cas.

Une technique non conventionelle universelle est présentée; du point de vue cinématique elle offre des possibilités modulables.

Etude des variables cinematiques et cinetiques pour la description de la marche avec prothese a l'aide du systeme enoch
K. Oberg et H. Lanshammar
Pros. Orth. Int. **6:1,** *43–47*

Résumé
Nous avons étudié les caractéristiques de la marche, des angles et mouvements des articulations, les forces de réaction au plancher chez des sujets normaux ainsi que chez des amputés de cuisse et du jambe.

L'étude montre que le diagramme de l'angle hanche-genou ainsi que différents diagrammes de symmétrie (par exemple angle du genou droit comparé à celui du genou gauche) donne des résultats facilement interprétables dans l'évaluation de la marche. Nous avons aussi trouvé qu'un diagramme combiné des vecteurs de force et caractéristiques de la marche était aussi adéquat à l'étude des articulations.

L'analyse comprenait aussi l'étude des forces de réaction au plancher et des moments (au sens physique du terme) musculaires au niveau articulaire. Ces moments au genou sont très différents selon l'amputation (au-dessus ou au-dessous du genou) et par comparaison avec le sujet normal. Certaines tendances intéressantes quant à la stabilité ont été montrées.

Un système appelé ENOCH a été utilisé pour les mesures et l'analyse. Il consiste en un mini-computer connecté linéairement pour les mesures de déplacement (Selspot) et de force de réaction au plancher (Kistler). Un computer graphique terminal (Tektronix) présentait les résultats.

Articulations en plastic d'un nouveau genre pour les ortheses en plastic
H. Watanabe, T. Kutsuna, H. Morinaga et T. Okabe
Pros. Orth. Int. **6:1,** *21–23*

Résumé
Les articulations en plastic ont plusieurs avantages sur celles en métal: elles sont plus légères, moins bruyantes, confortables à l'usage, ne rouillent pas, ne se corrodent pas et sont radiotransparentes.

Deux plastics différents ont été développés par les auteurs, l'un pour la chenille, l'autre pour le genou, le coude ou la hanche. Ils ont choisi le polypropylène pour sa flexibilité et le lisse de sa surface.

Italiano

Ortesi per il trattamento funzionale delle fratture della caviglia rapporto preliminare
S-A. Ahlgren, J. Hultin, J. Nilsson e L. Westman
Pros. Orth. Int. **6:1,** *24–26*

Riassunto
In questo rapporto si esamina la cura ortetica di pazienti con fratture della caviglia. Sono descritte due ortesi funzionali che permettono movimento nella giuntura talocrurale. Sono discussi i risultati ottenuti dall'uso di queste ortesi con dieci pazienti.

Prove di compressione di materiali in plastica e gomma espansa per suole interne su calzature ortopediche
G. Campbell, E. Newell e M. McLure
Pros. Orth. Int. **6:1,** *48–52*

Riassunto
31 materiali sono stati sottoposti alle prove di compressione per indurre la sollecitazione (forza per unità di superficie a sezione trasversale) rispeto al comportamento di deformazione (stress versus strain (deformation) behaviour), per la valutazione di varie plastiche o gomme espanse da utilizzare come materiale per le suole interne di calzature. Si è riscontrato che questi materiali potevano essere classificati in tre categorie distinte (molto rigido, moderatamente deformabile e molto deformabile) a seconda della forma di sollecitazione caratteristica nei confronti della curva di deformazione. Il gruppo a deformabilità moderata è stato considerato come uno dei più adatti per l'impiego clinico.

Valutazione pertinente all'andatura di bambini con mielomeningocele
N. C. Carroll, D. Jones, W. Maschuich, M. Milner e C. White
Pros. Orth. Int. **6:1,** *27–34*

Riassunto
All'Ontario Crippled Children's Centre (OCCC) vediamo molti bambini affetti da spina bifida che necessitano di un trattamento clinico e di una riabilitazione lunghi e costosi..

Lo scopo della terapia ortopedica può essere meglio raggiunto tramite la valutazione efficace di ciascun paziente, che serve da base per la definizione di una terapia conservativa e chirurgica durante tutta l'infanzia.

Gli ausili chirurgici ed ortottici sono il miglior mezzo correttivo e preventivo contro l'instaurarsi di una deformità ortopedica. Attualmente la sola possibilità di una valutazione di un'ortesi consiste nell'attendere e nell'osservare se essa migliori la funzionalità o prevenga una deformità.

Questo rapporto esamina uno studio pilota intrapreso per illustrare i fattori che contribuiscono al progredire delle deformità agli arti inferiori. Il nostro scopo è stato di valutare questi fattori secondo una tecnica per quanto possible non invasiva ed afficaco dal punto di vista del costi; inoltre, le informazioni ottenute dovevano essere utilizzate per la valutazione e la messa a punto di un'ortesi per questi giovani pazienti.

L'obiettivo iniziale dello studio è consistito nella quantificazione dell'effetto indotto da un ortesi prescritta sulla deambulazione di ciascun paziente. Sono stati esaminati circa 15 bambini affetti da mielomeningocele a livello lombare o sacrale per un totale di 59 studi controllati. Il metodo di raccolta dei dati comporta la stesura di un protocollo trifasico da parte dell'ortopedico, del fisioterapeuta e del tecnico. Nel laboratorio di deambulazione dell'OCCC sono stati effettuati un esame clinico esauriente, una valutazione deambulatoria visiva su video-cassette ed una valutazione deambulatoria strumentale. A partire dai dati ottenuti sono stati proposti dei criteri i quali erano stati definiti dopo l'esame della distribuzione di frequenza di alcune caratteristiche deambulatorie nel gruppo esaminato. Viene illustrato un caso di applicazione dei criteri valutativi.

Trattamento dei problemi di cicatrizzazione nell'amputato con vascolarizzazione anomala
G. Horne e J. Abramowicz
Pros. Orth. Int. **6:1,** *38–40*

Riassunto
Questo studio illustra il trattamento

dell'amputato con vascolarizzazione anomala. Sono descritte le tecniche chirurgiche e postoperatorie che permettono di prevenire i problemi di cicatrizzazione. Si esamina quindi il trattamento dei pazienti con problemi di cicatrizzazione già instaurati.

Nota tecnica—Norme per l'esecuzione di prove di protesi di mano monofunzionali a comando mioelettrico proposta
B. Ingvarsson, I. Karlsson, L-G. Ottosson e M. Thyberg
Pros. Orth. Int. **6:1,** *41–42*

Riassunto
Per ottenere uno standard accettabile di protesi ed ortesi, in Svezia l'Institute for the Handicapped si occupa della sperimentazione di questa categoria di ausilii. Una parte importante del lavoro consiste nella definizione di norme di sperimentazione.

In risposta ad una richiesta da parte del Swedish Institute for the Handicapped, il Laboratory of Rehabilitation Engineering della Clinica Universitaria di Linkoping, Svezia, ha suggerito alcune norme per la sperimentazione tecnica di protesi di mano monofunzionale a controllo mioelettrico. Queste norme includono vari fattori di controllo e di esame importanti per la funzionalità delle protesi di mano.

Incidenza delle grandi amputazioni in seguito alla cancrena degli arti inferiori
T. Mandrup-Poulsen e J. Steen Jensen
Pros. Orth. Int. **6:1,** *35–37*

Riassunto
E' stata calcolata l'incidenza delle grandi amputazioni in seguito alla cancrena degli arti inferiori durante il periodo 1971–1979 nel distretto di Copenhagen. L'incidenza globale corrisponde a circa 0,3 per mille abitanti di età superiore alla quarantina. Il rapporto uomo-donna è di 2:1. Detta percentuale sembra crescere esponenzialmente con l'età. Il tasso di amputazione agli arti inferiori è rimasto immutato per un periodo di 8 anni.

Studio retrospettivo su 14.400 invalidi civili trattati per 25 anni
I. C. Narang e V. S. Jape
Pros. Orth. Int. **6:1,** *10–16*

Riassunto
Questo rapporto esamina 14.400 invalidi civili trattati per un periodo di 25 anni all'Artificial Limb Centre, Pune, India. Esso esamina in particolare la distribuzione per sesso e età, la causa dell'invalidità, i livelli di amputazione, le fonti di finanziamento ed altri fattori collegati alla riabilitazione dei pazienti.

Protesi degli arti superiori per amputazione di braccio
J. K. Ober
Pros. Orth. Int. **6:1,** *17–20*

Riassunto
Questo rapporto offre l'analisi del moncone residuo e dei movimenti della parte superiore del corpo ed i loro effetti sulle funzioni manipolatorie del paziente. E' descritta una protesi ibrida di braccio comandata a cavo singolo applicata a vari casi individuali. La struttura cinetica ed il controllo della protesi variano in ogni caso singolo a seconda della caratteristica manipolatoria del paziente. E' descritta una tecnica universale non convenzionale con varianti modulari dal punto di vista cinematico.

Analisi delle varianti cinematiche e cinetiche per la descrizione della deambulazione protesizzata con impiego del sistema enoch
K. Oberg e H. Lanshammar
Pros. Orth. Int. **6:1,** *43–47*

Riassunto
Gli schemi di deambulazione, le angolazioni articolari, le forze di reazione al suolo e i momenti articolari durante il cammino sono stati esaminati nei soggetti sani e negli amputati di gamba e di coscia.

L'analisi ha dimostrato che il diagramma dell'angolazione ginocchio-anca ed altri diagrammi simmetrici (per es. angolazione sinistra del ginocchio rispetto all'angolazione destra) forniscono un valido aiuto

interpretativeo delle anomalie dello schema deambulatorio. Si è concluso che un diagramma delle forze vettore-schema di deambulazione è utile nella valutazione dei momenti articolari.

Le forze di reazione al suolo ed i momenti muscolari alle articolazioni sono stati inclusi nell'esame. I momenti articolari al ginocchio erano molto diversi negli amputati sia di gamba che di coscia, in confronti ai soggetti sani.

Per la misurazione e l'analisi è stato usato un sistema detto ENOCH. Esso è costituito di un minicalcolatore collegato in linea alle apparecchiature di misurazione dello spostamento (Selspot) e delle forze di reazione al suolo (Kistler). Un terminale a computer grafico (Tektronix) à stato impiegato nella visualizzazione dei risultati.

Nuove articolazioni per le ortesi in plastica
H. Watanabe, T. Kutsuna, H. Morinaga e T. Okabe
Pros. Orth. Int. **6:1,** *21–23*

Riassunto

Le articolazioni plastiche per le ortesi sono migliori rispetto alle articolazioni metalliche. Esse sono leggere, silenziose, facili da portare, non arruginiscono, non sono soggette alla corrosione e sono radiotrasparenti.

Gli autori hanno messo a punto due tipi di articolazioni in plastica, uno per l'articolazione della caviglia e l'altro per il ginocchio, il gomito e l'anca. Il polipropilene è stato scelto per la realizzazione dell'articolazione grazie alle sue caratteristiche di flessibilità e di robustezza.

Prosthetics and Orthotics International, 1982, 6, 125–128

Calendar of events

National Centre for Training and Education in Prosthetics and Orthotics

Short-Term Courses and Seminars 1982–83

Seminars

NC 701 Positional Seating for the Disabled; 3rd November, 1982.
NC 702 Biofeedback and Behavioural Engineering; 19th January, 1983.
NC 703 Seating for Patients with Tissue Sensory Loss, 16th March, 1983.

Courses for Physicians and Surgeons

NC 101 Lower Limb Prosthetics; 15–19 November, 1982.
NC 102 Lower Limb Orthotics; 4–18 February, 1983.
NC 103 Introductory Biomechanics, Prosthetics and Orthotics, 8–12 November, 1982.

Courses for Physicians, Surgeons and Therapists

NC 501 Functional Electrical Stimulation (Peroneal Brace); 21–24 February, 1983.
NC 502 Upper Limb Prosthetics; 22–26 November, 1982.

Courses for Prosthetists

NC 205 Above-Knee Prosthetics; 7–18 March, 1983
NC 211 Patellar-Tendon-Bearing Prosthetics (Cuff and Supracondylar Suspension); 10–21 January, 1983.
NC 212 Hip Disarticulation Prosthetics; 6–17 December, 1982

Courses for Orthotists

NC 203 Knee-Ankle-Foot and Hip-Knee-Ankle-Foot Orthotics; 21–31 March, 1983.
NC 206 Upper Limb Orthotics; 24–28 January, 1983.
NC 207 Spinal Orthotics; 25 October–5 November, 1982.
NC 213 Ankle-Foot Orthotics; 6–17 December, 1982.

Courses for Occupational and Physiotherapists

NC 301 Lower Limb Orthotics; 28 February–4 March, 1983.
NC 302 Lower Limb Prosthetics; 29 November–3 December, 1982.

Courses for Prosthetic Technicians

NC 603 Above-Knee and Hip Disarticulation Modular Prosthetics; 31 January–11 February, 1983.

Further information may be obtained by contacting Prof. J. Hughes, Director, National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, George Street, Glasgow G1 1XW, Scotland. Tel: 041–552 4400.

Castle Priory College
Course Programme Autumn 1982

18/56 Disabled School Leavers—guidelines for careers teachers, careers officers, social workers and counsellors; 4–8 October, 1982.
18/58 Behaviour Modification, Observation Techniques and Goal Planning with "problem" children; 14–17 October, 1982.
18/64 Technology and Disabled Children; 30 October, 1982.
18/50 Extended course—Certificate in Residential Care of the Handicapped.
Information: The Secretary, Castle Priory College, Thames Street, Wallingford, Oxon. OX10 0HE England.

Oxford Study Days 1982/83

Sexual Problems of the Physically Disabled; 8–9 November, 1982.
Management of Congenital Foot Problems in Children; 24 November, 1982.
Wheelchair Seating and Pressure Problems for the Severely Disabled; 19 January, 1983.
Rehabilitation after Stroke; 31 January and 1 February, 1983.
Management of Motor Neurone Disease; 16 February; 1983.
Psychological Management of Physical Disability and Chronic Pain; 2 March, 1983.
Orthoses for the Hand; 30 March, 1983.
Information: The Secretary, Demonstration Centre, Mary Marlborough Lodge, Nuffield Orthopaedic Centre, Headington, Oxford.

October 1982
Disabled Child/Latin America
International Seminar on "Analysis of the Situation of Children and Young People with Disabilities in Latin America".
Information: Instituto Panameno de Habilitacion Especial, Apartado 11349, Zona 6, Panama.

13–15 October, 1982
Naidex '82. National Aids for the Disabled Exhibition and Conference, London.
Information: The Conference Officer, RADAR, 25 Mortimer Street, London W1N 8AB, England.

13–15 October, 1982
6th Annual Meeting of the American Society of Biomechanics, Seattle, Washington, USA.
Information: Peter S. Walker PhD., Programme Chairperson, American Society of Biomechanics, Orthopaedic Biomechanics, Brigham and Woman's Hospital, 75 Francis Street, Boston, Massachusetts 02115, USA.

Autumn 1982
Third International Forum on Lifelong Integrated Education.
Information: Y. Nomura, Director-General, Lifelong Integrated Education Center, Yoyogi 1–47–12, Shibuya-ku, Tokyo 151.

31 October–6 November, 1982
3rd Far East and South Pacific (FESPIC) Games for Physically Disabled to be held in Hong Kong.
Information: Joint Council for the Physically and Mentally Disabled. G.P.O. Box 474, Wanchai, Hong Kong.

November 1982
11th Congress of the International Diabetes Federation, Nairobi.
Information: International Diabetes Federation, 3/6 Alfred Place, London WC1E 7EE, England.

17–19 November, 1982
6th Annual International Rehabilitation Film Festival. To be held in the American Red Cross Building, New York.
Information: Film Festival 82, Rehabfilm, 20 West 40th Street, New York, NY 10018.

19–27 November, 1982
8th Congress of the International League of Societies for the Mentally Handicapped, Nairobi.
Information: Kenya Society for the Mentally Handicapped, P.O. Box 42365, Nairobi, Kenya.

9–10 December, 1982
ISPO—"Boerhaave" Congress (10th anniversary conference) (Dutch Language) Noordwjkerhout—Holland.
Information: ISPO—Holland c/o P.C. Prakke, v. Heemstralaan 51, 6814 KC Arnhem—Holland Tel 085-433495

1983

23–25 March, 1983
1st European Conference on Research in Rehabilitation, Edinburgh, Scotland.
Information: Dr. John Hunter, Rehabilitation Studies Unit, Princess Margaret Rose Orthopaedic Hospital, Fairmilehead, Edinburgh, Scotland.

4–18 April, 1983
7th Asia and Pacific Conference of Rehabilitation International, Kuala Lumpur, Malaysia.
Information: Malaysian Council for Rehabilitation, 12 Long Kongan Jenjarom, Off Jalan Klang, Kuala Lumpur, Selangor, Malaysia.

10–16 April, 1983
7th Rehabilitation International Asia and Pacific Regional Conference on theme, "Prevention and Rehabilitation: A Task for the Community, the Family and the Disabled Person," Kuala Lumpur, Malaysia.
Information: Malaysian Council for Rehabilitation, 12 Lengkongan Jenjarom, Off Jalan Klang, Kuala Lumpur, Selangor, Malaysia.

11–13 April, 1983
Annual Scientific Meeting of the American Spinal Injury Association, Denver, Colorado, USA.
Information: ASIA's Meeting Co-ordinator, Mr. Lesley Hudson, Shepherd Spinal Centre, 3200 Howell Mill Road, NW Atlanta, Georgia 30327, U.S.A.

14–15 April, 1983
International scientific meeting—Biomechanical measurement in orthopaedic practice, Nuffield Orthopaedic Centre, Oxford, England.
Information: J. D. Harris, Director, Oxford Orthopaedic Engineering Centre, Nuffield Orthopaedic Centre, Headington, Oxford OX3 7LD. Tel: 0865-64811 ext. 514/510

18–22 April, 1983
Occupational Therapy—Balancing Environment and Individual, Portland, Oregon.
Information: Betty Cox, COTA, ROH, AOTA. Director of Communications, The American Occupational Therapy Association, Inc. 1383 Piccard Drive, Rockville, Maryland. 20850.

8–13 May, 1983
10th World Congress on the Prevention of Occupational Accidents and Disease, co-sponsored by International Labor Office and International Social Security Association, Ottawa, Canada.
Information: Canadian Organizing Committee, 500–300 Slater Ottawa, Canada K1P 6A6.

2–5 July, 1983
15th Congress of the World Federation of Hemophilia, Stockholm.
Information: SCB, Jacobs Torg 3, S–11152 Stockholm, Sweden.

5–9 September, 1983
I.S.P.O. Fourth World Congress, London.
Information: Conference Services Ltd., 3 Bute Street, London.

13–18 November, 1983
2nd International Symposium on Design for the Disabled, Israel.
Information: Design for Disabled Secretariat, P.O. Box 29784, Tel Aviv 61297, Israel.

1984
15th World Congress of Rehabilitation International, Lisbon, Portugal.
Information: Rehabilitation International, 432 Park Avenue South, New York, New York 10016, U.S.A.
Conference of the International Federation of Physical Medicine and Rehabilitation, Jerusalem, Israel.
Information: International Federation of Physical Medicine and Rehabilitation, 9001 West Watertown Plan Road, Milwaukee, Wisconsin, U.S.A.

1985
7–13 July, 1985
14th International Conference on Medical and Biological Engineering, Helsinki, Finland.
Information: Dr. Hiilo Saranummi, Finnish Society for Medical Physics and Medical Engineering, P.O. Box 27, 33 231, Tampere 23, Finland.

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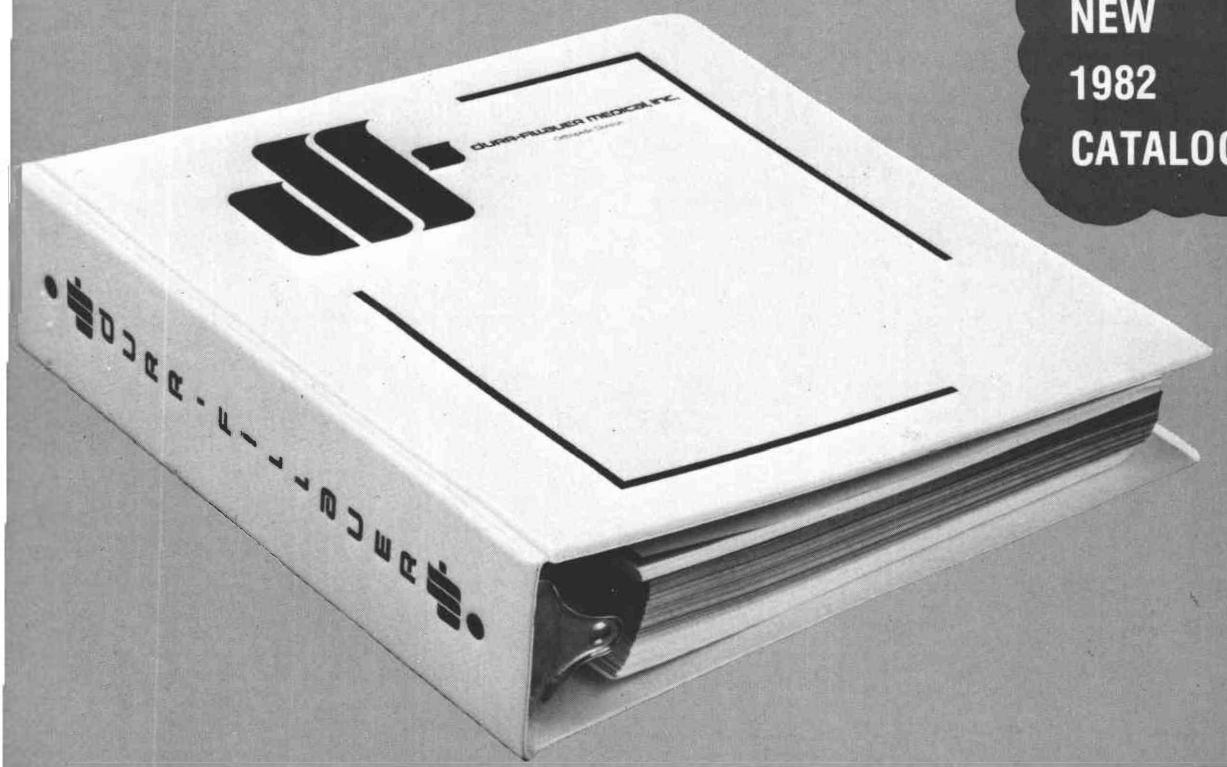

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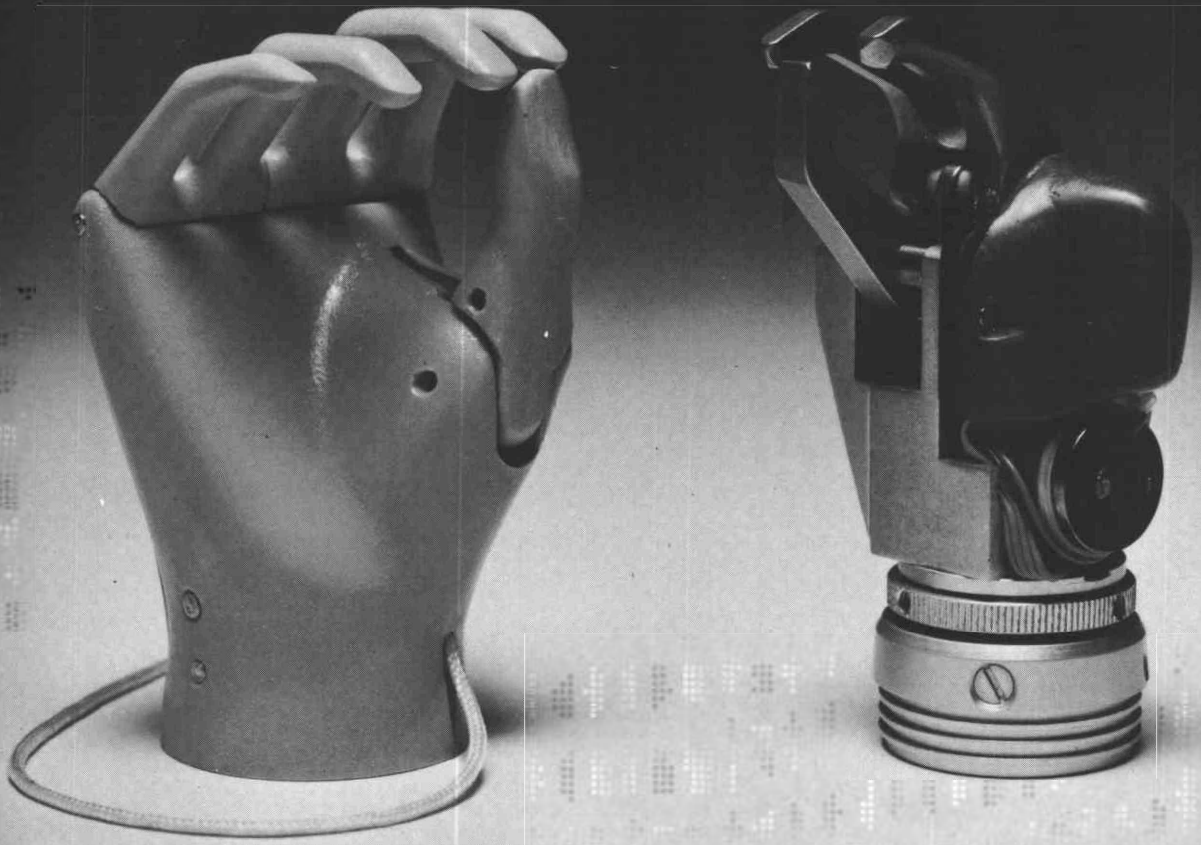

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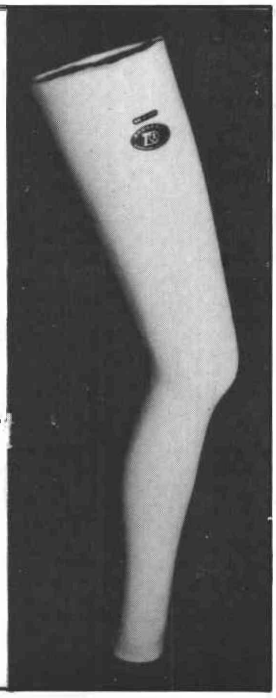

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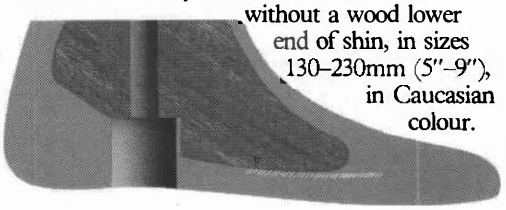

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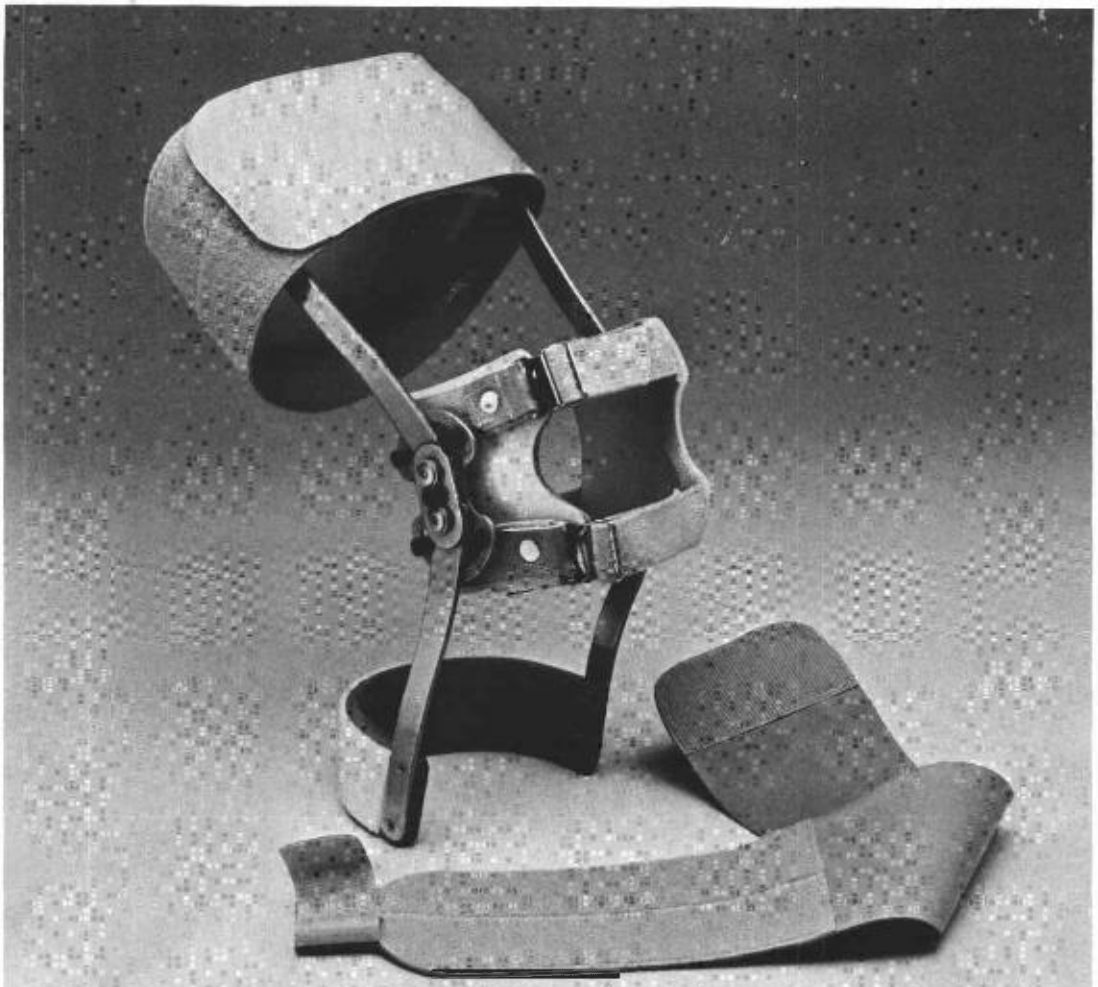

The Lerman Multi-Ligamentus™ Knee Control Orthosis.

The Lerman Multi-Ligamentus™ Orthosis controls knee motion with metal polycentric joints that closely simulate motion of the anatomical knee. The knee joints are completely adjustable and may be used to inhibit knee flexion or extension to any desired degree.

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