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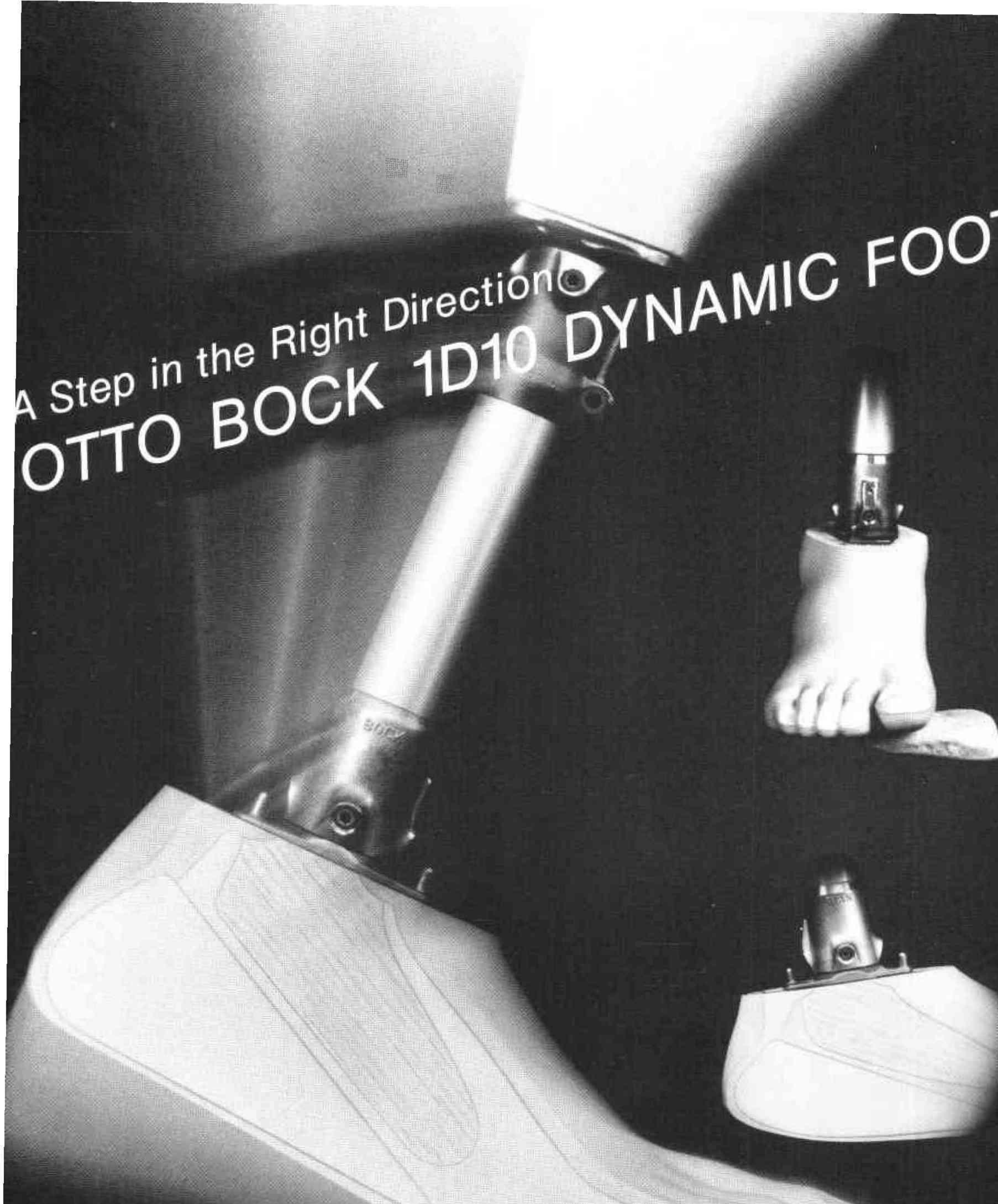

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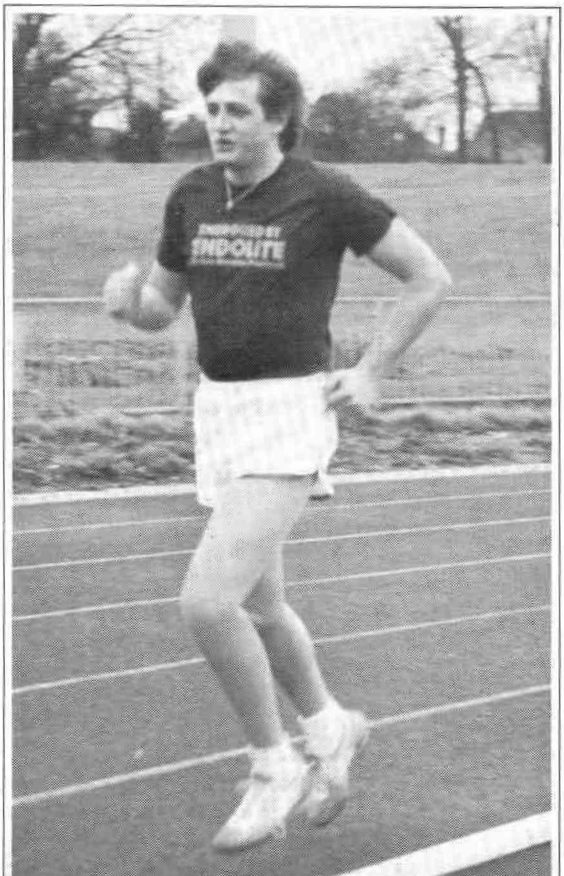

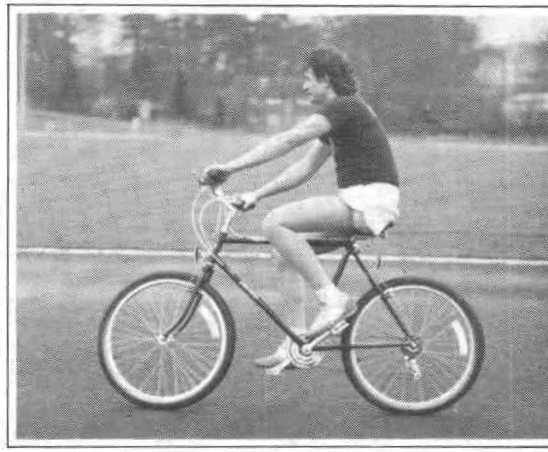

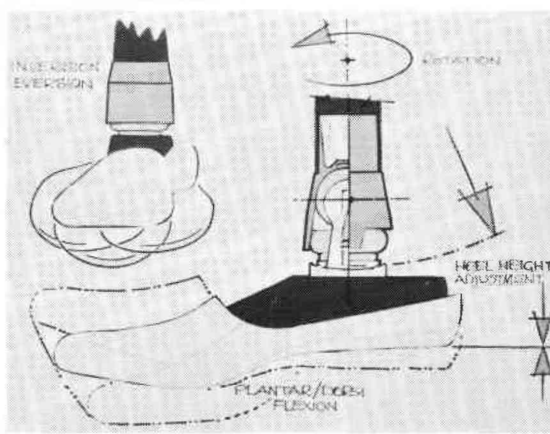

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Editorial

Evaluation—a fitting subject

At the risk of being trite, it seems obvious that when one has all the facts, making the proper decision usually presents no problems. Certainly facts are needed in order to develop and weigh the pros and cons of possible solutions to a given problem.

In no endeavour is this more obvious than in prosthetics and orthotics, where there are present so many factors involved in successful application and where only one or two of these factors can account for failure. The situation is compounded of course, because many of the factors are interdependent. Alignment is influenced by fitting, and fitting is influenced by alignment. The functional characteristics of the components used affect alignment; cosmesis can affect the attitude of the patient, which can affect performance, etc. Superimposed on this is the effect of the relationship between the patient and the prosthetist.

A prime example of a situation that requires decision making but where no one person has all the facts is the present plethora of above-knee fitting and alignment techniques that have been introduced and promoted somewhat vigorously during the past three or four years. In the United States three of the major university prosthetics education programmes were offering postgraduate courses in above-knee fitting techniques that differed markedly from the time-honoured quadrilateral above-knee socket and which differed to apparently a significant extent from each other. One of these same universities was also offering a course in application of the quadrilateral socket with a "flexible" socket. At the same time a group of private practitioners was promoting still more radical socket shapes with and without flexible walls.

To gather as many facts as possible as efficiently as possible, the ISPO Executive Board at its January 1987 meeting authorized the President and Teasurer to organize and conduct a workshop on above-knee fitting and alignment.

With the cooperation of the staff of the Prosthetics Orthotics Education Programme at Florida International University and the U.S. Veterans Administration a most successful Workshop was held at FIU, May 15–19, 1987.

Participation was by invitation only, and was composed of nine medical doctors, 31 prosthetists, eight engineers, and five individuals from other disciplines. All have had extensive experience in one or more aspects of prosthetics from basic research to education and provision of services. Eight countries were represented. Faculty from eight prosthetics education programmes participated. Many of the major prosthetics research programmes throughout the world were represented and approximately 15 private prosthetist practitioners were present.

The initial plenary session was started with a review of the history and principles of the quadrilateral socket. After each of the schools offering programmes in the newer techniques and a private group who were offering still another technique presented the details of their respective systems, the group was divided into six panels with the remit to identify the similarities and differences in the various techniques presented, explore how flexible walls might be used to the best advantage, and make recommendations for future work.

In the second plenary session each panel made a report on their respective deliberations and each recommended that some sort of evaluation programme be carried out.

The group was divided again, this time into five panels, and asked to recommend ways in which the evaluation might best be carried out.

On the concluding day each panel made a report and the meeting was opened for a general discussion.

Printed drafts of the panel reports are in the hands of the participants for corrections and the final versions will be used to develop a proposal for an evaluation programme that will provide the answers needed by clinicians, educators, and administrators in caring for above-knee amputees.

Meanwhile a comprehensive report of the workshop will prove helpful. Already, as a result of the meeting, the three universities in the US which have been offering different courses have taken steps to develop a more unified approach to teaching this newer technique which seems to have a very definite place in lower limb prosthetics. This step will of course be most helpful in carrying out an evaluation programme.

The workshop represents the first step in bringing order to the current practice of above-knee prosthetics, by first determining that the new techniques do indeed seem to have merit; by helping the education programmes to develop a more unified approach; and by setting the stage for an evaluation programme to complete the collection of the facts needed to apply these new techniques so that they serve the amputee in the best way possible.

It is hoped that this is but the first workshop of many to be sponsored, organized, and conducted by ISPO to help all who are concerned with prosthetics and orthotics to make the decisions needed in their work. Furthermore, it is hoped that the proposed international clinical evaluation project concerned with the current AK techniques will demonstrate to governments throughout the world the value of a broader evaluation programme.

A. Bennett Wilson Jr.

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Repeatability of kinetic and kinematic measurements in gait studies of the lower limb amputee

M. S. ZAHEDI, W. D. SPENCE*, S. E. SOLOMONIDIS and J. P. PAUL

Bioengineering Unit, University of Strathclyde, Glasgow

Abstract

During the last few years considerable attention has been given to the use of gait analysis as a tool for clinical use. The instrumentation for measurement of the kinetics and kinematics of human locomotion was originally designed for research use. Extension of its use into the clinical field calls for simplified methodology and clearly defined protocols with precise identification of the relevant parameters for the analysis. Force platforms, TV-computer and pylon transducer systems were used for collection of kinetic and kinematic data of five normal subjects, 10 below-knee, 10 above-knee and one hip disarticulation amputee. The repeatability tests showed significant differences in the measured parameters. These variations are attributed to the methodology of the analysis and the step to step variation of the subjects' gait. Differences in the degree of step to step variation between various amputee and normal subjects are quantified. In this presentation the capability of present day systems to perform repeatable gait measurements is discussed. A computational method for determination of representative measurements for the purposes of biomechanical evaluation and comparison as well as quantification of the degree of repeatability is described.

Introduction

The use of gait analysis for the assessment of several skeletal-neurological disorders, evaluation of the use of internal prostheses, measurement of effectiveness of orthotic devices, prescription of prosthetic components and fitting of lower limb prostheses is an inevitable and natural development of 50 years of research and development of instrumentation for the study of human locomotion undertaken at many centres throughout the world.

Since the beginning of contemporary gait studies by the University of California, commissioned in 1947, there has been an expansion in various parts of the world of development of instrumentation using modern technology and in planning long and short term programmes of research into the study of pathological and normal gait. Studies of biomechanics have allowed a new understanding of human locomotion particularly in respect of the forces developed at the joints of lower limbs and an indication of the proprioceptive feedback relating to position and velocity of the segments. The clinical application of gait analysis indicated the variability of the performance of normal and disabled subjects and highlighted the need for more repeatable and accurate measurements of kinetic and kinematic parameters. The use of such measurement facilities in the clinical situation assists the understanding of the gait process leading to identification and quantification of those variables which most accurately reflect the critical factors in gait of the disabled.

The most sophisticated methods of gait study have included force platforms for measurements of ground reaction forces, television/computer, infra red light sensing systems and cine photography systems in conjunction with passive and active body markers, for measurements of linear and angular displacements of limb segments. Additionally, studies have been performed to establish phasic muscular activity by utilizing EMG and metabolic energy cost by the use of respiratory gas analysis. Various computational techniques are used for data storage and reduction for the calculation of the position of joint centres and the loads developed there.

The accuracy and repeatability of the instrumentation used has been significantly improved. However the inaccuracies caused by the various assumptions made in calculations of

All correspondence to be addressed to M. S. Zahedi, Tayside Rehabilitation Engineering Services, Limb Fitting Centre, 135 Queen Street, Broughty Ferry, Dundee DD5 1AG, Scotland.

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joint centres and the repeatability of human movements and many other parameters have created a gap between the findings of the researchers in the field of human locomotion and the application of such findings in a clinical environment.

Repeatability of kinetic and kinematic measurements

Several factors may be identified which prevent the repeatable acquisition of gait data.

With regard to the instrumentation the leading commercial manufacturers of force platforms, Kistler and AMTI have reported the errors in signals from their load measuring devices as less than 2% with linearity errors less than 1% of full scale deflection. Similarly the TV/computer system marketed under the trade name "VICON" by Oxford Metrics for measurement of displacements is claimed to have an accuracy of 0.1% of the field of view. Other leading commercial systems in this field the CODA, SELSPOT and ELITE also claim similar or even better accuracy. Various individually designed instruments for load measurement such as the strain gauged pylon transducer for prosthetic load measurement and various kinds of kinematic measurement apparatus developed and used for gait analysis at research centres enjoy similar degrees of accuracy and repeatability.

The effects of computational analysis with fourth generation 16 and 32 bit computers have been quantified by several workers. Philippens (1981) examined the assumptions made in the analysis of data obtained from cameras in the study of motion and particularly the effect of accuracy of their placement and alignment. An error analysis was performed in the calibration, parallax correction and the use of three-dimensional direction cosine matrices for transformation of measured loads to various joint axes. Applying Philippens' analysis, a discrepancy of as little as two degrees in the alignment of camera and/or calibration frame, resulted in a typical calculation of antero-posterior hip joint bending moment being in error by 8% (The loads for these calculation were derived from a typical normal test performed as described in the methodology section).

Knowledge of the physical properties of the limb segments is a fundamental requirement in the study of human locomotion. The effects of various anatomical assumptions made in the calculation of the positions of joint centres have been reported by several workers. Various cadaveric studies have resulted in the development of constants for calculation of the positions of joint centres from the surface markers, and the calculation of mass and centre of gravity position of limb segments. A summary of the differences in the reported body segment parameters is given by Goh (1982). For the purposes of this presentation, the position of the hip joint centre was calculated with the aid of markers at the anterior superior iliac spines, sacrum, iliac crest, greater trochanter and the constants derived by Dempster (1955) and Ishai (1975). The estimated joint centre was then used for further calculation of antero-posterior bending moment at the hip. A difference of 12% was noted on using different derived constants.

There have been several publications quantifying the error produced by the use of passive and active body markers. These errors are in two groups. First the positioning of these markers and second the error caused by skin movement during the dynamic phase. In the latest reported work Andriacchi and Strickland (1985) have shown differences of 1.6 cm in positioning of an active body marker at the greater trochanter which is used for calculation of the hip joint centre. This resulted in a difference of 13.5% in the antero posterior bending moment at the hip. This effect varies at other levels (knee) and in other planes. Macleod of Oxford Metrics in a personal communication reported on the skin movements under passive markers during the swing and stance phases of normal subject locomotion. This effect has been illustrated by calculation of apparent differences of 3 to 4 cm in the length of the thigh segment.

Another major source of variation is the repeatability of the human gait itself. In quantification and identification of the individual source of variations it is important to measure the influence of each variable independently using controlled experimentation. Winter (1984) reported on the repeatability of the locomotion of normal subjects. A coefficient of variation as a measure of total variability was calculated. This parameter represented the root mean square of the standard deviation of the moment over the stride period divided by the mean of the absolute moment of force over the stride period. It was shown that on nine trial walks of one subject this coefficient of variability for ankle, knee and hip joint angle increased from nine to 10 and 19% respectively. Similarly for the axial load and shear force measurements, values of seven and 20% were reported. When moments at ankle, knee and hip in the antero-posterior plane were considered, the coefficient of variability increased from 22 (ankle) to 67 (knee) and 72% (hip) respectively.

The use of gait analysis has been reported in the evaluation of the use of orthoses in cerebral palsy children (Gage, 1983; Meadows, 1984), on evaluation of hip joint replacement (Kelly et al, 1983), on the influence of alignment on amputee gait (Zahedi, et al, 1987) and many other applications which require evaluation and comparison of individual gait patterns. However very few attempts have been made to describe the repeatability of measurements from one test subject. For the purposes of such comparison it is necessary to quantify the repeatability due to the method of measurements and step to step variation before attempting any biomechanical comparisons.

To show the step to step variations in normal subjects requires facilities such as long force platforms for measurements of ground reaction forces or devices such as load measuring devices inserted in the shoes or attached on the outside. Mobile camera systems or other systems or goniometers are required to acquire data accurately over several steps. Moreover the independent influence of anatomical assumptions, computational analysis, body marker positioning and skin movements have to be controlled and quantified. At present most clinical gait analysis laboratories have only conventional force platforms and kinematic measuring systems capable of handling data acquisition for one single step. It is therefore necessary to quantify the run to run variation from the selected measured steps. With regard to the prosthesis of a lower limb amputee, the effect of the anatomical assumptions, the positioning of the markers and skin movement do not apply. Further, the strain gauged pylon transducer incorporated in the shank of the prosthesis allows the monitoring of the three orthogonal forces and moments for an infinite number of successive steps constrained only by the testing environment. Furthermore, by measuring the alignment of the prosthesis it is possible to establish the position of the ankle, knee and hip relative to the pylon transducer and thus calculate the loadings at various levels. Thus it is possible to quantify the actual degree of step to step variation independently.

Methodology

Figure 1 shows the set up of the biomechanics laboratory at the Bioengineering Unit,

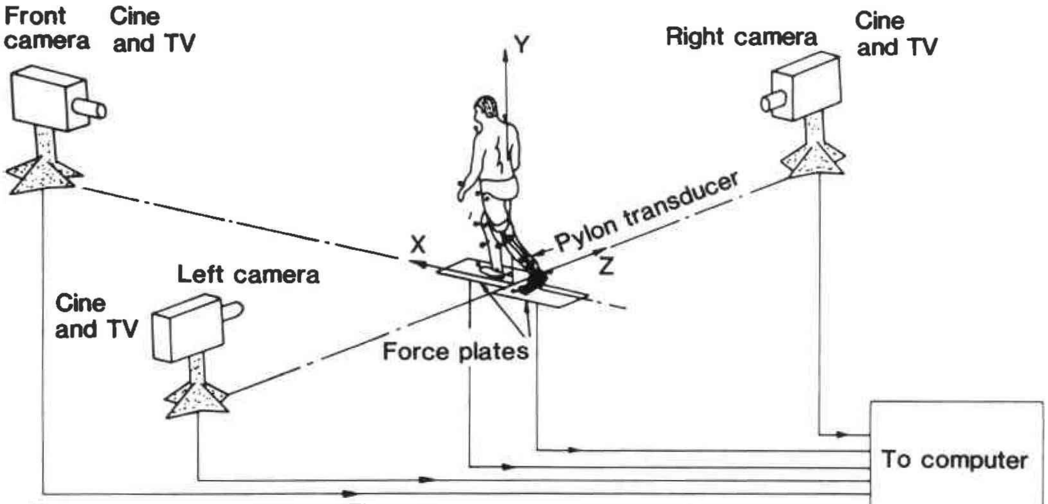

Fig. 1. The biomechanics laboratory.

University of Strathclyde, Glasgow, Scotland. The Kistler force platforms, and the Strathclyde Television Computer system were used for measurements of kinetic and kinematic parameters. Additionally the strain gauged short pylon transducer (Berme et al, 1976) as seen in Figure 2 incorporated in the shank of the prosthesis was used. A single axis goniometer measured the flexion angle of the prosthetic uniaxial knee joint. This angle in conjunction with alignment measurements, defining the relative geometrical position of the socket, the knee (as appropriate) and the foot of the prosthesis allowed the positions of the hip, knee and ankle relative to the transducer to be calculated. The Unit's PDP 11/34 computer was used for all data acquisition and data reduction.

Four categories of test subjects were considered as shown in Table 1.

With the exception of the last category of these subjects, the remainder were selected for participation in the programme of the study of alignment of lower limb prostheses undertaken at the Bioengineering Unit (Zahedi et al, 1986; Zahedi et al, 1987). The hip disarticulation subject was used for the study of the biomechanics of hip disarticulation. (Solomonidis et al, 1977).

As far as possible, standard prostheses and means of measurement were used to allow comparison of the results with those of other investigators. Otto Bock modular prostheses were used to allow easy change of alignment. The mass of the prostheses excluding the cosmetic covering but with the addition of the 500 g transducer was on average within ± 10% of the mass of the definitive prosthesis to which the amputees were accustomed. The prostheses were fitted to the amputees and dynamically aligned by the prosthetist. In each case two other prosthetists verified the fit of the socket and the alignment. Questionnaires completed by them were later correlated. An unsatisfactory collective rating of a fitting

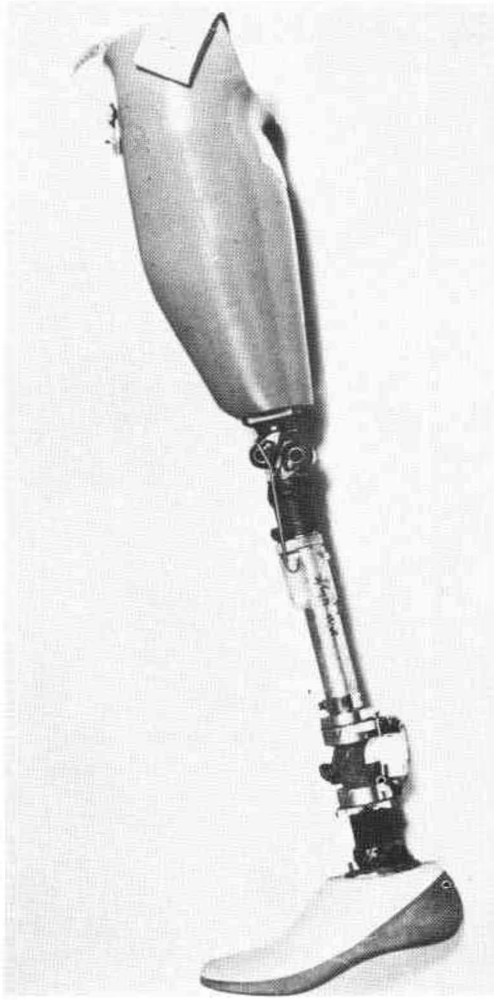

Fig. 2. Pylon transducer incorporated in the shank of an AK prosthesis.

Table 1. Details of subjects.

Subjects		Age		Years since amputation		Activity level*	
Type	No.	Mean	S.D.	Mean	S.D.	Mean	S.D.
Normal	5	36.2	± 9.51	—	—	—	—
Above-knee	10	46.1	±11.80	13.3	8.2	31.10	±13.43
Below-knee	10	55.33	±10.66	17.4	±10.5	37.5	±10.28
Hip Disarticulation	1	21	—	2	—	17	—

*Activity level determined according to Day (1981) (above +30 very active, below −40 inactive).

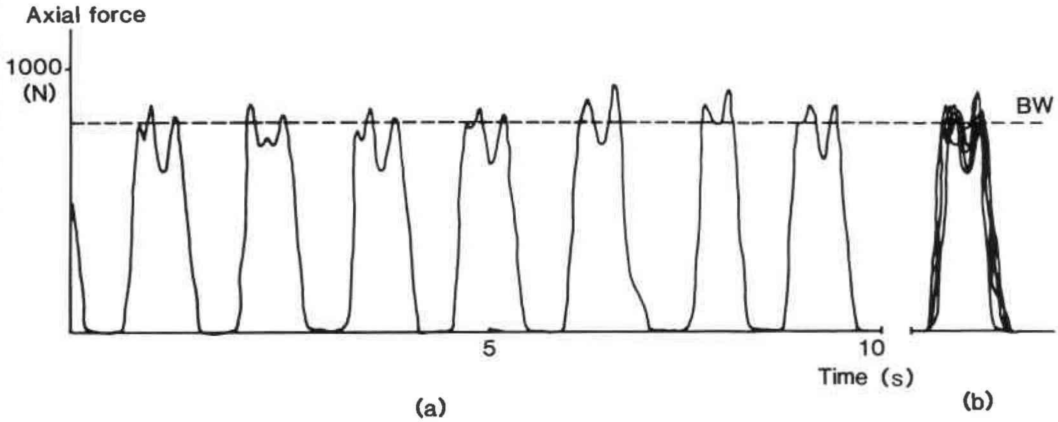

Fig. 3a. Prosthesis axial force for 9 successive steps by BK amputee; 3b, superimposition of cycles shown in 3a.

resulted in elimination of that test. A time of approximately one hour was given to the amputee to get used to the prosthesis and surroundings.

The force plate/TV data acquisition system was used for measurement of single steps of several successive runs for the normals and the sound and prosthetic side of the amputee subjects. The pylon transducer was used for prosthetic load measurement of successive steps on each run of the amputee subjects inside the laboratory. A series of Fortran programs allowed the sampling, filtering and calibration of data and transformation of loads to the ankle, knee and hip levels and the display of the results.

As the results from several successive steps and the selected steps from various runs showed different periods for stance and swing phase timing, it was not possible to simply average the data by the simple process of adding the signals without significantly distorting the picture. Simple normalization to the minimum time base also resulted in distortion of the amplitude of the signal (since the variation in the cycle time could be as much as 25%). Thus a technique was developed which firstly, automatically selected individual steps with the start, end and period of each step determined. Then a Fourier analysis technique was used to normalize the waveforms of different periods. Once the harmonics were determined, a new data sequence was generated based on a normalized period, without distorting the

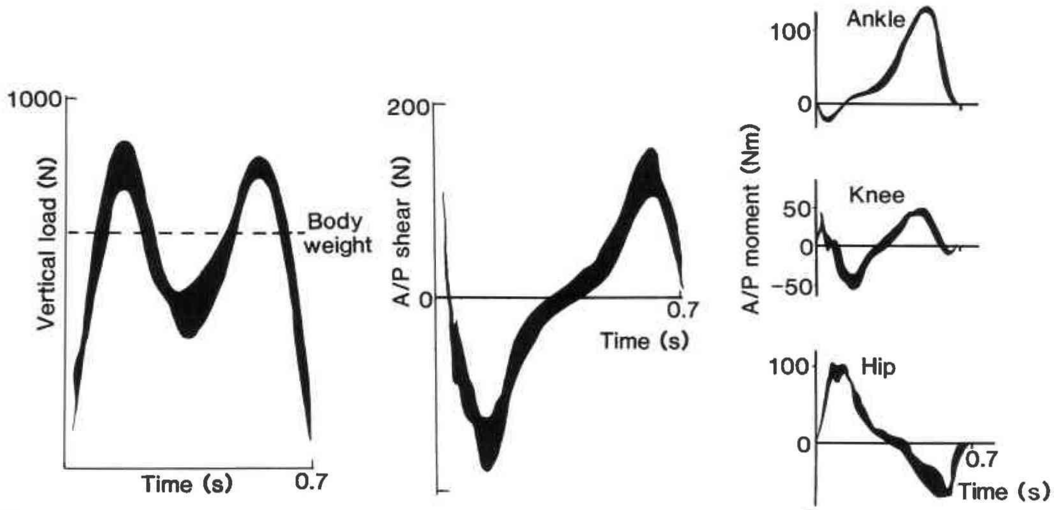

Fig. 4. Vertical load, AP shear force and AP moments at ankle, knee and hip—14 steps of a normal subject.

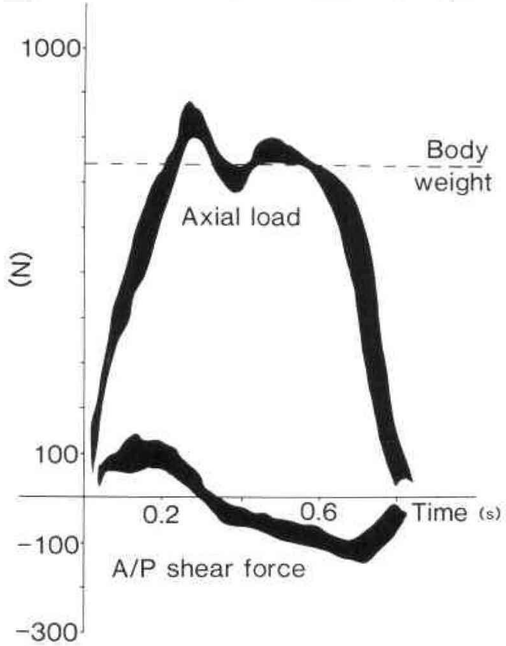

Fig. 5. Axial load and shear force of 14 steps by an AK amputee.

amplitude of the signal. The data can now be averaged for all successive steps of a run or the selected individual steps of several runs. The standard deviation is also quantified. The plot of the averaged data as the representative step for analysis and the standard deviation plot as the representation of the degree of repeatability is produced. The superimposition of the individual signals for each step develops an envelope which produces another form of representation of the degree of the repeatability. Figure 3 shows the axial loading of the prosthesis for a below-knee amputee: nine successive steps are presented, measured using the pylon transducer. The degree of step to step variation is shown by the thickness of the superimposed envelope of the individual signals.

Results

Figure 4 shows the curves of variation with time of load actions on the leg in the sagittal plane for a typical normal subject walking 14 times over the force platform. The composite curves are formed by superimposing the values for the fifth step in each of the 14 tests. The thickness of the envelope represents the variation between tests. These variations are due to step to step variation and skin movements under markers. Figure 5 represents the axial and shear force in the pylon frame of reference for an amputee using the same

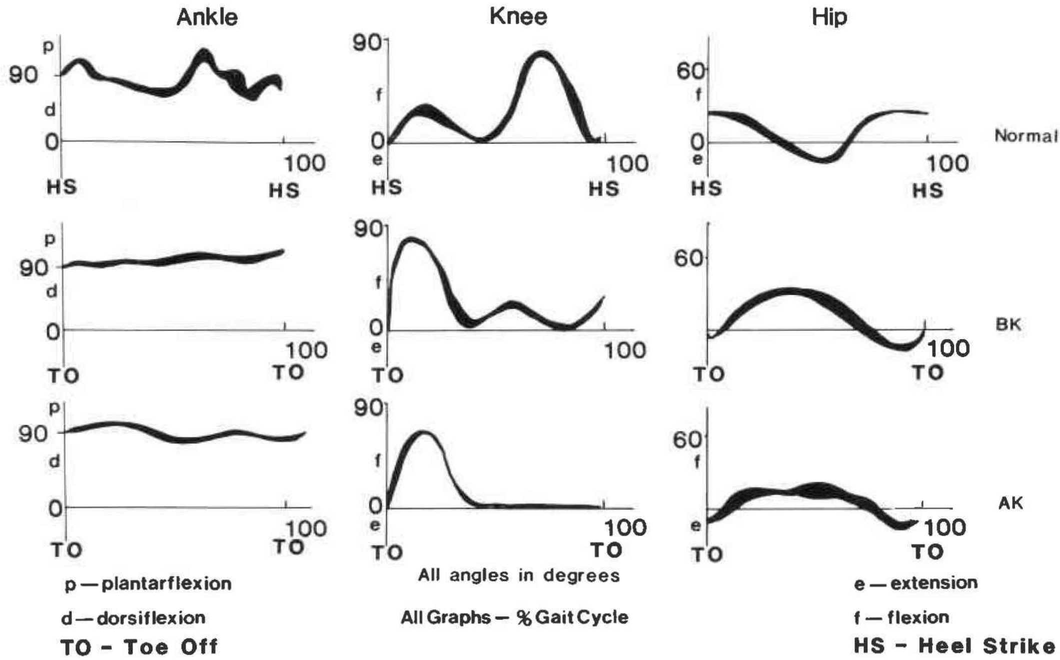

Fig. 6. Joint angle variation with time for 14 steps of a normal subject, an AK amputee and a BK amputee.

prosthesis over 14 steps. There are no skin movements and no errors due to positioning of the markers as the joints are mechanical and the socket reference land marks, used for measurement of alignment (Zahedi et al, 1986), are permanently marked. However there is variation corresponding to the step to step variability. The repeatability of kinematic data is illustrated in Figures 6 and 7, which show joint angle plotted against time and angle versus angle diagrams, for a normal subject, a below-knee and an above-knee amputee. Figure 8 shows the forces and moments measured by the pylon transducer incorporated in the shank of an above-knee prostheses for 60 successive steps on six runs. These signals are normalized and averaged using the previously described technique (Fig. 9). The standard deviation represents purely the degree of step to step variation (Fig. 10). The repeatability of

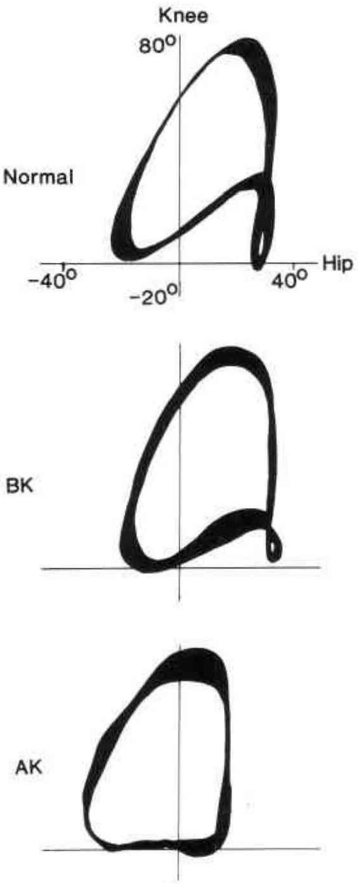

Fig. 7. Hip v knee angle diagram for 14 steps by a normal subject, an AK amputee and a BK amputee.

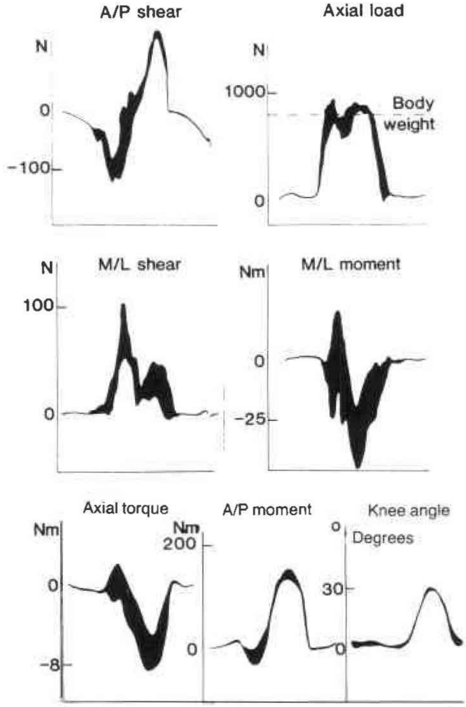

Fig. 8. Forces and moments of 60 steps by an AK amputee at pylon level.

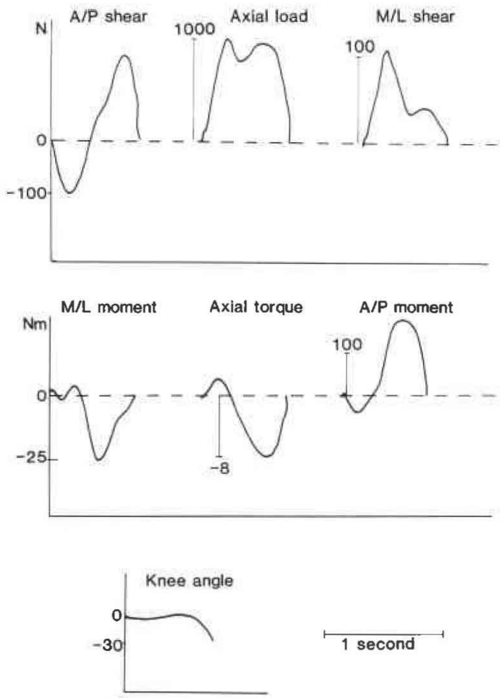

Fig. 9. Average of data at ankle level in shank frame of reference.

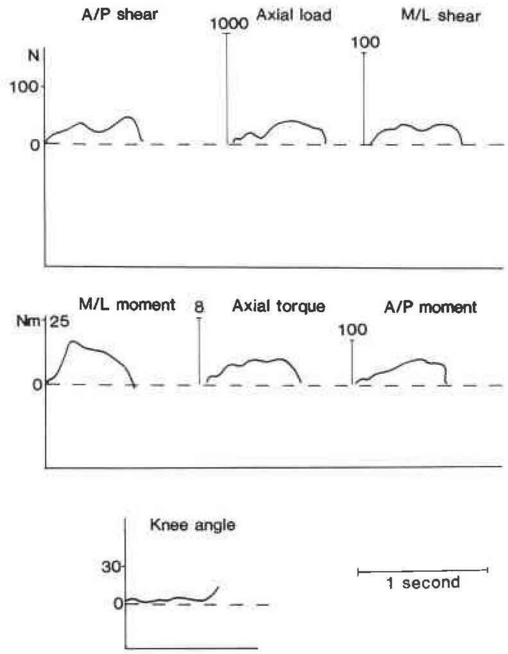

Fig. 10. Standard deviations of data (positive values only are shown).

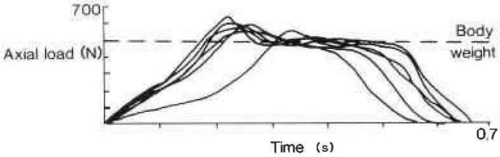

Fig. 11. Axial load in the prosthesis of a hip disarticulation amputee measured during 6 steps.

the axial load measurement for a normal subject is compared with the below-knee, above-knee and hip disarticulation data from Solomonidis et al (1977) in Figure 11.

The prosthesis for an above-knee amputee was dynamically aligned by three different prosthetists. Both the amputee and prosthetists were satisfied with the alignment and amputee's gait. Figure 12 shows the axial loading, the antero posterior, and mediolateral external bending moments at the ankle level. Figure 13 shows the comparison of the representative signal taking into account the step to step variation and the standard deviation graphs as the value for the degree of step to step variation for each condition.

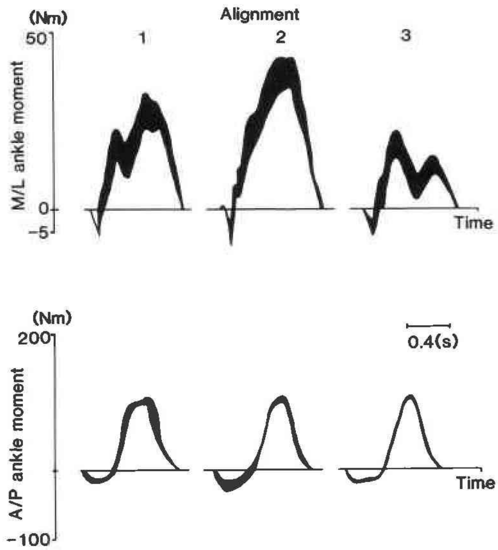

Fig. 12. Repeatability of AP and ML bending moment on 1 AK amputee with 3 different acceptable alignment configurations.

Discussion

The repeatability of the kinetic and kinematic measurements in the study of gait is determined by the errors produced in the instrumentation, techniques of data acquisition and the repeatability of the human gait. The errors caused by anatomical assumptions for location of the joint centres can be reduced by subjective selection of the reported constants (body segment parameters) and direct measurement where possible. The body segment parameters derived by using subjective averaging (Goh, 1982) provided better results. Where X-rays are possible direct anatomical measurements can be made reducing the errors

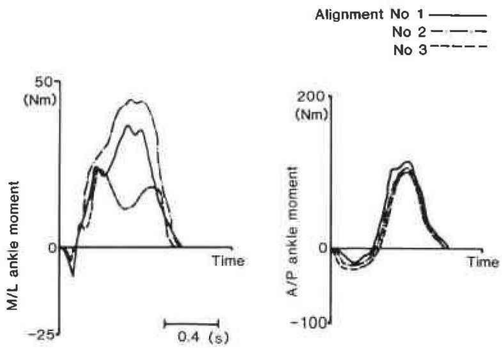

Fig. 13. Representative moments for 3 acceptable alignments.

even further (provided parallax is corrected). The errors caused by computational analysis to some extent have been overcome by the use of space calibration techniques as described by Marzan and Karara (1975) which take account of camera alignment and lens distortion by spatial calibration. The error in body marker positioning can be reduced to a very insignificant level by using small passive markers of 10 mm diameter assembled on a flexible base which can be accurately and repeatedly mounted on the anatomical land mark. The mass of a passive marker can be made very small (approximately 1g) made from solid foam. Mounting sites with least soft tissue under the skin can be selected. This method may require more computation of static and dynamic analysis than simple markers over joint centres. However by combining the above criteria, faster sampling of data (increase to 100 Hz from 50 Hz) and introduction of an averaging technique during the sorting of each marker position, the errors are reduced significantly.

The main source for the inability to produce repeatabie results is therefore demonstrated to be the step to step variation within several successive steps and the variation between the selected step of several runs. It was found that the best representative step for analysis, in a sufficiently long gait laboratory (15 m) which covers 15 steps of a normal subject is step number seven for force plate/TV data acquisition. Using video recording of the subjects' locomotion, the first and last three steps were found to be visually different from the rest of the steps.

The quantification of the repeatability of individual subjects also allowed the overall assessment of variation from one subject to another. Therefore it becomes important to have a measure of the repeatability for every subject tested, especially when biomechanical comparisons are made, or long term biomechanical changes in subjects' gait are to be monitored. This is crucial if small changes such as small differences between equally acceptable alignments in a lower limb amputee are being considered. It has been a conventional belief that there is a single optimum alignment for a prosthesis corresponding to 'best' gait performance. It has however been shown that, under the present method of prosthetic fitting the amputee could be satisfied with more than one alignment configuration (Zahedi et al, 1986). As can be seen, small differences in alignment influence the degree of repeatability as well as the pattern of the actual load actions, although the subject and the prosthetist are not aware of such differences and are satisfied with all alignments. From biomechanical considerations, alignment No. 3, which has also the least variation in loads from step to step in antero posterior bending moment and axial loading parameters seems better and nearer to the optimum condition. It is suspected that the amount of step to step variation is dependent on the degree of control during the gait. As equilibrium has to be maintained in each step, the reaction forces and moments measured indicate the required balance of forces and moments for spatial stability and equilibrium. Thus the subject has to correct for changes in forces, position and velocity using the proprioceptive feedback control for each step. This is seen as a variation from one step to another in the measured parameters. This could also be seen in variation of step to step changes in the normals and in further comparison with amputee subjects. A loss of a limb will reduce the degree of control of the musculo-skeletal structure and reduce proprioceptive feedback for the ambulation of the subject. Further this loss of control is greater in an above-knee amputee and even greater still in a hip disarticulation patient than in the below-knee amputee. The variation from step to step can be visually detected in the gait of a hip disarticulation amputee and to a lesser extent in above-knee amputees. However, visually it is very difficult to detect the step to step variation in all active below-knee amputees and it is almost impossible to detect this variation in normal subjects. Thus it can be concluded that for optimization of gait in an amputee it is necessary to quantify the degree of repeatability and to use the true representative signal for the biomechanical analysis.

Conclusion

In the use of gait analysis for the study of normal and pathological locomotion and for the purposes of clinical comparison, it is first necessary to quantify the degree of the repeatability due to the method of measurement and step to step variation, before attempting biomechanical comparison.

The degree of step to step variation varies for different normal subjects. This degree of step to step variation increases for the amputee population compared with normals. It appears that within the amputee population it increases further as the level of amputation becomes more proximal.

The degree of repeatability of kinetic and kinematic parameters measured at various levels increases with proximal increase in level from the ground.

A method is now available for quantification of step to step variation and reduction of the data to a representative signal for the purposes of biomechanical comparison.

Acknowledgements

The participation of all the test subjects is acknowledged. The prosthetic alignment part of this study was undertaken with financial support from the Scottish Home and Health Department. The assistance of Mr. Simon Scott in preparation of the diagrams is acknowledged.

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COMMERCIAL INFORMATION

AMTI, 141 California St. Newton, USA.
CODA 3, Technology Centre, Loughborough, United Kingdom.
ELITE, G. Ferrigno, A. Pedotti, via Gozzadini 7, Milan, Italy.
KISTLER INSTRUMENTE AG, Eulachstrasse 22, CH-8408, Winterthur, Switzerland.
SELSPOT, Selcom, Box 4032 Josered, Sweden.
VICON, Oxford Metrics, 7 West Way, Oxford, United Kingdom.

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Bondgraph modelling and simulation of the dynamic behaviour of above-knee prostheses

P. G. VAN DE VEEN, W. VAN DER TEMPEL* and J.DE VREISS**

Department of Mechanical Engineering, Twente University of Technology, Groningen

**Department of Rehabilitation, University of Groningen*

***'t Roessingh, Rehabilitation Centre, Enschede*

Abstract

A mathematical model was used to investigate the dynamic behaviour of an above-knee (AK) prosthesis in the swing phase and to analyse the influence of mass and mass distribution on the maximal stump load and the required energy. The model consists of a bondgraph model of the prosthesis and a "walking" model which predicts the walking velocity, step length and the femoral trajectory. Equipment was developed to measure the inertial properties of the components of the prosthesis.

Through computer simulation, stickdiagrams of the swing phase and graphs of the variation with time of the hip and stump forces were obtained. It was found that for a normal AK prosthesis with a knee-lock mechanism the axial stump load is greatest at the beginning and at the end of the swing phase. At a walking velocity of 5 km/hr the maximum axial stump load amounts to 2.1 times the static weight of the prosthesis.

The maximum axial stump force appeared to be almost directly proportional to the total mass of the prosthesis but independent of the mass distribution. The required energy also increased with the mass of the prosthesis but is dependent on mass distribution.

Because of their comparable weights the influence of the shoe is almost equal to the influence of the prosthetic foot. Thus lightweight shoes should be used with lightweight prosthetic feet in order to add to their advantages.

Introduction

Since the days of Ambroise Paré (1580), the history of the development of lower limb prostheses shows deep concern for the problems of mass and mass distribution. The famous prosthesis of Paré had a total weight of 7 kg and was made of steel. In the 17th century, wood and leather became the most important materials in prosthetics, later followed by aluminium. The success of the these light and strong materials can be concluded from their frequent use until this day. Because of the rapid development of modern plastics in the last half century, it is nowadays possible to make a complete AK prosthesis with a total weight of less than 2 kg. All manufacturers of prosthetic components are developing their lightweight prostheses, usually based on modern materials such as titanium, "aircraft" aluminium alloys and carbon fibre reinforced plastics. All these new designs satisfy some standards for the required strength and stiffness, e.g. the Philadelphia Standards (ISPO, 1978). It is remarkable that there are as yet no standards for the optimal mass of lower limb prostheses. The final mass is the result of the available materials, rather than conforming to well-understood design criteria. The aim of this research is to establish scientific criteria for the optimal mass and mass distribution.

In general there are two approaches:

Experimental It is possible to evaluate walking patterns of amputees wearing prostheses of varying mass. However, in this way it would be difficult to do serious experiments with prostheses with a much lower weight than the currently available types. Therefore, it was decided to approach the problem in a theoretical way.

Theoretical A method, was developed based on bondgraphs, to obtain mathematical models of the dynamic behaviour of lower limb prostheses. At this moment the authors have models for the swing phase of locked and unlocked AK prostheses. A model for the stance phase is in preparation.

All correspondence to be addressed to Ir. P. G. Van de Veen, Department of Mechanical Engineering, Twente University of Technology, P. O. Box 217, 7500AE Enschede, The Netherlands.

The subject of this paper is the model of the swing phase of a locked knee AK prosthesis. The influence of the mass and mass distribution on the resulting stump load and the required energy was investigated.

The model incorporates two submodels:
1. A bondgraph model of the prosthesis.
2. A model which describes the swing phase.

Bondgraph model of a locked-knee AK prosthesis

For an analysis of the dynamic behaviour the prosthesis was conceived as the physical system represented in Figure 1. The mathematical description is normally presented in differential equations. One of the main disadvantages of this approach is the impossibility of changing the system, e.g. by adding a knee control mechanism, without the laborious derivation of new system equations. Bondgraph description is a way to overcome this problem. In modelling and simulation with bondgraphs it is not necessary to derive equations. The physical system is expressed graphically and this graphical description is fed into a bondgraph-simulation-computerprogram. Because of the unfamiliarity of most readers with bondgraphs, the prosthesis-bondgraph will not be discussed further. Purely as an example of the compactness of this method, Figure 2 represents the bondgraph scheme of a 4-axial hydraulic damped knee mechanism.

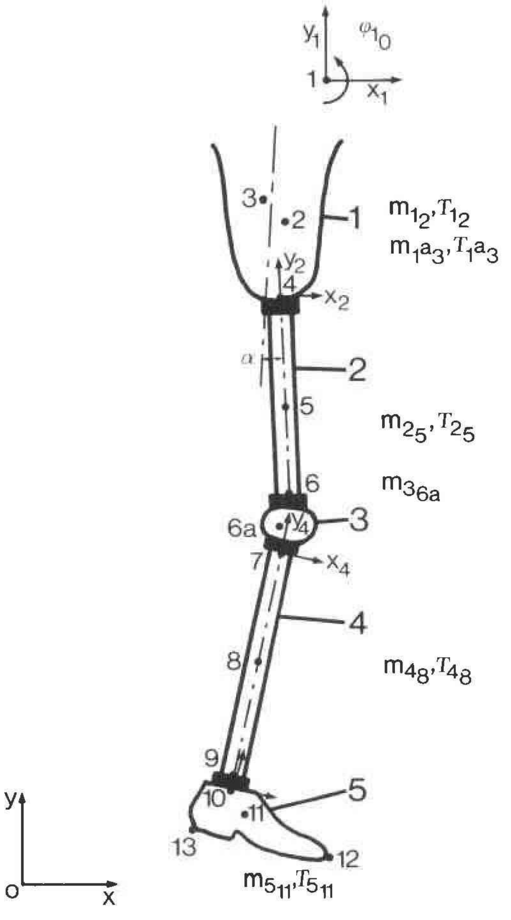


Fig. 1. The prosthesis conceived as a dynamic system.

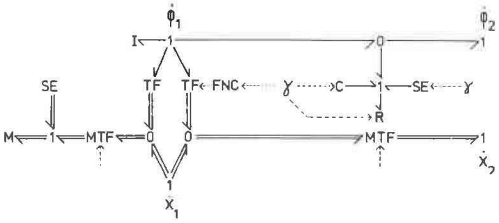


Fig. 2. Bondgraph of a four axial hydraulic knee.

Derived in a simple way, the bondgraph model is in fact identical with the set of Lagrange differential equations of the system. There are two ways to calculate the dynamic behaviour:

1. Using forces as input parameters and calculating the movements of the system. In practice this is rather difficult because it would be necessary to change the input forces until a normal swing phase was obtained.
2. Using the described movements of the prosthesis as input parameters and calculating the required forces, work and energy.

This second way was considered to be very useful. Of course it raises the question as to which movements are desired. The answer is the developed "walking"-model which predicts the optimal movements of a locked-knee AK prosthesis in swing phase.

The "walking" model

It would be possible to record (e.g. with Selspot equipment) the movements and walking patterns of amputees and to use these data as input for the prosthesis model. In this way it would be possible to calculate the required energy and the resulting stump load in particular cases. However, the aim was to investigate the influence of mass and mass distribution as a design criterion. Therefore, it was decided to use a model of "ideal walking". The calculated forces, stump load, work and energy must be conceived as values which are necessary to perform a normal walking pattern with an AK prosthesis. Of course, no amputee has a normal walking pattern and they would not produce these predicted forces, work and energy.

Figure 3 represents the swing phase from toe-off until heel contact.

The model incorporates the following experimental facts (Inman, 1980):

1. Swingtime in relation to the walking cycle frequency N.
2. Thigh angle at toe-off and heel contact in relation to N.
3. Total pelvic rotation.

The following assumptions were made:

4. The walking pattern is symmetrical and regular.
5. The rotation of the thigh and pelvis can be described with goniometric functions.
6. The trajectory of the femur can be approximated by a 3-degree polynome.

The model predicts:

7. The walking velocity, stride and step length in relation to N.
8. The trajectory and velocity of the femur as a function of time and in relation to N.

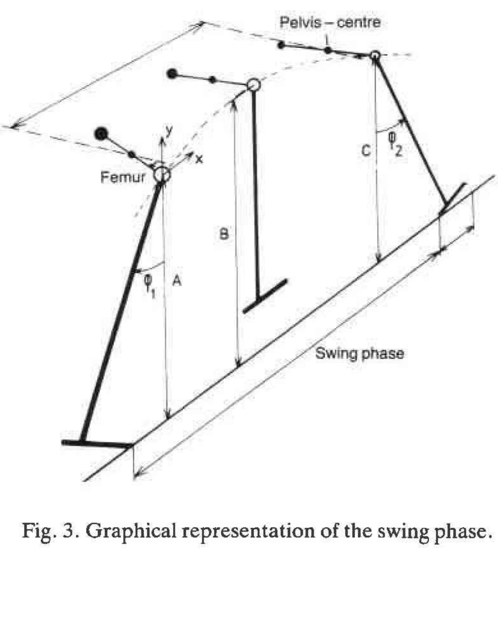


Fig. 3. Graphical representation of the swing phase.

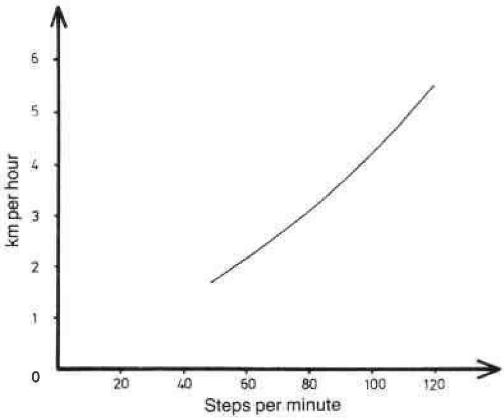


Fig. 4. Predicted walking velocity in relation to the step frequency.

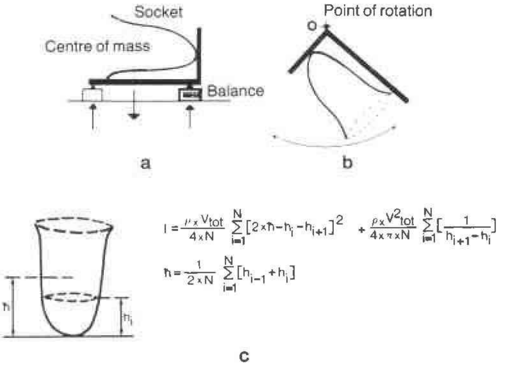
Verification of the "walking" model

Figure 4 represents the predicted velocity in relation to N. The walking velocity is too high at low frequencies and too low at the highest frequencies. However the overall inaccuracy of the model is less than 10%, which is rather good for a not very complex model. Improvements—better values for the parameters and extension from 2-dimensions to 3-dimensions—will raise the accuracy.

Parameters

To calculate the dynamic behaviour of the prosthesis, information is needed about the dynamic properties of the components of the system. Therefore, equipment was designed for measuring the first and second moments of inertia.

The first moment of inertia can be measured with a moment-equilibrium-table, illustrated in Figure 5a. With an oscillation time-measurement, the second moment of inertia can be established, as shown in Figure 5b. With these devices, the dynamic properties of all prosthesis components can be established with an error of less than 5%. Because the stump is part of the dynamic system, moments of inertia must be measured too. Obviously this is not possible with the equipment already mentioned. When it is assumed that the stump fills the socket entirely and an assumption is made for the relative weight of the stump (Drills et al, 1964) the mass and the moments of inertia can be calculated from the level heights when filling the socket with small equal amounts of water (Fig. 5c).



$$I = \frac{\rho \times V_{tot}}{4 \times N} \sum_{i=1}^{N} [2 \times \bar{h} - h_i - h_{i+1}]^2 + \frac{\rho \times V_{tot}^2}{4 \times \pi \times N} \sum_{i=1}^{N} \left[\frac{1}{h_{i+1} - h_i}\right]$$

$$\bar{h} = \frac{1}{2 \times N} \sum_{i=1}^{N} [h_{i-1} + h_i]$$

Fig. 5. Equipment for measuring the dynamic properties of the components of the prosthesis (see text).

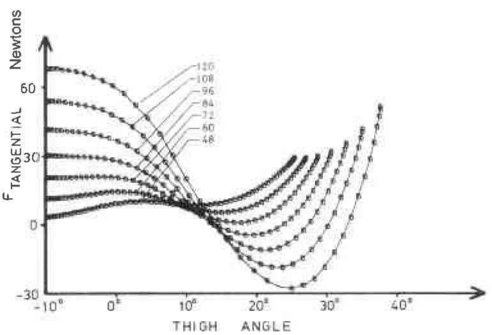


Fig. 7. Tangential hip force in relation to the thigh angle at seven step cycle frequencies.

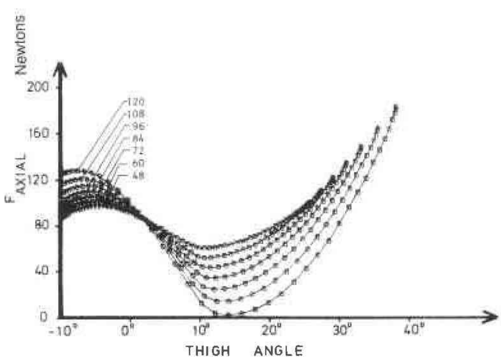


Fig. 8. Axial hip force in relation to the thigh angle at seven step cycle frequencies.

Results

Figure 6 shows stick diagrams of the calculated swing phase at two different walking velocities. The defined direction of the hip forces is also indicated.

The stick diagrams were obtained by simulation of the combined "walking" model and the bondgraph model, fed with the inertial properties of a normal AK prosthesis. This prosthesis consisted of a rather long socket, a Böck locked knee, pylon and uni-axial foot. The total weight of the prosthesis was 3.9 kg. Figures 7, 8 and 9, show the pattern of the hip forces which must be applied to the prosthesis to obtain this swing phase.

The axial force is particularly interesting; the tangential and rotational hip forces are rather moderate forces, but the axial hip force reaches high values. Therefore, the real axial stump force between stump and socket was investigated, acting as a shear force on the

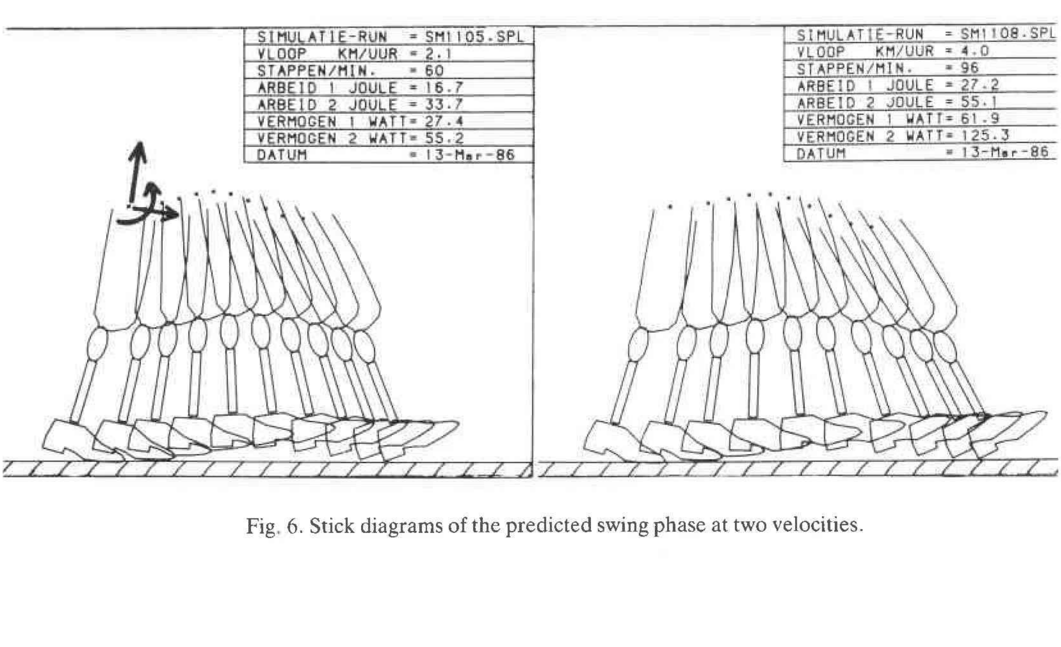


Fig. 6. Stick diagrams of the predicted swing phase at two velocities.

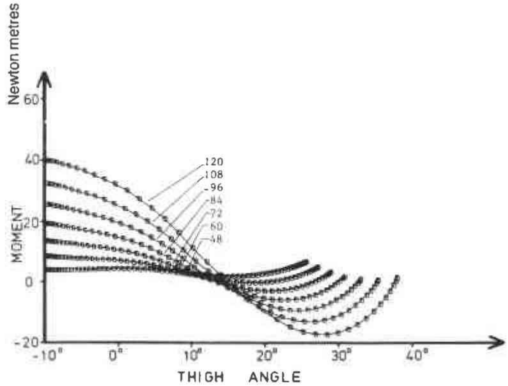


Fig. 9. Moment at the hip in relation to the thigh angle at seven step cycle frequencies.

stump's surface (Fig. 10). At higher frequencies, this force reaches values which are certainly not comfortable and probably impossible to endure.

To investigate the influence of mass and mass distribution on the maximal axial stump force, simulations were made for the already mentioned prosthesis, but also for the same prosthesis with a massless foot, a massless shoe, massless knee mechanism, massless pylon and a massless socket. In this way, the boundaries of "lightweight" prostheses were explored.

Numbering of the prosthesis and its variations:

Nr.	Weight in kg.	Remarks
1	3.9	Normal prosthesis
2	3.6	Massless pylon
3	3.3	Massless shoe
4	3.13	Massless knee mechanism
5	3.10	Massless prosthetic foot
6	2.5	Massless socket

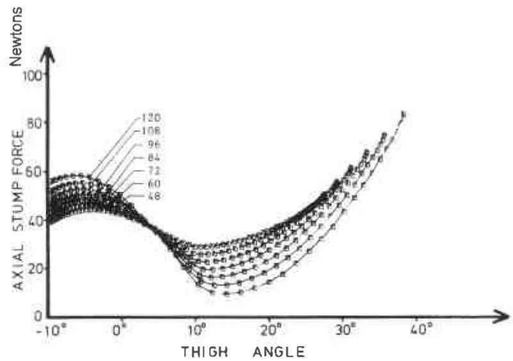


Fig. 10 Axial stump force in relation to the thigh at seven step cycle frequencies.

Figure 11 shows the maximum axial stump force that occurs in relation to the walking velocity for the prosthesis and its variations.

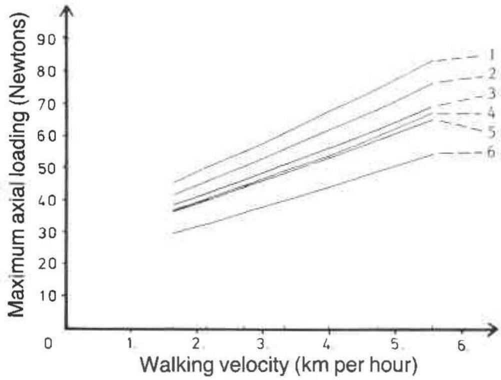


Fig. 11. Maximum axial stump force in relation to the walking velocity for an AK prosthesis with mass variations.

Some interesting results can be seen in this figure:

1. The maximum axial stump force is proportionate to the walking velocity.
2. The maximum axial stump force decreases almost linearly with the total mass of the prosthesis. Therefore, the lowest stump load in the swing phase of a locked knee AK prosthesis will occcur when the total mass is as low as possible.

   (This statement holds only for a locked knee prosthesis, because in the case of an unlocked knee, the femoral trajectory will be different).
3. The influence of the mass of the shoe is quite comparable with the influence of the mass of the knee mechanism and with the influence of the mass of the prosthetic foot. Therefore, lightweight prosthetic feet require lightweight shoes!

Figure 12 represents the total mechanical power put into the prosthesis in the swing phase, calculated for the prosthesis and for its variations, in relation to the walking velocity.

The figure shows the strong influence of the walking velocity on the required power in the swing phase. It can be seen that the required power is not completely linear with the total prosthetic mass, because of rotational effects.

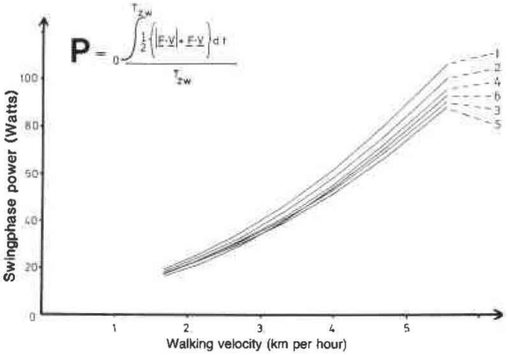


Fig. 12. Required swing phase power in relation to the walking velocity for an AK prosthesis with mass variations.

A massless shoe or a massless foot lead towards a lower required power than a massless knee mechanism, because of their greater distance to the point of rotation.

Discussion

1. The simple "walking" model gives rather good results for the walking velocity and the step length in relation to the step frequency.
2. The required power and the maximum axial stump force occurring in the swing phase of an AK prosthesis increases with the walking velocity.
3. The influence of the shoe is almost equal to the influence of the prosthetic foot on the maximal axial stump force and on the required power.
4. The required power and axial stump load decreases when the total mass of the prosthesis is decreased.
5. This theoretical approach proves to give useful information about the influence of mass and mass distibution on the swing phase of a locked AK prosthesis.

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The value of stump split skin grafting following amputation for trauma in adult upper and lower limb amputees

M. R. WOOD,* G. A. HUNTER,** and S. G. MILLSTEIN

Amputee Clinic, Ontario Workers' Compensation Board
** St. Joseph's Health Centre, Toronto*
*** Sunnybrook Medical Centre, Toronto*

Abstract

One hundred and twenty adult patients were reviewed in whom split skin grafts were applied to the stump following traumatic amputation of the upper limb (44 amputees) or lower limb (76 amputees). The average follow-up period was seven and a half years after initial amputation. There was delay in prosthetic fitting in all patients. Approximately one third of patients complained of occasional minor ulceration, controlled by removing the prosthesis for a few days or modification of the prosthesis. Further revision surgery, including excision of the grafted skin often combined with proximal bone resection, but not removal of the proximal joint, was necessary in 29% of below-elbow amputees and approximately 50% of below and above-knee amputees. At the above-elbow level, use of skin grafts allowed prosthetic fitting because of preservation of sufficient length of the stump. Despite the fact that revision surgery may often be necessary, split skin grafting has a definite place in the early management of the stump following traumatic limb amputation in the adult. Preservation of stump length with the knee or elbow joint allows easier rehabilitation and lower energy expenditure when using the prosthesis.

Partial foot amputation, when combined with skin grafting usually requires subsequent revision to a more proximal level to obtain a satisfactory result.

Introduction

Preservation of peripheral joints allows enhanced function following amputation. The below-knee amputee functions more efficiently than the above-knee amputee and the below-elbow amputee functions more satisfactorily than the above-elbow amputee. (Perry and Waters, 1981). Occasionally following trauma, the only way to preserve length is by the use of split skin grafts. The question arises whether the increased length of stump outweighs the disadvantages of a grafted stump. Thomson, Martin and Murray (1980) and Rosenfelder (1970) reported favourable results with grafting in lower limb amputations in children. The results in adults have been controversial. Harris (1981) stated that skin grafts of any type cannot tolerate the presence of a prosthesis. This opinion is shared by Hulnick, Highsmith and Boutin (1949) and Thompson and Alldredge (1944). Burgess (1981), Ascott (1954), Canty and Bleck (1952), and Dupertuis and Henderson (1946) do not agree. The authors' aim was to further study this problem. Both the early problems in the management and fitting of the skin grafted amputee and the long term problems with prosthetic wear, graft breakdown and revision surgery were assessed.

Materials and methods

One hundred and fifty-five patients with skin grafted stumps were seen at the Amputation Clinic of the Ontario Workers' Compensation Board.

Sufficient information was obtained following postal questionnaire, telephone interview and personal examination on 120 patients. The time the graft was performed, delay in fitting, subsequent prosthetic wear, minor ulceration and wound breakdown were recorded in all patients. All but two patients were men. All patients sustained the injury that led to their amputation in a work related

All correspondence to be addressed to Mrs. S. G. Millstein, Research Associate, Amputee Clinic, Ontario Workers' Compensation Board, Downsview Rehabilitation Centre, 115 Torbarrie Road, Downsview, Ontario M3L 1G8.

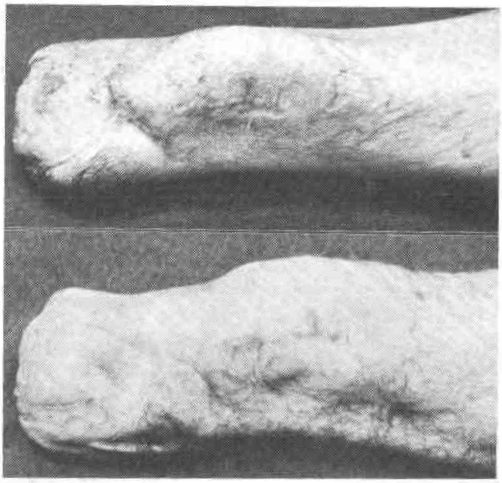

Fig. 1. Maturation of skin graft in below-knee amputation which saved length in a triple amputee (12 month interval between photographs).

accident. Only patients with a minimum follow-up of one year following their most recent operation were included in the study. The longest review was 21 years and the average follow-up was seven and a half years after initial amputation. There were 44 upper limb amputees and 76 lower limb amputees in the study.

Results

There was delay in fitting a prosthesis in all patients. The graft had to be sufficiently stable to withstand the shear stress of the prosthesis. Such maturation of a graft is shown in Figure 1, the time interval between the two photographs being 12 months. Most patients required modifications of the prosthesis including use of a temporary slip socket, incorporation of a thigh or ischial socket to offload the grafted stump in the lower limb amputee, and occasional fleece lined sockets were used in both upper and lower limb amputees to avoid friction on the skin graft (Bochmann, 1981). The delay in fitting a prosthesis ranged in the upper limb from six to 14 weeks following surgery and 10 to 26 weeks following surgery in the lower limb. The number of revisions required is shown in Table 1. None of the below-elbow and below-knee amputees required revision to a higher amputation level (Table 2).

Twenty of 27 partial foot amputations associated with skin grafts had to be revised (Table 3) and 14 were revised to the Syme (12) and below-knee (2) level. Although the partial foot amputation was preserved in 13 patients, skin grafts used in plantar or terminal aspects of the partial foot amputation resulted in unsatisfactory stumps because of skin graft intolerance and pain (Table 3 and Fig. 2).

Table 1. Revisions following skin grafts

Level	Number of patients	Revisions (%)
Below-elbow	34	10 (29%)
Above-elbow	10	1 (10%)
Partial foot	27	20 (74%)
Below-knee	26	13 (50%)
Above-knee	23	12 (52%)
Total	120	56

Table 2. Revisions to preserve proximal joint

Amputation level	Revisions	Revisions at same level to preserve proximal joint
Below-elbow	10	10
Below-knee	13	13
Partial foot	20	6

Table 3. Outcome of 27 partial foot amputees with skin grafts

Revision to level more proximal	12 2	Syme Below-knee
Local revision	6	(1 satisfactory)
No further revision	7	(1 satisfactory)

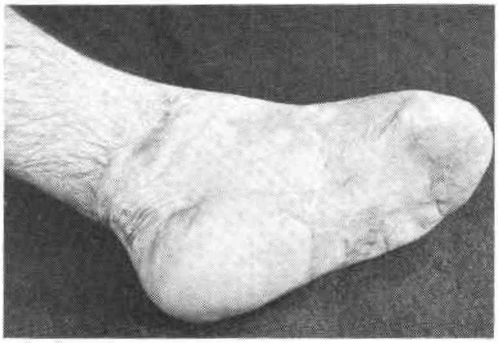

Fig. 2. Unsatisfactory skin grafting to plantar and terminal aspects of partial foot amputation.

Fig. 3. Successful revision following modification of a skin grafted stump allowing preservation of the knee joint (4 month interval between photographs).

Figure 3 illustrates successful revision of a skin grafted below-knee stump. Initial skin grafting followed by successful revision of the stump allowed preservation of the knee joint in below-knee amputees.

One third (40/120) of patients with split skin grafts complained of occasional minor ulceration at the junction of the graft with normal skin. This breakdown was often easily controlled by leaving the prosthesis off for a few days or making minor prosthetic adjustments. This minor problem did not detract from the use of a prosthesis and was considered by most users to be preferable to having a shorter stump. In the upper limb, eight above-elbow amputees were fitted with a prosthesis who would otherwise have had insufficient stump length for prosthetic fitting (See Fig. 4).

The results in Table 1 initially suggest that those who decry skin grafting and amputees are correct. There is a high rate of revision following skin grafting. Table 2 helps to clarify the situation. It shows that despite revision being necessary in a high proportion of below-elbow amputees and an even higher proportion of below-knee amputees, the subsequent revision still enables preservation of the proximal joint in all patients. However, only a small number of partial foot amputations were saved and these resulted in poor stumps due to skin graft intolerance and pain (Tables 2 and 3).

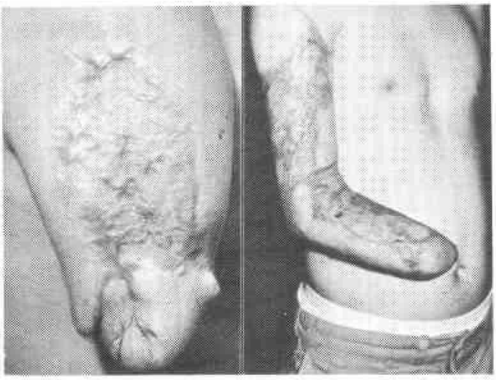

Fig. 4. Use of skin graft to preserve length in above-elbow amputation (left) and below-elbow amputation (right).

Often the skin graft could be totally removed by excision and primary suture following stump shrinkage and this two stage amputation left the patient with a satisfactory stump for prosthetic fitting (Fig. 5). The resulting stump was covered by normal skin with intact sensation. This situation is to be preferred to that which would arise if amputation had been done primarily at a higher level when either the elbow joint or knee joint would have been sacrificed.

Preservation of stump length with the knee or elbow joint allows better function with less energy expenditure to the patient. Similar

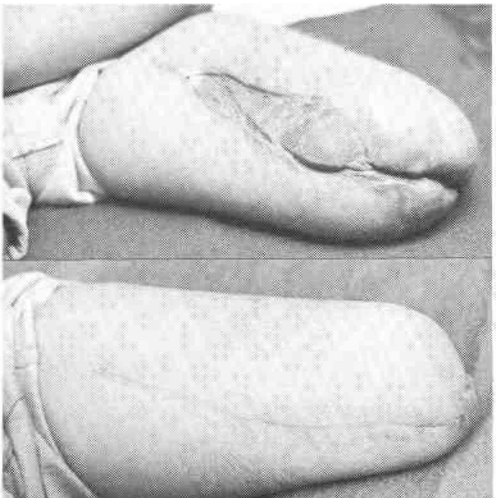

Fig. 5. Excision of skin graft area to preserve length and improve prosthetic fitting in above-knee amputation (4 month interval between photographs).

success has not been achieved with partial foot amputations and skin grafts should not be applied to plantar or terminal aspects of partial foot stumps.

In the adult traumatic amputee, split skin grafting has a definite place, and many patients benefit from this approach despite the fact that revision surgery may become necessary at a later date to provide skin with normal sensation, allowing the surgeon the ability to preserve the proximal joint above the amputation.

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Effect of anaerobic and aerobic exercise promoted by computer regulated functional electrical stimulation (FES) on muscle size, strength and histology in paraplegic males

P. J. PACY, R. H. EVANS and D. HALLIDAY

Nutrition Research Group, Clinical Research Centre, Harrow.

Abstract

The influence of anaerobic and aerobic exercise, promoted by computer regulated functional electrical stimulation (FES) was evaluated in four paraplegic males. Quadriceps muscle bulk was monitored by serial computerised axial tomography (CT) scanning and histology by muscle biopsies from the vastus intermedius. Anaerobic exercise consisted of 60 degree straight leg raising against increasing weights (range 1.4–11.4 kg) over a period of ten weeks. Aerobic exercise consisted of pedalling a modified Monark bicycle ergometer at 50 rpm against a fixed load ranging from 0–3/8 kilopond (0–18.75 watts) over a period of eight months. In both exercise studies the same work was not achieved by each paraplegic. FES was regulated by a closed loop system which is not presently commercially available, the frequency of the sequential muscle stimulator was 40 Hz with a pulse width of 300 μs.

Quadriceps muscle area of both legs increased 62.7% ($p<0.01$) after anaerobic exercise; similar but less pronounced effects followed aerobic exercise. Histologically two distinct patterns were noted from the outset, one had normal fibre type distribution the remainder had marked Type 1 loss. Both exercise regimens failed to change these although the number of internal nuclei per 100 fibres steadily increased (from 7.0% to 13.8% to 26.0%) as did the % of fibres with internal nuclei (5.4% to 10.5% to 25.7%) throughout the exercise periods. The significance of these observations is not immediately apparent but may signify continuing damage which may be due to the eccentric rather than the concentric nature of FES promoted muscular contraction.

All correspondence to be addressed to Dr. P. J. Pacy, Nutrition Research Group, Clinical Research Centre, Watford Road, Harrow, HA1 3UJ, United Kingdom.

Introduction

Maintenance of muscle mass requires continuous stimulation via the α-motor neurone. Disruption of such impulses that occurs following spinal cord trauma results in decreased muscle bulk, termed disuse atrophy.

It has been known for several centuries that isolated muscle preparations can be electrically stimulated to move. However clinical application of such a phenomenon has only relatively recently been developed which in part reflects advances in microcomputer technology which have made functional electrical stimulation (FES) a more practical proposition.

Preliminary evidence suggests that aerobic exercise, stimulated by computer regulated FES on a modified bicycle ergometer for 30 days increased endurance and average leg circumference by 1.9cm (Petrofsky et al, 1984). Bajd et al (1985) reported that a two-channel stimulator system regulating FES allowed practical crutch assisted walking in five individuals with incomplete spinal cord lesions.

These data are encouraging but the studies seem designed to evaluate the feasibility and perhaps practicality of FES rather than its influence on paralysed musculature.

It has been reported that in a study on ten paraplegics they displayed abnormal fibre type distribution below their spinal cord lesion with marked reduction of Type 1 (slow or red) fibres. In contrast above the lesion fibre type distribution was normal (Grimby et al, 1976). Comparable data have been published from animal studies. For instance cordotomy profoundly reduces Type 1 fibre numbers while increasing Type 2 (fast or white) fibres in guinea pig soleus muscle (Karpati and Engel, 1968). Interestingly Pette et al (1973) observed that in intact rabbits the composition of fast muscle could be altered to one resembling predominantly slow muscle by changing the pattern of electric stimulation.

The authors have therefore conducted a study in paraplegics to determine the influence of both anaerobic and aerobic exercise, promoted by FES, on muscle size, strength and histological appearance.

Subjects

Four paraplegic males who were involved in the *WALK Fund Project were studied and their clinical details are shown in Table 1. In addition to these details subject 2 was taking diazepam in a mean daily dose of 50mg which remained unchanged throughout the study. Only one (subject 4) exercised regularly prior to the study. This took the form of swimming three days per week for 30 minutes a session and again this was maintained constant.

Table 1. Clinical details of the four paraplegic men

Age (years)	27.3±6.4
Weight (kg)	67.9±4.5
Body Mass Index $\left(\frac{kg}{m^2}\right)$	20.5±0.5
Duration of paraplegia (years)	3.9±2.6

Figures=mean±SD

Methods

FES was administered via an as yet non-commercially available four channel stimulator. The pulse width was 300μs with a frequency of 40Hz. The electric stimulation used during anaerobic exercise ranged from 65–90 volts and for aerobic exercise 80–125 volts.

The muscle biopsy specimens were frozen in isopentane cooled to its melting point by liquid nitrogen. Cryostat transverse sections were stained at pH=9.4 to demonstrate myosin adenosine triphosphate activity so allowing identification of Type 1 and 2 fibres (Round et al, 1980). Sections were examined on a Magiscan image analysis system and fibre diameters calculated with an interactive computer program (Slavin et al, 1982). The fibre diameter represented the maximum distance across the lesser aspect of the fibre and when possible at least 100 fibres of each type were examined.

*WALK: Charitable organisation concerned with promoting the provision of locomotion systems for paraplegic patients.

Protocol

Each paraplegic was studied on three separate occasions: initially prior to any exercise, after ten weeks of anaerobic exercise and finally after eight months of aerobic exercise.

Anaerobic exercise consisted of straight leg raising by extending the knee through 60 degrees against a gradually increasing work load. An individual cycle lasted six seconds (three up and three down) followed by six seconds rest with the weight adjusted to enable the individual to exercise for 15 minutes. Both legs were subjected to the same protocol. The weights ranged from 1.4–11.4kg during the course of this exercise which was undertaken for five days per week for ten consecutive weeks. Aerobic exercise consisted of pedalling at 50rpm on a modified Monark bicycle ergometer. This had a high-back seat with seat belt and shoulder harness for postural support, if required. Cycling was performed for 15 minutes for five days per week for a total of eight months. Exercise was carried out against a fixed load ranging from 0 to a maximum of 3/8 kilopond (0–18.75 watts). As with anaerobic exercise identical work was not performed by the subjects. Maximum work achieved by subject 1 was 18.75 watts, subjects 2 and 3, 6.25 watts, and subject 4, 12.5 watts. Figure 1 shows the position of the electrodes used to stimulate the various muscle groups which enabled cycling. Figure 2 shows the modified bicycle ergometer.

Initially and at the end of each exercise regimen the following were performed; a single quadriceps muscle biopsy which was taken from

Fig. 1. Positions of the electrodes used for FES.

Fig. 2. Paraplegic subject on the modified bicycle ergometer.

approximately 5cm into the lateral mass of the muscle (Bergstrom, 1962) at the level of the femur via a UCH biopsy needle (Edwards et al, 1980). It was probable that fibres from the vastus intermedius were taken (Young et al, 1980). The same leg was used at each study.

Cross-sectional CT scans of both quadriceps muscle were taken and a representative picture is shown in Figure 3. Three separate cuts were taken although for the purposes of this paper only the one 6 inches from the lateral femoral condyle was used.

In addition to these investigations the paraplegics weight was recorded on a beam balance.

The protocol was approved by the Ethical Committee of the hospital and all individuals gave informed consent. Statistics were paired by t-test between the original results and subsequent ones.

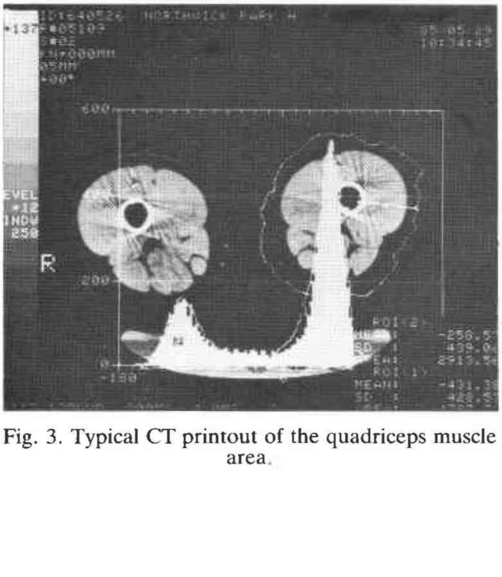

Fig. 3. Typical CT printout of the quadriceps muscle area.

Results

During the course of this study the size of the thigh obviously increased; this was more apparent following anaerobic exercise. However what was also noted was the friable nature of the biopsied muscle on all occasions. Histologically the paraplegics fell into two distinct categories; three (subjects 1, 2 and 3) having marked depletion of Type 1 fibres while the other had more normal distribution. These are shown in Figure 4. Neither of these patterns was influenced by the exercise. It was not possible to make any valid comment on the effect of exercise on Type 1 fibre number or diameter in the three paraplegics initially depleted while in the remaining individual these variables remained unchanged. However in all four, Type 2 fibre diameter increased progressively from $43.4\pm1.4\mu$ to $47.0\pm7.9\mu$ ($p=$ NS) to $61.3\pm8.5\mu$ ($p<0.05$). The number of internal nuclei per 100 fibres steadily increased from 7.0 ± 5.4 to 13.8 ± 5.4 ($p=$NS) to 26.0 ± 5.9 ($p<0.001$). Likewise so did the % of fibres with

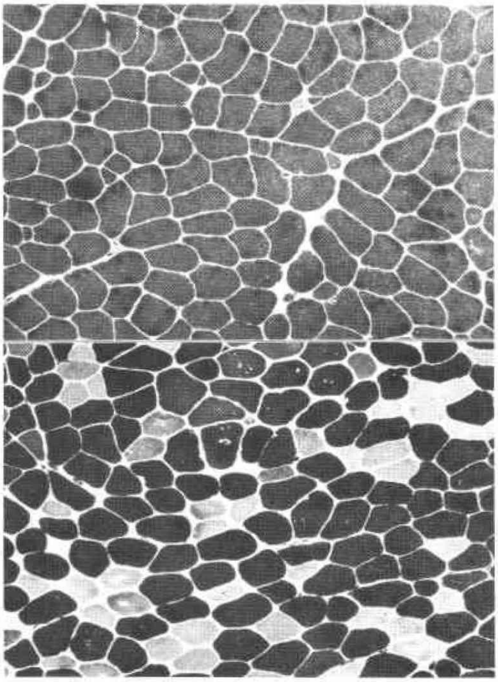

Fig. 4. Top, almost complete absence of Type 1 fibres. Bottom, a more normal fibre type distribution in quadriceps muscle.

internal nuclei from 5.4±3.8% to 10.5±3.7% (p=NS) to 25.7±3.1% (p<0.001).

The influence of both exercise regimens on total quadriceps area measured by CT scanning are shown in Table 2. Anaerobic exercise increased this area 63.5±23.3% (range 49.5–98.4%; p<0.01). In three paraplegics aerobic exercise resulted in a further but considerably smaller increase, 12.6±4.8% (range 8.5–19.3%). Subject 4 actually had a reduced area although this remained significantly larger than originally.

These paraplegics increased weight during the study from 68±5 to 70±4 to 73±4kg (p=NS) which was accompanied by a feeling of well-being.

Table 2. Changes in the quadriceps area of both legs assessed by CT scan following anaerobic (Ex 1) and aerobic (Ex 2) exercise.

Subject	BASAL	Ex I	Ex 2
2	2958	4528	5400
1	3224	6397	6943
4	3053	4671	4343
3	4337	6482	7161
Mean ± SD	3393 ± 639	5520* ± 1064	5962* ± 1334

Number=pixel unit *p<0.01

Statistics by paired t-test between basal area and those after each exercise. No statistical difference was found between exercise 1 and 2.

Discussion

This study has formally confirmed, by serial CT scans, preliminary observations that exercise stimulated by FES in paraplegics increases muscle bulk. This appeared more influenced by the shorter anaerobic exercise while the more prolonged aerobic activity had little additional effect on size. Both regimens increased the work performed by the paraplegics.

Although FES promoted exercise, increase in muscle bulk seemed to allow increased work, the maximum amount attained was considerably less than most non-trained, non-paraplegics would be expected to achieve (Edwards et al, 1977). Interestingly the level of work achieved appeared not to relate to the actual quadriceps area of both legs. Despite this there was a suggestion that the percentage increase of area was important with respect to work load. Subject 1, who had the greatest increase, approximately 115%, achieved the greatest work.

Histologically two patterns were noted from the outset of the study with three paraplegics having marked depletion of Type 1 fibres; the other had more normal distribution. These findings are somewhat at variance to data previously reported (Grimby et al, 1976). This study observed an almost universal absence of Type 1 fibres. It is possible that the minor differences between these findings may relate to the actual technique of muscle biopsy. It is known that fibre types are heterogenously distributed in muscle of mice (Susheela et al, 1968) and rats (Pullen, 1977). Likewise a study in man (Lexell et al, 1983) showed marked differences in fibre type from the vastus lateralis at various depths of the biopsy.

The study failed to demonstrate that fibre type distribution changed during FES promoted anaerobic and aerobic exercise. This is in marked contrast to findings in rabbits (Pette et al, 1973) and man (Munsat et al, 1976). This latter study showed that when muscle contracted isometrically Type 1 fibres increased in size (p<0.001) and percentage (mean increase SD=22.2±21.2%). In one individual a tenotomised muscle subjected to similar electric stimulation, and thus undergoing isotonic contraction, showed a different response. Type 1 fibre number decreased some 14.8% as did their mean diameter (p<0.001). Electrical stimulation was applied for far longer than the authors achieved which may be relevant to the differences in response between the studies. However no relationship was noted between the length of stimulation and degree of fibre type change (Munsat et al, 1976). In addition the type of electrical stimulation may be important (Hudlicka et al, 1982). Munsat et al (1976) employed intermittent stimulation at 33Hz while the authors used 40Hz via a sequential stimulator.

The most striking histological feature was the increase in the internal nuclei during the course of the study. The significance of this finding is unclear however. Certainly electrical stimulation is known to result in muscle fibre damage (Hudlicka et al, 1982). It has been suggested that the type of muscle contraction may be important with respect to possible muscle damage. Edwards et al (1984) postulated that eccentric muscular contraction which occurs during FES may be more deleterious than concentric contraction. The authors suggest that the increase in internal nuclei represents continued muscle damage although further work is required to confirm this. However despite such concern it has to be stressed that muscular work achieved did increase throughout the study.

In conclusion this study of computer regulated FES in paraplegic males shows measurable improvement in several variables of muscle bulk and performance. However future research needs to continue to refine this technique and to determine precisely the importance of the increase in internal nuclei before it is more widely used in the clinical setting.

Acknowledgements

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Limb perfusion in the lower limb amputee—a comparative study using a laser Doppler flowmeter and a transcutaneous oxygen electrode

S. L. E. FAIRS, R. O. HAM, B. A. CONWAY and V. C. ROBERTS

Department of Medical Engineering and Physics, King's College School of Medicine and Dentistry, London

Abstract

Accurate and objective assessment of amputation level in the lower limb plays an important role in patient management. Laser Doppler flowmetry (LDF) is a new and non-invasive technique for skin blood flow measurement and has been used pre-operatively in 25 patients undergoing amputation for vascular disease and in five normal controls. Baseline flux measurements were made at room temperature on the medial aspect of legs and then again after local heating of the skin for five minutes. Transcutaneous oxygen measurements were made at the same site for comparison and amputation level in patients selected on this basis.

Significant differences ($p<0.001$) in $TcPO_2$ values were found between controls (10.9 ± 0.5kPa), below-knee (BK) amputees (6.0 ± 1.5kPa) and above-knee (AK) amputees (1.5 ± 0.6kPa). Baseline LDF flux did not differ significantly between any group. Heated flux values did however show a significant difference ($p<0.005$) between controls (52.4 ± 23.5) and both BK (20.6 ± 9.2) and AK groups (8.1 ± 7.7) and also between the amputee groups. The relative increase in flux (heated flux/baseline flux) differed significantly between the BK (3.3 ± 1.5) and AK (1.2 ± 0.3) groups ($p<0.001$) and between these two and the controls (11.2 ± 5.4) ($p<0.001$). The correlation between relative increase in flux and $TcPO_2$ was 0.7 ($p<0.001$).

It is concluded that laser Doppler flowmetry used in conjunction with thermal stressing could provide a quick, simple and non-invasive method for objectively determining amputation level in the lower limb.

All correspondence should be addressed to Miss S. Fairs, Department of Medical Engineering and Physics, King's College School of Medicine and Dentistry, Denmark Hill, London SE5 8RX.

Introduction

In spite of advances in vascular surgery over the last decade, patients with degenerative vascular disease are still coming to major amputation when reconstruction is no longer feasible. In 1984, 85% of all new patients referred to the Artificial Limb and Appliance Centres (ALAC) in England and Wales suffered from vascular disease or diabetes and 78% were over 60 years of age (HMSO, 1985). In a society with an ageing population it is important that each amputation heals primarily, with preservation of the knee joint (Robinson, 1976) giving the patient the greater chance of successful rehabilitation on a prosthesis. Accurate assessment of amputation level is therefore of considerable importance in the management of the patient.

Clinical evaluation of the limb has been the standard way in which amputation level has been selected but it is highly subjective. Objective methods for the determination of amputation level that have been widely used in the last few years include clearance of radioisotopes (Malone et al, 1981) and transcutaneous oxygen measurements (Burgess et al, 1982). The former technique is often regarded as the "gold standard" because it provides a quantitative measure of skin blood flow, whereas the adequacy of the blood supply is only implied by measurement of percutaneous oxygen tension.

Laser Doppler flowmetry is a relatively new and non-invasive technique for measuring blood flow in the microcirculation of the skin (Nilsson et al, 1980). Doppler shifts in laser light produced by moving red blood cells are detected and electronically processed to provide a measure of blood flow, termed flux. This flux is the product of the concentration of cells in the measurement volume and their average velocity. Since the penetration depth of helium-neon laser light in the skin is of the order of 0.6mm only flow in the microcirculation is detected (Anderson and Parrish, 1981).

Radioisotope clearance and transcutaneous oxygen measurements are both time consuming and somewhat complex methods of determining skin blood flow. Laser Doppler flowmetry would appear to offer a quick and simpler alternative. The authors have accordingly attempted to evaluate this new instrument by comparing its use alongside transcutaneous oxygen measurements made in a group of patients proceeding to lower limb amputation and in a group of normal control subjects.

Patients and methods

Transcutaneous oxygen measurement ($TcPO_2$) and laser Doppler flowmeter (LDF) flux values were obtained pre-operatively on 25 patients undergoing lower limb amputation for vascular disease and from five healthy subjects with no evidence of vascular disease.

All measurements were made with the subject in a semi-recumbent position after allowing at least 15 minutes for them to equilibrate. Sites on the medial aspect of both legs, 13cm from the knee joint line and 5cm from the tibia were used for both instruments. A measurement at a lateral site, 13cm from the knee joint and 3cm from the tibia was also made with the LDF. Both LDF and $TcPO_2$ measurements were made sequentially on identical sites. Baseline flux values at room temperature were first obtained from the LDF, (Periflux PFId) followed by readings made after 5 minutes local heating of the skin at 42°C (the maximum possible using the instruments integrated heating element). A time constant of 3 secs was selected on the LDF and a frequency cut off of 4kHz, settings which are optimum for obtaining mean, rather than pulsatile readings. The more slowly varying mean flow is easier to interpret, unlike the pulsatile signal which exhibits a "hunting phenomenon" in critically ischaemic skin. This phenomenon is variable between patients and difficult to quantify.

The LDF evaluation was followed by a $TcPO_2$ measurement (Roche $TcPO_2$ monitor 632) in which the heating element temperature was set at 45°C, the temperature at which previous work in determining critical levels for amputation with the $TcPO_2$ has been done. Measurements of $TcPO_2$ were made when the temporal variations in measured level were within 0.1kPa. This usually occurred about 15 minutes after placement of the electrode on the skin. Amputation levels were selected by the referring clinician on the basis of the $TcPO_2$ readings using pre-determined levels (Butler, 1985). A transcutaneous oxygen tension of less than 2.7kPa is taken as indicating above-knee amputation. By following the LDF measurements by the $TcPO_2$ estimations the procedure overcomes the possible errors caused by heating to 45°C just prior to heating to 42°.

$TcPO_2$ measurements, LDF baseline and heated flux values were expressed as mean ± standard deviation. The relative increase in LDF flux produced by heating was calculated (heated flux/baseline flux) and was similarly expressed.

Apparent differences in the mean values of all measured and derived parameters, between the controls, BK amputees and AK amputees were analysed for statistical significance using the Student's t-test for unpaired data.

Results

All amputations performed in this small series healed primarily at the level chosen. Transcutaneous oxygen and LDF values for each group are presented in Table 1. Significant differences exist ($p<0.001$) between the $TcPO_2$ measurements of each group (This would be expected since the amputation level was largely

Table 1. LDF flux and $TcPO_2$ values measured on medial and lateral aspects of the legs of controls (C), BK amputees and AK amputees. Mean values are given ± one standard deviation.

	Medial Site			Lateral Site		
Variable	C	BK	AK	C	BK	AK
Number	5	14	11	5	13	10
Baseline flux	5.4± 1.5	6.9±3.6	7.0±6.3	5.8± 2.9	6.4± 4.4	5.0±3.4
Heated flux	52.4±23.5	20.6±9.2	8.1± 7.7	47.3±12.3	22.8±15.8	5.5±3.6
Relative increase in flux	11.2± 5.4	3.3±1.5	1.2± 0.3	9.9± 4.9	4.5± 4.2	1.2±0.3
$TcPO_2$, kPa	10.9± 0.5	6.0±1.5	1.5± 0.6	—	—	—

determined on the basis of this measurement).

The baseline LDF flux values for the controls, BK and AK patients were not found to differ significantly. However, a significant difference existed between the heated flux of the control and both the BK and AK groups ($p<0.001$). The difference between the heated flux of both amputee groups was also significant ($p<0.005$).

A more significant difference between the BK and AK group was found when the relative increase in flux was considered ($p<0.001$). The difference between the controls and BK or AK groups using this parameter was also significant ($p<0.001$). Figure 1 illustrates the correlation between $TcPO_2$ and the relative increase in LDF flux ($r=0.7$, $p<0.001$).

Discussion

Laser Doppler flowmetry is a recent addition to the range of techniques available for microcirculatory measurements. It has been previously used in assessing skin flap viability (Jones and Mayou, 1982), muscle blood flow (Oberg et al, 1979) and the evaluation of peripheral vascular disease (Karanfilian, 1984). Good agreement between standard skin blood flow measurement techniques and laser Doppler flux has been demonstrated (Watkins and Holloway, 1978; Stern et al, 1977). Although LDF cannot provide a quantitative measure of skin blood flow, comparisions of flux values are possible enabling stimulus/ response type experiments to be performed.

As would be expected, skin oxygen levels are related to blood flow (LDF flux). Figure 1 shows that $TcPO_2$ rises asymptotically towards the normal levels which for patients over 60 is typically 10.6 kPa (the 95% range extending from 8.32 to 12.84 kPa). Figure 1 shows the correlation between relative increase in flux on heating as a function of $TcPO_2$ and demonstrates how sensitive $TcPO_2$ levels are to relatively small changes in vascular reactivity. Relative flux was plotted rather than absolute flux as the authors' experience in monitoring reconstructed vascular surgery has indicated that vascular reactivity is likely to be a better indicator of potential for healing than absolute flow level. In fact the correlations are very similar though marginally poorer for the absolute heated flux/$TcPO_2$ characteristic.

Support for this approach comes from the

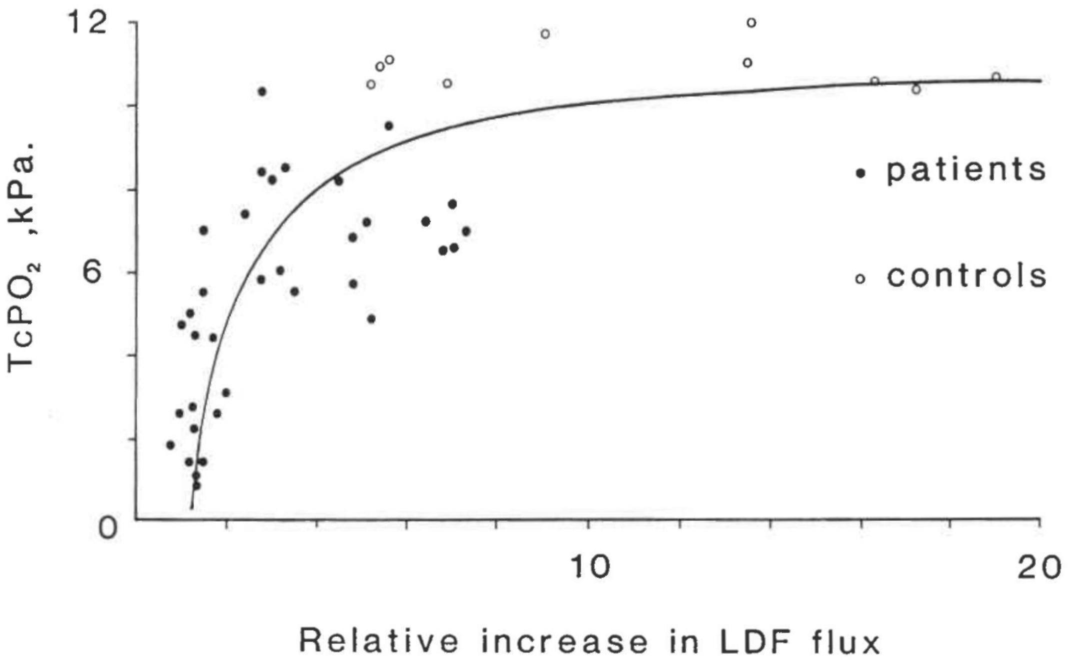

Fig. 1. Plot showing the results of relative increase in blood flow flux following heating as a function of skin $TcPO_2$ in kPa. The open circles represent normal subjects, the closed circles the amputee patients.

results of the present study which show that baseline LDF values which are in fact a measure of resting blood flow, do not permit discrimination between controls and patients. A similar observation has been made in patients and controls on the lateral aspect of the leg which would agree with the authors' findings (Pabst et al, 1985).

After local heating of the skin, the flux values recorded allowed the control group to be distinguished from patients with vascular disease and separation of BK/AK groups was possible. By considering the relative increase in flux (heated flux/baseline flux) it was also possible to separate BK amputations from AK, and both of these from the controls. The latter parameter is considered to be of most use in discriminating between the BK and AK groups because it is not influenced by the actual LDF value after heating but is a measure of microcirculatory reactivity.

In a previously reported uncontrolled pilot study using an LDF for amputation level selection, the increase in flux brought about by local heating was also measured (Holloway and Burgess, 1983). Although no exact values are given, successful amputation in both AK and BK patients was indicated when the flux achieved after heating was at least one third of that seen in controls. The findings in the present study differ substantially in that all AK amputations were found to heal primarily in situations where the heated flux was less than 30% of the control levels. However, exact comparisons are difficult because of the paucity of information in the previously published work.

A study of LDF in controls and patients with peripheral vascular disease (PVD) has shown that a significant difference in baseline level exists between controls and severely ischaemic limbs in the great toe only (Pabst et al, 1985). Differences were not found at more proximal sites between controls and either mild or severe PVD, a finding which the authors' observations confirm. The derivation of a reactive hyperaemic index in these groups showed a significant difference between the control and severe PVD groups at all sites on the limb. However, a distinction between mild and severe ischaemia could not be made at any site on the leg, an indication that the technique would probably be of little value in assessing amputation level. In any event, producing reactive hyperaemia in patients with severe ischaemia is questionable in terms of the pain it produces. In the authors' experience, pre-operative pain in the amputation patient is not always well controlled. These patients are often acute admissions in need of immediate amputation surgery, or their condition has deteriorated rapidly following other unsuccessful vascular prodecures. Therefore any technique which could painlessly obtain the same information would be preferable.

It might be argued that the scatter of the points in this limited series of observations suggests that there is only a poor correlation between skin oxygen levels and flux change. However, a one-to-one correlation would not be expected. Indeed, since healing is as likely (if not more so) to be influenced by tissue perfusion rather than skin oxygen the results suggest that laser Doppler flowmetry could be equally effective in determining amputation level. From these present results, local heating of the skin and evaluation of its response to this stimulus with an LDF would appear to be a more sensitive (and certainly painless) test for determining the degree of ischaemia in the lower limb and hence the most distal level for successful amputation.

The authors' experience over several years in measuring $TcPO_2$ levels in amputees suggests that the technique demands meticulous and time consuming attention to detail and system recalibration, even when moving the instrument between two patients on the same ward. More recent experience with the LDF has shown it, for the most part, to be far more reliable, simpler and quicker to use and intrinsically more stable. However, on a number of occasions some instability has been noted in the readings indicated. This instability appears to be a function of movement of the fibreoptic bundle and is related to the multimode nature of the fibres (Moore, 1985). The artefact can be overcome by stabilizing the fibre bundle, by taping it to the bed clothes for example.

Despite this reservation it is believed that the LDF in conjunction with thermal stressing of the skin could provide the best method available for determining the level for lower limb amputation. Its use now needs evaluation in a controlled prospective trial.

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Mobility after major limb amputation for arterial occlusive disease

W. FRANCIS and C. J. C. RENTON

County Hospital, Hereford

Abstract

This study is concerned with the degree of mobility achieved by patients following major amputation for arterial occlusive disease of the legs and its relationship to the level of amputation. Eighty-four out of a possible 85 consecutive amputees form the basis of the study and the degree of mobility was assessed and graded in survivors six months after amputation.

Of the 69 survivors 74% were mobile to some degree and 57% walked daily with a prosthesis. Sixty-five per cent of all the amputations were below-knee. Seventeen per cent of below-knee stumps in patients surviving two weeks failed to heal. In amputees who attained a unilateral mobile healed stump 78% with below-knee amputations and 50% with above-knee amputations walked daily with a prosthesis. To obtain maximum mobility the knee should be retained whenever practical even though this results in some unhealed stumps requiring revision.

Introduction

When revascularization of an ischaemic limb is not practical or has failed, major amputation is necessary to save life, relieve pain and remove necrotic tissue. An above-knee (A/K) amputation fulfills these aims (Haynes and Middleton, 1981), but many reports have stressed that retention of the knee joint increases the chances of mobility using a prosthesis, improving independence and quality of life (Robinson, 1980; De Cossart et al, 1983; Jamieson et al, 1983). Attempts to retain the knee joint can result in a significant proportion of unsatisfactory stumps due to non-healing or flexion contracture of the knee (Moffat et al, 1981; De Cossart et al, 1983). However, since 1969 the authors have increasingly retained the knee joint. From 1969 to 1973, 34% of all amputations were below-knee (B/K), from 1974 to 1978, 53% were B/K, and from 1979 to 1983, 65% were B/K. This study represents the experience, in the last five-year period, of mobility and stump healing in amputees followed up for six months post-operatively.

All correspondence should be addressed to Mr. C. J. C. Renton, County Hospital, Hereford HR1 2ER, United Kingdom.

Patients and methods

All amputations described in this paper were performed by one general surgical firm, with an interest in arterial surgery, situated in a rural hospital. Many of the amputations were done by a succession of rotating registrars under the supervision of one consultant.

Between 1979 and 1983, 85 consecutive patients had their first major amputation for arterial occlusive disease of the legs. Many had previously had aortography, direct arterial surgery, sympathectomy or local amputation of the toes or forefoot. Eighty-four out of the 85 amputees were followed up for six months and then the mobility of the survivors was graded by a physiotherapist according to the scale described in Table 1. Those amputees who walked daily with the prosthesis, as categorized C2, C3, C4, were judged to be walking usefully.

Selection of amputation site

Patients with severe cardiac, respiratory, neurological, skeletal or mental disorders which

Table 1. Grading of mobility in amputees six months after amputation.

A	Chairbound
B	Walking with aids but without a prosthesis
C	Walking with a prosthesis
C_1	Occasionally
C_2	Daily indoors
C_3	Daily indoors and outdoors
C_4	Indoors, outdoors and stairs

would preclude them from walking with a prosthesis had an A/K or supracondylar amputation. The supracondylar amputation was favoured by the authors as they believed it produced a long balancing stump.

Massive gangrene or skin ischaemia at the potential site of a B/K amputation necessitated an A/K amputation. With absent femoral and distal pulses an A/K amputation was also favoured unless the ischaemia was confined to rest pain with or without limited gangrene of toes. However with a palpable femoral pulse and absent distal pulses a B/K amputation was preferred unless some of the above contra-indications were present. The visible skin circulation in the elevated foot proved helpful in doubtful cases. Doppler ultrasound ankle blood pressures identified some patients, usually diabetics, who had predominantly small vessel disease with clinical femoro-popliteal blocks. The presence of a popliteal pulse was usually an indication to attempt a B/K amputation.

Operative Details

In A/K amputations equal anterior and posterior flaps were fashioned with suture of the muscles. The Gritti-Stokes technique was used in supracondylar amputation, recently with lateral skin flaps. In B/K amputations the tibia was usually divided 9cm distal to the knee joint, although the distance could vary between 8 and 14cm. The gastrocnemius muscle was retained in a long posterior skin flap. Drainage in early cases in the series was by corrugated plastic drains via the wound and latterly by redivac suction drains. The skin was closed with interrupted nylon sutures. A well padded plaster of Paris was applied to B/K stumps for five days. Antibiotics were given.

Post operative care

Physiotherapy started pre-operatively in a minority of patients and continued immediately post-operatively in all patients. Pneumatic post-amputation mobility aids were used for walking as soon as practical, usually between seven and 14 days.

Once healed, the amputee was usually referred to the Artificial Limb and Appliance Centre at Selly Oak, Birmingham. The A/K amputees had ischial weight-bearing sockets with prostheses of either metal exo-skeletal construction or modular carbon fibre construction, each having semi-automatic knee locks. The B/K amputees had modular patellar-tendon-bearing prostheses with plastic sockets of polypropylene or glass reinforced plastic construction. The delay between operation and the first prosthesis was 7-10 weeks in B/K prostheses and 6-20 weeks in A/K prostheses.

Physiotherapy continued on an outpatient basis until the amputee was able to use the prosthesis reliably on his own.

Results

Mortality

Table 2 shows the progressive mortality for the peri-operative period up to two weeks, from two weeks post-operatively to discharge from hospital, and from discharge to six months after amputation. The total hospital mortality up to discharge was 11%. There was no mortality amongst the four amputees who lost their second leg, or amongst the six B/K amputees requiring A/K revision.

Age and sex

The average age at amputation was 73 years for A/K amputees, 82 years for supracondylar amputees and 71 years for B/K amputees. Seventy-five per cent of the amputees were male.

Diabetes

The prevalence of diabetes in 84 amputees was 30%. In patients with iliac blocks the

Table 2. Mortality rates in major amputations for lower limb ischaemia.

Amputations	No.	Within 2 weeks of amputation	2 weeks to discharge from hospital	Discharge to 6 months after amputation	Total No.	Total %
A/K	25	2	1	3	6	24
Supracondylar	4				0	—
B/K	55	3	3	3	9	16
TOTAL	84	5	4	6	15	18

prevalence was 18%, in femoral popliteal blocks 25% and in tibial blocks 78%. In 30 non-diabetic patients with femoral popliteal blocks, 23% of B/K stumps failed to heal, while in 11 diabetic patients with femoral popliteal blocks all B/K stumps healed.

Stump healing

All amputation stumps in A/K or supra-condylar amputees, who survived two weeks, healed. Table 3 relates stump healing in B/K amputees, who survived two weeks, to the clinical level of the arterial block. Seventeen per cent of A/K stumps and 33% of B/K stumps, which eventually healed, showed a delay in healing due to minor ischaemic lesions or sepsis. Thirty per cent of patients with iliac blocks, 73% of those with femoral popliteal blocks and 89% of those with tibial blocks had B/K amputations.

Table 3. Stump healing in relation to level of amputation and arterial block in amputees surviving two weeks.

Amputation	Site of Arterial Block	Total Numbers	Healed	Necrosed
B/K	Iliac	3	2 (67%)	1 (33%)
	Femoro-popliteal	41	33 (80%)	8 (20%)
	Tibial	8	8 (100%)	
	TOTAL	52	43 (83%)	9 (17%)

Flexion contractures

Three out of 46 B/K amputees discharged from hospital had a significant flexion contracture of the knee joint. Identifiable causes included early pre-operative contraction, delayed healing, painful stumps and inability to cooperate.

Mobility

Table 4 shows grades of mobility in the 69 survivors of the 84 amputees followed up for six months. Of all the amputees who walked usefully (C2 + C3 + C4), 20% walked daily indoors (C2), 16% walked indoors and outdoors (C3) and 20% also climbed stairs (C4). With a unilateral healed mobile stump 78% of B/K amputees and 50% of A/K amputees walked usefully.

Table 5 shows the progress of amputees surviving six months. Out of six B/K amputees requiring A/K revision, two walked usefully (C2, C3). Neither of the two B/K amputees with unhealed stumps walked usefully. Out of three B/K amputees with flexion contracture of the knee, one walked usefully (C2). Of the three bilateral B/K amputees, one walked usefully (C2). In spite of 30% of the B/K amputees not having unilateral mobile B/K stumps, this group of 46 amputees showed marked improvement in overall mobility (B + C) and in useful mobility with a prosthesis (C2 + C3 + C4) compared with the 19 A/K amputees.

Nine patients became bilateral B/K amputees during 1979-1983, six having had their first leg amputated before this period. Fifty-six per cent of these walked usefully with bilateral pros-theses (C2, C3). None of the authors' bilateral A/K amputees has walked with two full-length pros-theses.

Hospital stay, 1979-1983

The average hospital stay to discharge or death of the 25 A/K amputees was 32 days, and

Table 4. Mobility in amputees six months after amputation.

Initial Amputation		A/K	Supra-condylar	B/K	Total
Survivors		19	4	46	69
Mobility	A	7	4	7	18
	B	2		4	6
	C_1	1		5	6
	C_2	2		12	14
	C_3	3		8	11
	C_4	4		10	14
B + C					
% of survivors		63		85	74
$C_2 + C_3 + C_4$					
% of survivors		47		65	57

Table 5. progress of amputees surviving six months.

Initial Amputation	A/K	Supra-condylar	B/K
Survivors	19	4	46
Healed mobile stumps	18	4	32
Revision to A/K			6
Unhealed*			2
Flexion contracture			3
Bilateral	1		3

* 1 amputee was judged unfit for futher surgery
1 amputee had A/K revision at 7 months

for the 55 B/K amputees 38 days. If deaths and amputees becoming bilateral within the initial hospital admission are excluded, the hospital stay for A/K amputees was 35 days and for B/K amputees 34 days. If, in addition, amputees requiring A/K revision are discounted the stay of B/K amputees dropped to 28 days.

Discussion

The loss of a leg is a major disaster for all patients, limiting mobility and independence. The possibility of revascularizing ischaemic legs by arterial surgery, interventional angioplasty or sympathectomy must always be considered.

The progressive mortality and involvement of the contra-lateral leg in this study confirms previous reports (Finch et al, 1980; Rush et al, 1981; Bodily and Burgess, 1983). It emphasizes the importance of a short hospital stay and early rehabilitation to fully exploit the limited life expectancy. The higher mortality rate in A/K amputations compared to B/K is known (Robinson, 1980; Rush et al, 1981). The high proportion of diabetic amputees with distal arterial blocks and the marked improvement in below stump healing in diabetics with clinical femoro-popliteal blocks over non-diabetics confirms the prevalence of small vessel disease in diabetics.

The percentage of B/K amputees in other reported series varies between 11% and 67%, reflecting the wide range of importance given to retaining the knee joint (Robinson, 1980; Finch et al, 1980; Haynes and Middleton, 1981). When this is given high priority the knee joint can be retained in two-thirds of cases, and this is similar to the authors' experience (Barnes et al, 1976; De Cossart et al, 1983).

In A/K amputees healing occurs in over 90% of stumps (Kihn et al, 1972; Haynes and Middleton, 1981). In B/K amputees reports of stump healing vary between 65% and 89% (Kihn et al, 1972; Barnes et al, 1976; Couch et al, 1977; Finch et al, 1980; De Cossart et al, 1983; O'Dwyer and Edwards, 1985). In series giving a high priority to retaining the knee, non-healing is reported in 15 to 20% of amputees (Barnes et al, 1976; De Cossart et al, 1983).

As previously described, following past experience, the authors' indications for offering B/K amputation in patients with iliac blocks have been severely restricted, so no conclusions should be drawn from the results in the small numbers in this series. The healing rate in B/K amputees with femoral popliteal blocks was 80%, and with tibial blocks 100%.

The main problem is to judge which patients with femoral popliteal blocks will heal a B/K stump. Although A/K revision in an unhealed B/K stump carries a low mortality and heals well, further surgery with added suffering, prolonged hospitalization and rehabilitation must remain unsatisfactory. There is an urgent requirement for a reliable technique for selecting the most distal level at which an amputation stump can be relied upon to heal. Peripheral blood pressure estimations using Doppler ultrasound, transcutaneous oxygen pressures, isotope scanning, skin perfusion pressures and fluor-escein angiography have been found useful by some writers (Couch et al, 1977; Barnes et al, 1976; Robinson, 1980; Creaney et al, 1981; Holstein, 1982; Tanzer and Horne, 1982; Ito et al, 1984). However they have not been generally accepted (Finch et al, 1980; De Cossart et al, 1983; O'Dwyer and Edwards, 1985).

In B/K amputation the incidence of flexion contracture of the knee joint can be limited by intensive physiotherapy and by predicting pre-operatively those patients who are especially at risk. Moffat et al (1981) suggested that useful mobility with a prosthesis was still possible if the contracture was under 15° and this has also been the authors' experience.

Mobility studies ideally need to indicate grades of mobility and include all amputees and not just special groups. Jamieson and Ruckley (1983) reported from a general surgical unit that 70% of survivors were able to walk with or without assistance. Barnes et al (1976) reported that 60% of 50 amputees with a B/K prosthesis walked independently. Finch et al (1980) reported that 56% of a mixed group of A/K and B/K amputees were walking, 46.5% with a prosthesis. Robinson (1980) reported from the Roehampton Limb Fitting Centre that 82% of B/K and 38% of A/K amputees walked. Reports from the centres with integrated facilities specializing in the management and rehabilitation of the amputee appear the most encouraging (Robinson, 1980; Finch et al, 1980; Malone et al, 1981). However, the published reports for mobility are difficult to compare due to the variation in groups, criteria and times of review (Kihn et al, 1972; Barnes et al, 1976;

Couch et al, 1977; Robinson, 1980; Finch et al, 1980; Moffat et al, 1981). The authors' results, with 74% of all amputees who survived 6 months achieving some degree of mobility (B + C) and 57% walking usefully with a prosthesis (C2 + C3 + C4) to varying standards, appear favourable. There is a need for more detailed studies to allow comparisons to be made and targets set.

These results provide encouragement to continue giving a high priority to retaining the knee joint, so allowing two-thirds of amputees the best chance of mobility and independence. The presence of diabetes in patients with femoral popliteal blocks is a factor in favour of B/K amputation. If patients become bilateral B/K amputees there is still over a 50% chance of walking with two prostheses.

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Technical note

An alternative design of extension prosthesis

T. J. C. HARTE

Rehabilitation Engineering Centre, Musgrave Park Hospital, Belfast.

Abstract

Some patients with a congenitally shortened lower limb can be fitted with a total contact socket of one piece construction, dispensing with the need for removable panels or split socket construction. This gives advantages in weight, strength and cosmesis. The technique is described and compared with those conventionally used.

Introduction

The provision of a prosthesis for a patient with a congenitally shortened lower limb frequently presents more problems than that for the more commonly encountered amputee. The presence of a reasonably normal foot, although capable of load bearing, may give special difficulties. Often the proximal joints are inherently unstable and will require the support of a total contact socket, yet the relatively large foot produces problems of access to such a socket.

Socket design criteria

a) If the foot can take weight on the heel pad and also on the heads of the metatarsals, then suitable areas in the socket should be provided to preferentially stress them. The unstable proximal joints will require the general support of a total contact socket to maintain alignment.
b) The foot will usually be held in an equinus position within the socket. A system of forces must be provided to hold the heel pad on its seat.
c) Provision must be made to allow the foot to gain entrance to the socket yet retain the support of total contact.
d) Since ankle movement will be lost within the socket an attempt must be made to provide this function prosthetically.
e) The appearance of the finished limb must be acceptable cosmetically especially when fitted to the young and socially active.
f) The limb must be durable without being excessively heavy.

Conventional extension prostheses

The patten ended orthosis

This device consists of a metal "rocker" attached to the shoe by two or more struts. The foot is held plantar grade by the shoe, a Knee Ankle Foot Orthosis may be used to provide support for the proximal joint (Fig. 1, left).

While function is acceptable, cosmesis and weight are not.

Internal shoe extension

A leather boot is used to secure a raised heel under the patient's foot. This boot then fits securely into his own shoe. The foot is held in equinus, bearing weight on the heel and metatarsal heads.

Access is achieved through a laced front section which, when secure, will maintain the foot in position supporting the proximal joints.

Fig. 1. Left, patten ended orthosis. Right, internal shoe extension.

All correspondence to be addressed to Mr. T. J. C. Harte, Rehabilitation Engineering Centre, Musgrave Park Hospital, Stockman's Lane, Belfast BT9 7JB, Nothern Ireland.

Cosmetically the device is reasonable, but since the material of construction will deform, it is lacking in function and strength (Fig. 1, right).

The panelled socket

This is a total contact device manufactured in glass fibre reinforced plastic. It contains the shortened limb up to the level of the knee with careful moulding around the tibial crest to control rotation.

The rigid material and close fitting construction does not allow easy access. To permit this, an aperture is cut in the side of the socket allowing the foot to pass the narrow region and then locate in the distal portion. The cut out panel is then secured in position by a strap, so producing total contact and positive suspension.

Normal load-bearing areas of the foot are used, with the foot held in the maximum possible degree of plantar-flexion. This facilitates good cosmesis since the profile of the foot can be fitted inside a shank of normal dimensions. Ankle movement is simulated by a prosthetic foot of S.A.C.H. design. Cosmesis is good and function is excellent (Fig. 2, left).

Strength and durability are compromised by removal of the panel as loss of material weakens the structure. Higher stresses are experienced at the edges of the aperture, which can lead eventually to failure. Extra material may be used to increase the strength, with a consequent weight penalty.

The split socket design

Functionally this device behaves exactly like the panelled limb. The load-bearing areas in the socket and the positioning of the foot are the same, as is the material of construction.

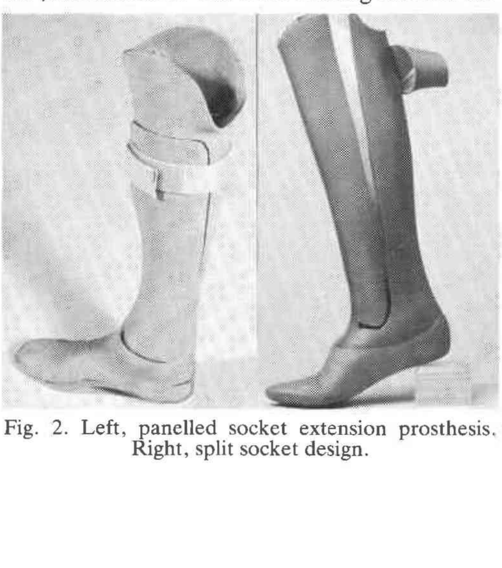

Fig. 2. Left, panelled socket extension prosthesis. Right, split socket design.

Access is gained by splitting the socket from the proximal brim down the medial and lateral aspect to a hinge at the distal end. When donning, the anterior shell is pulled away from the posterior shell and the foot inserted. The two halves are secured by a strap (Fig 2, right).

Strength is compromised by cutting the socket, and weight increase may be necessary to maintain strength.

An alternative design—the flexi-liner socket

Design

By continuing the use of glass reinforced plastic, but in the form of a conical one piece socket, the strength of the structure, can be greatly increased thus decreasing the weight.

Total contact fit is attained by using a soft flexible liner. The internal profile matches that of the limb remnant while the external profile matches the socket (Fig. 3).

When donning the device the patient first pulls the liner on to the limb and then pushes the liner and contained limb into the rigid socket. Entry to the soft liner is eased by two or more vertical cuts corresponding to the narrowest part of the limb. Once fitted these splits close regaining total contact.

Load-bearing areas in the socket remain as previously described, prosthetic foot construction is also unchanged.

Manufacturing technique

The liner is manufactured directly to a modified plaster cast of the affected limb and is moulded from P.E. Lite. Successive thicknesses are built up until the outer profile appears conical, the smallest circumference being distal.

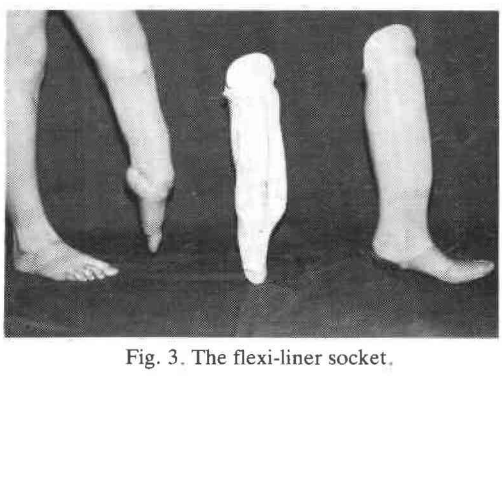

Fig. 3. The flexi-liner socket.

The cross-sectional profile should not be circular so as to provide positive resistance to rotation within the prosthesis or outer rigid socket. The outer rigid socket is manufactured in glass fibre over the outside of the finished liner using conventional techniques. The socket is then attached to the S.A.C.H. foot ready for dynamic alignment. The final exterior limb profile can now be shaped using filler material and completed by a second lamination integral with the foot.

Conclusions

Since changing to this method of construction the author has recorded a reduction in mechanical failures in extension prostheses. Weight has also been significantly reduced. Patient comments on cosmesis are encouraging. They also report an increase in overall comfort. The results to date would appear to indicate the usefulness of this technique.

FURTHER READING

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Prosthetics and Orthotics International, 1987, 11, 93–95

Technical note

An economic cushioned seat of variable easily adaptable configuration for cerebral palsied children

L. C. NAVA and V. H. PALLUZZI

Institute of Applied Mechanics, Puerto Belgrano Naval Base, Argentina

Abstract

This paper describes a low cost cushioned seat of variable configuration for cerebral palsied children in the 5 to 10 year age bracket.

Economic considerations are presented and the manufacturing process is described. It is believed that the system has advantages specially related to socio-economic conditions prevailing in a developing country.

Introduction

Several seating systems have been designed for cerebral palsied children (Ferguson-Pell and Paul, 1981). These include:

- moulded or vacuum-formed custom shaped shells
- shapeable matrices from interlocking injection-moulded components
- foam in place and foam in box
- parapodia and other standing support devices.

This paper describes the design philosophy and objectives for a special cushioned seat of variable configuration for cerebral palsied children in the 5 to 10 year age bracket. In view of the socio-economic conditions prevailing in a developing country like Argentina, a low cost design was a fundamental consideration in the development of the "IMA 509" Seat described in this publication*.

IMA Seat "509"—Design factors

Clinical

The two major clinical factors which must be considered in designing a seat for cerebral palsied children are: fixed skeletal deformities and pathological reflexes.

Cerebral palsied children are normally characterized by the predominance of spasticity, athetosis or hypotonicity.

Children with scoliosis, pelvic obliquity or other structural asymmetry must be provided with a seating system contoured to accommodate their irregular shapes.

A "custom seating" system has both seat and back specially shaped. A basic weakness, for instance in the moulded seating concept is the fact that it relies upon the hypothesis that a seat shape determined in the clinic on one day will be comfortable and clinically acceptable for an extended period thereafter (Ferguson-Pell and Paul, 1981). From this viewpoint a variable geometry seat which permits a wide range of parameters to be adjusted to find an optimum configuration for the disabled is highly desirable (Holte, 1983).

Technical and socio-economic

A basic design parameter in a developing country is low cost. This requires fabrication with materials readily available in the developing country. In Argentina high ductility aluminium plate, polyurethane foam and a non-toxic paint* fell into this category.

Other parameters taken into account were:

- usefulness to a large population of disabled children
- simplicity in construction in order that it may be fabricated by disabled workers or even members of the family of the children affected by cerebral palsy
- versatility to allow for (a) manual transportation (b) attachment to an automobile seat and (c) mounting on a standard wheelchair.

All correspondence to be addressed to Mr. L. C. Nava, Institute of Applied Mechanics, Puerto Belgrano Naval Base, 8111 Argentina.

* The development described resulted from collaboration between the "Instituto de Mecánica Aplicada" (IMA) and the School for Retarded Children No 509, Bahía Blanca.

** A roof covering paint was used with good results.

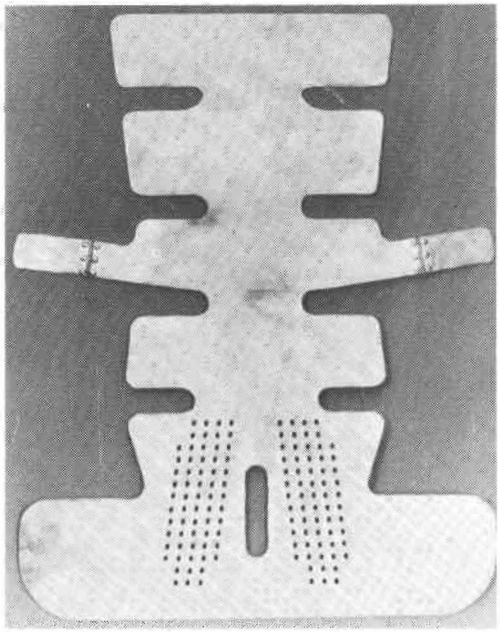

Fig. 1. Aluminium frame.

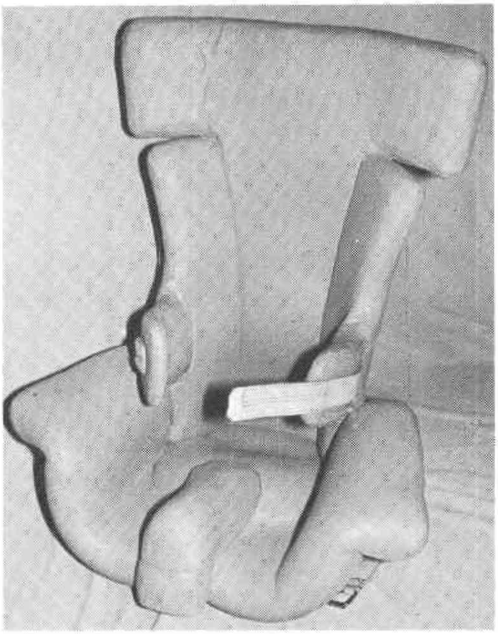

Fig. 2. Front view of the IMA 509 seat.

Manufacturing process

It is possible to obtain three aluminium "skeletons" from an aluminium plate of 1×2m (1.5mm thick) commercially available in Argentina.

Figure 1 shows the detail of a single aluminium frame. Holes are bored in the lower portion to permit ventilation of the foam.

The jointed arms are designed in such a manner that waist straps can be secured to them as shown in Figure 2. The waist straps mould around the patient's trunk for support when properly positioned.

The polyurethane foam is cemented to the aluminium plate with a commercially available adhesive.

Successive paint coatings are applied once the basic seat geometry has been adjusted to the patient (Figures 2 and 3).

Preliminary results and conclusions

Results obtained so far with the IMA 509 seat are satisfactory.

The cost of a single seat is approximately $100 (US) which is considerably lower than seats commercially available in highly industrialized countries. At the same time as can be appreciated in Figures 2 and 3 the IMA 509 variable configuration seat is aesthetically pleasing.

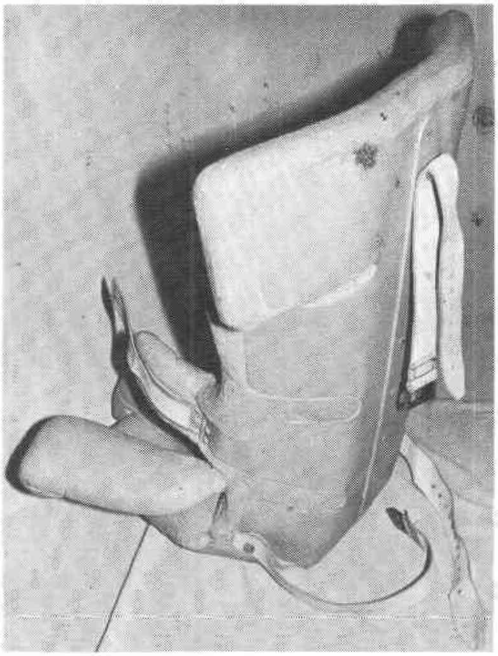

Fig. 3. Side view.

Acknowledgements

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The authors are grateful for the valuable cooperation and constructive criticism given by the IMA Director, Dr. P. A. A. Laura.

The authors are indebted to Maurice Leblanc and Richard Holte (Children's Hospital at Stanford, Palo Alto, California) and to M. W. Ferguson-Pell and J. P. Paul (Bioengineering Unit, University of Strathclyde, Glasgow) for the extremely valuable scientific information provided to the Institute of Applied Mechanics and their kind support. The authors express a deep feeling of gratitude to David Werner (The Hesperian Foundation, Palo Alto, California) for his interest, advice and moral support with respect to the present project.

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Prosthetics and Orthotics International, 1987, 11, 96–97

Technical note

Tangential goniometry-ANGULATOR

M. POKORA and L. OBER*

UL Gubinowska 7–115, Warsaw, Poland,

**UL Brzechwy 6, Poznan, Poland.*

Abstract

Tangential goniometry based on the principle of mechanical strain measurement is proposed for evaluating human body deformities or mobility. This features continuous body contour matching; defined measuring plane P2 due to rigidity in measured surface P1 (Fig. 1); stable physical characteristics; simple mechanical structure; ease of transformation of the measured angle into electric signal for CAD contour and shape reconstruction; can be integrated into orthopaedic appliances for monitoring activity.

The novel design of ANGULATOR — flexible polycentric goniometer is presented.

Principle

Two identical elastic steel flat strips (Fig. 2A) are joined together at the one end (B). The flat strips are kept in continuous contact by holders (C) in such a manner that the sliding of one element in respect to another is assured (exception: point B).

Bending such a structure (Fig. 2B) by the angle $\Delta\varphi$ causes a linear relative displacement $\Delta L_{\widehat{AB}}$ of both free ends of the strips. The amplitude of occurred displacement is proportional to the angle of bending $\Delta\varphi$ and the thickness of the single flat b as follows:

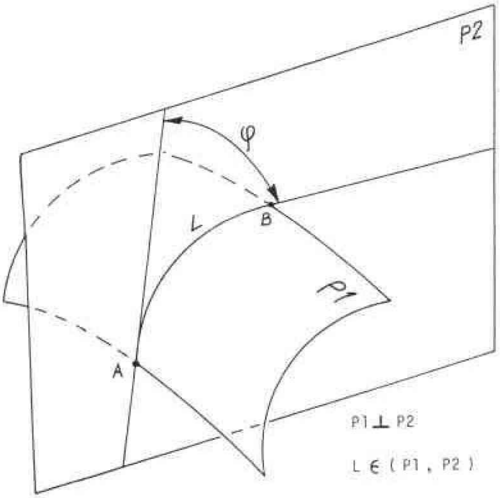

Fig. 1. Measurement plane P2 and plane of measured surface P1.

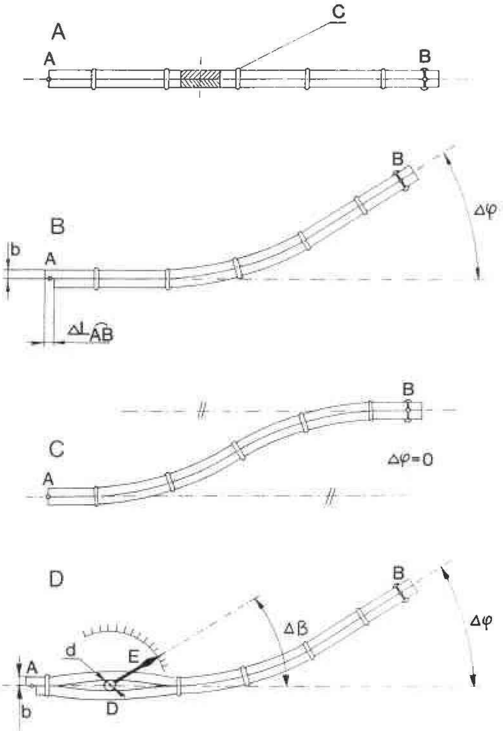

Fig. 2. A) Elastic steel flat strips joined at one end. B) Bending causes linear displacement of free ends. C) With A and B parallel the displacement is zero. D) Simple means of reading angle. Indicator E attached to shaft D.

All correspondence to be addressed to Mr. M. Pokora, UL Gubinowska 7–115, 02–956, Warsaw, Poland.

* The name ANGULATOR suggested by Dr. Per Udden.

$$\Delta L_{\widehat{AB}} = b \cdot \int_A^B d\varphi \quad (1)$$

or $$\Delta L_{\widehat{AB}} = b \cdot \Delta\varphi \ [\text{rad}] \quad (2)$$

When distal parts (A and B) are parallel (Fig. 2C) the displacement $\Delta L_{\widehat{AB}} = 0$.

For easy read out of the displacement (being angle equivalent) a linear displacement transducer can be applied. However the simplest mechanical solution is presented in Figure 2D. The shaft (D) with attached indicator (E) rolls between both ends of the strips at their free ends (point A). The angular displacement $\Delta\beta$ of the indicator is given by:

$$\Delta\beta = \tfrac{b}{d} \cdot \Delta\varphi \text{ where d – shaft diameter} \quad (3)$$

or $$\Delta\beta = \Delta\varphi \text{ if } d = b \quad (4)$$

Figure 3 displays the model of tangential goniometer ANGULATOR*.
Its measuring range is ± 195° with an error ± 1°.

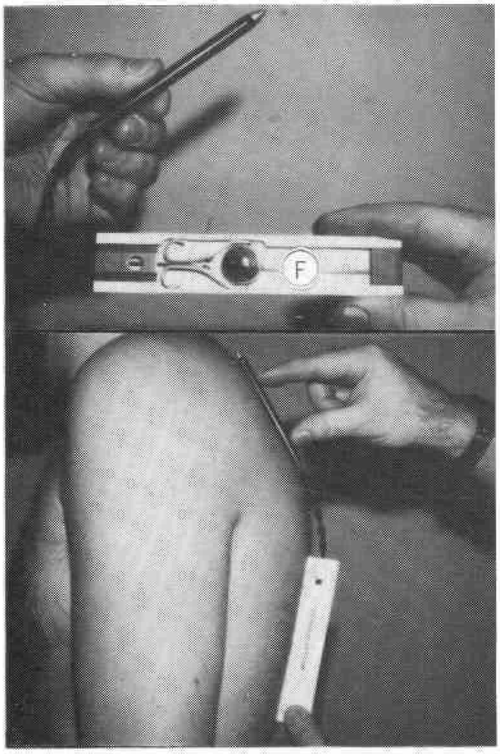

Fig. 3. Tangential goniometer ANGULATOR (see text).

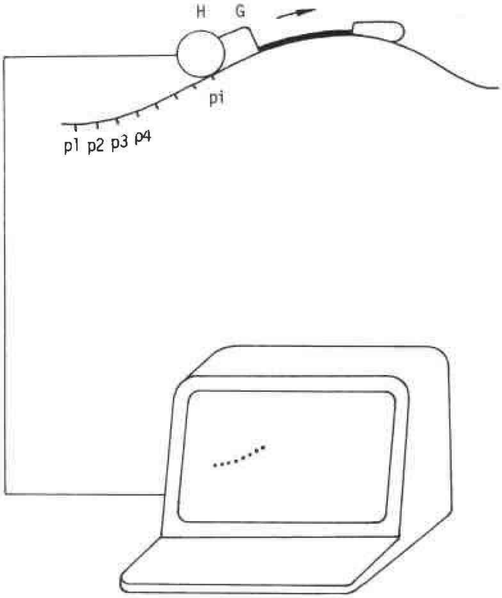

Fig. 4. Possible combination of the tangential goniometer and a microcomputer as an aid to deformity evaluation.

Within ± 90° the hysteresis is below 0.5°. To decrease the hysteresis within the range ± (90°–195°) the reverse spring (F) is applied acting above ± 90°. This device reduces the hysteresis to less than 1.5°. Having in mind the limited accuracy of positioning the goniometer arms in relation to the human body segments the ANGULATOR accuracy can be considered as adequate.

Future

Because of its flexibility and ease of application to follow the body contour, tangential goniometry will be helpful for deformity evaluation. A possible arrangement (Fig. 4) would combine the tangential goniometer with a photoelectric sensor (G); the coding wheel (H) supplies the system with a traced distance signal. Such a microcomputer based system can control the measurement and reconstruct the traced body contour to present it for evaluation.

This method could be an alternative to X-ray when used for assessment of scoliosis progression. It can also be a valuable aid for congenital deformity evaluation as well as in prosthetic and orthotic fitting.

Prosthetics and Orthotics International, 1987, 11, 98–99

Technical note

Splinting for CDH—initial impressions of a 'user-friendly' alternative

R. N. VILLAR, P. M. SCOTT and A. RONEN

Department of Orthopaedic Surgery, Addenbrooke's Hospital, Cambridge

Abstract
A lightweight, washable, and easily adjusted splint for the congenitally dislocated hip, designed to improve maternal compliance, is described. Observations are currently scientifically uncontrolled, though initial impressions are favourable.

Introduction

The treatment of congenital dislocation of the hip (CDH) in the newborn is an emotive subject, meeting with as many different ideas as there are surgeons treating it. Some elect to use splintage at an early stage (Fredensborg, 1976; Dunn et al, 1985; Visser, 1985) whilst others adopt a more relaxed approach, feeling that splintage may damage the capital epiphysis (Ilfeld and Makin, 1977) and that many unstable hips may stabilize without any treatment at all (Palmen, 1984). The choice of splints, if splintage is adopted, is diverse in the extreme (Heikkila and Ryoppy, 1984; Klisic et al, 1984; von Rosen, 1962; Sahlstrand et al, 1985; Ramsey et al, 1976; Wiersma 1976), the aim being to maintain the hip in sufficient abduction to prevent redislocation but not so widely abducted that avascular necrosis is the result. The length of time the appliance stays in place can vary from several weeks (Hadlow, 1979) to many months (Dunn, 1985; Ilfeld and Makin, 1977).

The authors have found, over several years of treating the unstable hip in their unit, that maternal compliance is a vital factor governing the success of a splint. A device that is cosmetically ugly, difficult to clean, and interferes with nappy-changing, ensures frequent visits by disgruntled parents to an overloaded clinic. This 'user-friendly' abduction splint devised with both mother and baby in mind is therefore presented.

Description of the device

The splint comprises a semi-circular metal waistband connected to two leg pieces by L-shaped metal bars (Fig. 1). All metal

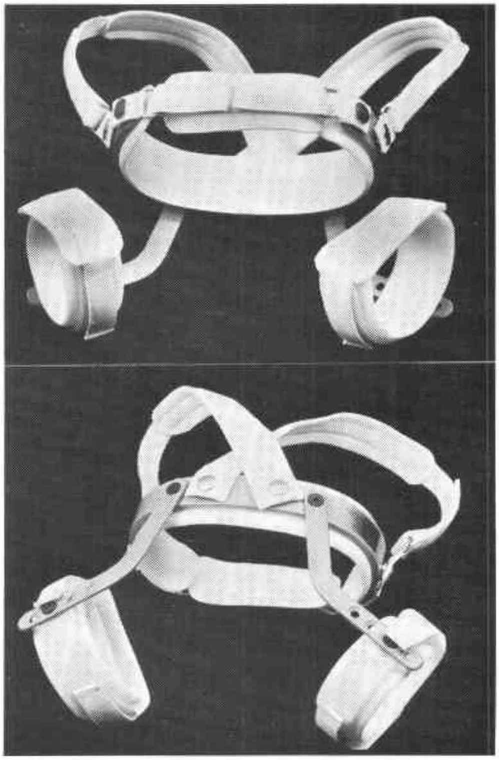

Fig. 1. Top, the abduction splint from the front. Bottom, the abduction splint from behind demonstrating the Allen nut adjustment mechanism.

All correspondence to be addressed to Mr. R. N. Villar, Senior Orthopaedic Registrar, Addenbrooke's Hospital, Hills Road, Cambridge, CB2 2QQ. United Kingdom.

attachments are lined by removable (and washable) liners. Straps are broad to prevent them from cutting into the baby. Each leg piece is connected to the back of the waistband by one Allen nut to allow adjustment to the precise splintage position required. Though the splint is provided in different sizes, all its components are detachable and it is thus possible to adapt any splint to fit a baby of any size. It has been found that babies with large waists do not necessarily have large thighs, and vice versa. Shoulder straps are also available if required. Mothers frequently request a second set of liners to act as a reserve and are now provided with these as routine so that the splint can avoid that much-used appearance that only babies can produce. These is no need to remove the splint if the liners need to be changed and nappy-changing is simple.

The device has now been used for a year with 20 patients completing treatment. All hips remain reduced and no instances of avascular necrosis have been recorded. However, this experience is currently uncontrolled and a longer-term controlled study is underway. At present however, the mothers involved are understandably enthusiastic that, given the constraints of splintage in the first place, the device is light, comfortable and hygienic, even permitting the occasional cuddle (Fig. 2). On such a basis it is felt that it can be strongly recommended to others as a 'user-friendly' alternative.

Fig. 2. The "occasional cuddle".

Acknowledgement

The authors are grateful to Cumbria Orthopaedic Ltd, London, for providing the necessary splints.

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Prosthetics and Orthotics International, 1987, 11, 100–103

Calendar of events

National Centre for Training and Education in Prosthetics and Orthotics
Short Term Courses and Seminars 1987/1988

Courses for Physicians, Surgeons and Therapists

NC503 Introductory Biomechanics; 19th–21st October, 1987.
NC508 Orthopaedic Footwear; 26th–27th November, 1987.
NC504 Lower Limb Orthotics; 30th November–4th December, 1987.
NC505 Lower Limb Prosthetics; 18th–22nd January, 1988.
NC502 Upper Limb Prosthetics and Orthotics; 25th–29th January, 1988.
NC510 Wheelchairs; 7th–8th March, 1988.
NC511 Clinical Gait Analysis; 14th–16th March, 1988.
NC506 Fracture Bracing; 11th–15th April, 1988. (Also suitable for orthotists and plaster technicians).
NC501 Functional Electrical Stimulation; 3rd–6th May, 1988.

Courses for Prosthetists

NC212 Hip Disarticulation Prosthetics; 29th October–6th November, 1987.
NC211 PTB Prosthetics; 7th–18th December, 1987.
NC205 Above-knee Prosthetics; 15th–26th February, 1988.

Course for Orthotists and Physiotherapists

NC217 Ankle–Foot–Orthoses for the Management of the Cerebral Palsy Child; 29th February–3rd March, 1988.

Seminar for Physicians, Surgeons, Therapists and Orthotists

NC717 Orthotic Management in Paraplegia; 2nd May, 1988.

Further information may be obtained by contacting Prof. J. Hughes, Director, National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, Curran Building, 131 St. James' Road, Glasgow G4 0LS, Scotland. Tel: 041-552 4400 ext. 3298.

24 September, 1987
Biological Engineering Society. Annual Scientific Meeting, Oxford.
Information: Miss J. Upton, B.E.S. Royal College of Surgeons, Lincoln's Inn Fields, London WC2 3PN.

1–3 October, 1987
Multidisciplinary International Meeting on Care of Babies Hips, Belgrade, Yugoslavia.
Information: Meeting on Care of Babies Hips, SAVA Centre, PO Box 1, 11070 Belgrade, Yugoslavia.

3–8 October, 1987
Western Orthopaedic Association, Colorado Springs, CO.
Information: 2975 Treat Blva,-E5, Concord 94518, U.S.A.

7–9 October, 1987
International Meeting of Rehabilitation Centres for the Physically Handicapped, Madrid, Spain.
Information: Secretariat, Fundacion Mapbre, International Meeting of Rehabilitation Centres for the Physically Handicapped, Apartado de Correos 36.237, Madrid, Spain.

7–9 October, 1987
3rd International Symposium on Biomedical Engineering, Madrid, Spain.
Information: 3rd International Symposium on Biomedical Engineering, Lab. de Bioengeniería, Clínica Puerta de Hierro, c/ San Martín de Porres 4, 28035 Madrid, Spain.

7–10 October, 1987
American Academy of Cerebral Palsy and Development Medicine Annual Meeting, Boston, U.S.A.
Information: AACPDM, PO Box 11083, Richmond, VA 23230, U.S.A.

9–11 October, 1987
3rd National Congress of the Chinese Biomedical Engineering Society.
Information: Professor Yang Zibin, Institute of Basic Medical Sciences, Chinese Academy of Medical Sciences, 5 Dong Dan San Tiao, Beijing, People's Republic of China.

20–22 October, 1987
3rd International Conference on Rural Rehabilitation Technologies, North Dakota, U.S.A.
Information: ICRRT III Headquarters, Box 8202, University Station, Grand Forks, ND 58202, U.S.A.

23–24 October, 1987
6th Southern Biomedical Engineering Conference, Dallas, U.S.A.
Information: Dr. Robert C. Eberhart, Conference Chairman, Dept. of Surgery/Biomedical Engineering Program, University of Texas Health Science Center at Dallas, 5323 Harry Hines Blvd., Dallas, Texas, U.S.A.

26–28 October, 1987
The Spine—Current Concepts and Techniques, Nashville, U.S.A.
Information: Dan Spengler, AAOS, 222 South Prospect, Park Ridge, IL 60068, U.S.A.

28–30 October, 1987
2nd Protech International Congress, Groningen, Netherlands.
Information: Dept. of Rehabilitation, University Hospital, Oostersingel 59, 9713 E Z Groningen, Netherlands.

4–8 November, 1987
17th Annual Meeting of the American Association for Hand Surgery, San Juan, Puerto Rico.
Information: Myra Josephson, Central Office Manager, American Association for Hand Surgery, 2934 Fish Hatchery Road, Suite 218, Madison, Wisconsin 53713, U.S.A.

13–14 November, 1987
Cincinnati Spine Course: Current Concepts in the Management of Degenerative and Traumatic Spine Disorders.
Information: Sheila Stuckey, Coordinator, Mayfield Neurological Institute, 506 Oak St., Cincinnati, OH 45219, U.S.A.

26 November, 1987
Cerebal Palsy in Children, Wakefield, England.
Information: The Secretary, National Demonstration Centre, Pinderfields General Hospital, Aberford Rd, Wakefield WF1 4DG, West Yorks., England.

2–5 December, 1987
Autumn Meeting of the French Society for Surgery of the Hand, Paris, France.
Information: Aldebert Kapandji, SOCFI, 14 Rue Mandar, 75002 Paris, France.

7–11 December, 1987
3rd Handitec Symposium and Hospital Expo, Paris, France.
Information: Handitec, 12 rue du Val d'Osne, 94410 Saint Maurice, France.

1988

4–9 February, 1988
American Academy of Orthopaedic Surgeons Annual Meeting, Atlanta, Georgia.
Information: AAOS, 222 South Prospect, Park Ridge, Il 60068, U.S.A.

1–4 March, 1988
Intermeditech. International Medical Technology Exhibition and Conference, Glasgow, Scotland.
Information: SEC Exhibition and Conference Centre, Glasgow G3 8YW, Scotland.

18–20 April, 1988
ISPO UK Scientific Meeting, Bath, England.
Information: Dr R. G. S. Platts, Institute of Orthopaedics, Brockley Hill, Stanmore, Middlesex HA7 4LP, England.

May, 1988
Quadrennial Congress of the International Federation of Physical Medicine and Rehabilitation, Toronto, Canada.
Information: Dr John M. Melvin, 1000 N 92 St., Milwaukee, Wisconsin 53226, U.S.A.

May, 1988
5th International Conference on Mobility and Transport for Elderly and Disabled Persons, Stockholm, Sweden.
Information: 5th International Conference Secretariat, c/o Swedish Board of Transport, Box 1339, S–171 26 Solna, Sweden.

5–8 May, 1988
Pediatric Orthopaedic Society of North America, Colorado Springs, Colorado.
Information: AAOS, 222 South Prospect, Park Ridge, IL 60068, U.S.A.

10–13 May, 1988
Orthopädie+Reha-Technik 1988 International (International Trade Fair and Congress for Orthopaedics and Rehabilitation Technology), Nürnberg, West Germany.
Information: NMA Nürnberger Messe-und Ausstellungsgesellschaft mbH, Objektleitung, Messesentrum, D–8500 Nürnberg 50, West Germany.

18–20 May 1988
2nd S. M. Dinsdale International Conference in Rehabilitation, Ottowa, Canada.
Information: Education Dept., Royal Ottowa Regional Rehabilitation Centre, 505 Smyth Ottowa, Ontario K1H 8M2, Canada.

June, 1988
Congress of the World Federation of Occupational Therapists, Australia.
Information: Ms. S. Degilio, Plaistow Hospital, Samson St., London E13, U.K.

June, 1988
3rd European Conference on Research in Rehabilitation, Rotterdam, The Netherlands.
Information: Office for Post Graduate Medical Education, Erasmus University Rotterdam, PO Box 1738, 3000 DR Rotterdam, The Netherlands.

20–22 June, 1988
American Orthopaedic Association Annual Meeting, Hot Springs, Virginia.
Information: AOA, 222 South Prospect, Park Ridge, IL 60068, U.S.A.

20–23 June, 1988
7th Congress of the International Society of Electrophysiological Kinesiology, Enschede, The Netherlands.
Information: ISEK–88, Congress Secretariat, PO Box 310, 7500 A H Enschede, The Netherlands.

26 June–1 July, 1988
Spinal Disorders, 1988, Gothenburg, Sweden.
Information: Spinal Disorders 1988, Prof. Alf L. Nachernson, Dept. of Orthopaedics, Sahlgren Hospital, S–413 45 Gothenburg, Sweden.

5–9 September, 1988
16th World Congress of Rehabilitation International, Tokyo, Japan.
Information: Secretary General, 16th World Congress of Rehabilitation International, Japanese Society for Rehabilitation of the Disabled, 3–13–15, Higashi, Ikebukuro, Toshima-ku, Tokyo 170, Japan.

20–23 September, 1988
Scoliosis Research Society, Baltimore, U.S.A.
Information: Vern Tolo, SRS, 222 South Prospect, Park Ridge, IL 60068, U.S.A.

22–23 September, 1988
2nd Biomechanics and Orthotic Management of the Foot Meeting, Newcastle, England.
Information: Dr. D. J. Pratt, Orthotics and Disability Research Centre, Derbyshire Royal Infirmary, London Rd., Derby DE1 2QY, England.

25–30 October, 1988
American Academy of Orthotists and Prosthetists Annual National Assembly, Washington, DC.
Information: American Academy of Orthotists and Prosthetists, 717 Pendleton St., Alexandria, VA 22314, U.S.A.

26–29 October, 1988
American Academy of Cerebal Palsy and Development Medicine Annual Meeting, Toronto, Canada.
Information: AACPDM, PO Box 11083, Richmond, VA 23230, U.S.A.

3 December, 1988
Shriners Hospital Pediatric Orthopaedic Seminar, Philadelphia, U.S.A.
Information: Randal Betz, Shriners Hospital, 8400 Roosevelt Blvd, Philadelphia, PA 19152, U.S.A.

1989
9–14 January, 1989
American Academy of Orthopaedic Surgeons Annual Meeting, Las Vegas, U.S.A.
Information: AAOS, 222 South Prospect, Park Ridge, IL 60068, U.S.A.

12–17 November, 1989
ISPO World Congress, Kobe, Japan.
Information: Secretariat, 6th ISPO World Congress, c/o International Conference Organisers Inc., 5A Calm Building, 4-7 Akasaka 8-chome, Minato-Ku, Tokyo 107, Japan.

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(Closing date for applications — 15th September, 1987)

DUNDEE '88
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For further information contact: Secretariat,
Dundee '88,
c/o Dundee Limb Fitting Centre,
133 Queen Street,
Broughty Ferry,
Dundee DD5 1AG,
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Organized by the Department of Orthopaedic and Traumatic Surgery, University of Dundee.

INTERNATIONAL SEMINAR ON PROSTHETICS AND ORTHOTICS
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TOPICS will include:

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Venue:	The Dan Accadia Hotel, Herzliya, Israel.	
Fees:	Before July 1, 1987	After July 1, 1987
Physicians	US $ 180.00	US $ 220.00
Para-medical participants	US $ 140.00	US $ 170.00
Accompanying persons	US $ 80.00	US $ 80.00

Further information from the Secretariat, ISPO 1987, PO Box 50006, Tel Aviv 61500, Israel. Telephone (03) 654571.

INTERNATIONAL CONGRESS

PROTECH II

AMPUTATION TECHNIQUE AND
PROSTHETIC MANAGEMENT OF THE LOWER LIMB

OCTOBER 28-30 1987

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partial and total foot • below-knee • knee disarticulation
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SIMULTANEOUS TRANSLATION INTO ENGLISH, GERMAN AND DUTCH

Organization:
Congress Bureau, University of Groningen, telex: RUGEB NL 77360

Information:
Department of Rehabilitation, University Hospital, Oostersingel 59, 9713 EZ Groningen, The Netherlands. Telex: AZGN NL 53942

ISPO Publications

IV World Congress, September 1983
Book of Abstracts.
Contains abstracts of Plenary and Concurrent session papers, Films/Videotapes and Poster presentations $7.50 (US)

V World Congress, July, 1986 Copenhagen
Programme and Abstracts $50 (US)

Standards for Lower Limb Prostheses
Report of a Conference, Philadelphia 1977
$10 (US)

The Deformed Foot and Orthopaedic Footwear
Report of the ISPO Workshop in Stockholm March 1977
Edited by Bo Klasson
Co-Editors A. Forchheimer, J. Hughes, G. Murdoch ISPO Members $15 (US)
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Directory of Films in Prosthetics and Orthotics
Compiled and Edited by J. E. Edelstein and R. G. Donovan
Published 1980 $5 (US)

Prosthetics and Orthotics International
August 1983
Special Issue — Through-knee Amputation and Prosthetics
Edited by J. Steen Jensen $18.00 (US)

Prosthetics and Orthotics in the Developing World with Respect to Training and Education and Clinical Services
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Edited by N. A. Jacobs, G. Murdoch
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Non-Members $10 (US)

Planning and Installation of Orthopaedic Workshops in Developing Countries
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Non-Members $10 (US)

Training and Education in Prosthetics and Orthotics for Developing Countries
Report of an ISPO Workshop Jönköping, Sweden
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Orders which must be accompanied by the appropriate remittance (cheque or international bank draft made payable to ISPO), should be sent to:

ISPO Publications,
National Centre for Training and Education in Prosthetics and Orthotics,
The Curran Building,
University of Strathclyde,
131 St. James' Road,
Glasgow G4 0LS,
Scotland.

INTERNATIONAL SOCIETY FOR PROSTHETICS AND ORTHOTICS

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The Executive Board has agreed that individuals from Developing Countries who wish to join the Society can do so at a reduced annual membership fee of DKK150 (that is one hundred and fifty Danish Crowns). This reduced fee also applies to existing members. Those individuals who wish to take advantage of this reduced fee should apply in writing to:

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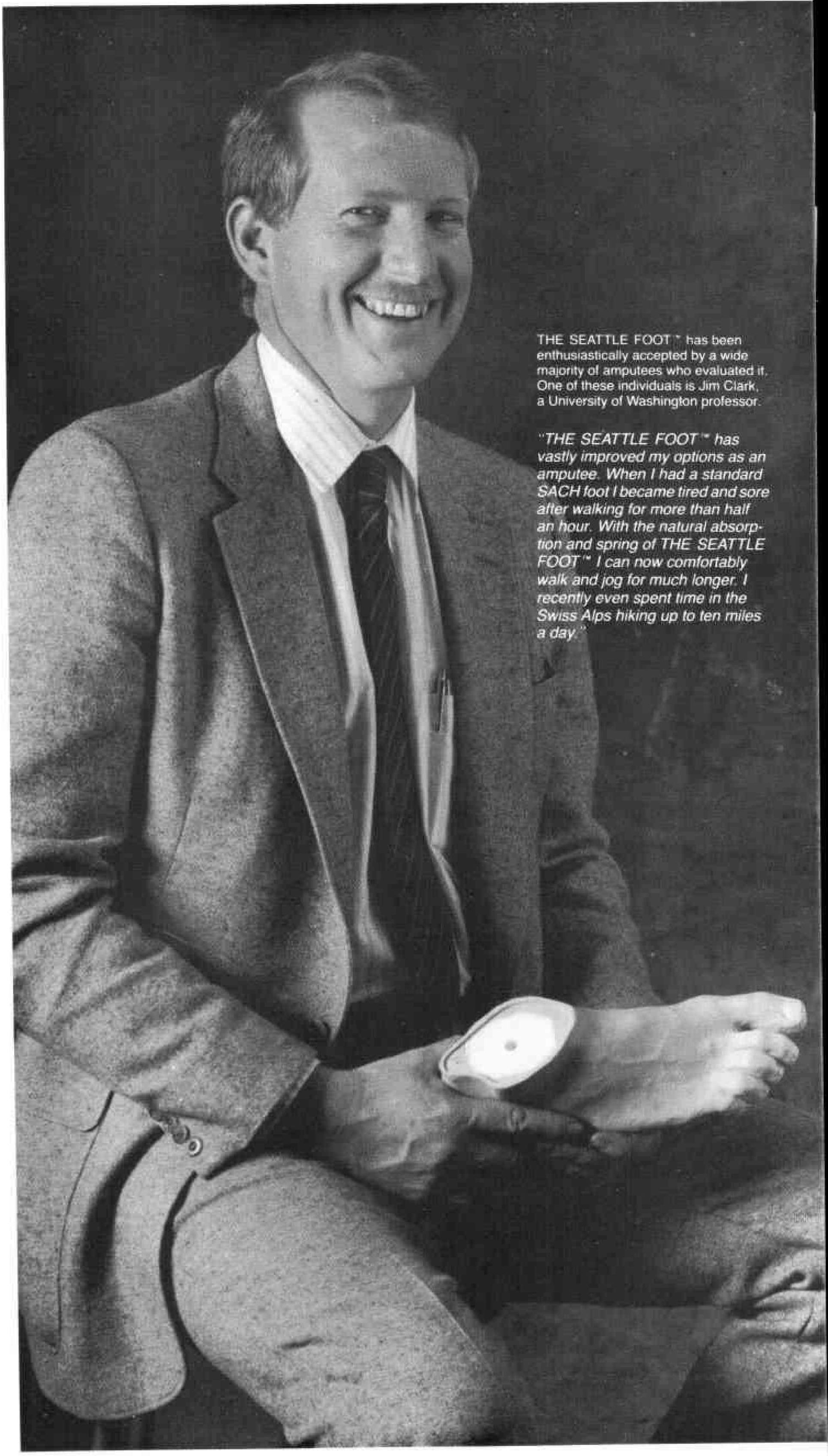

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