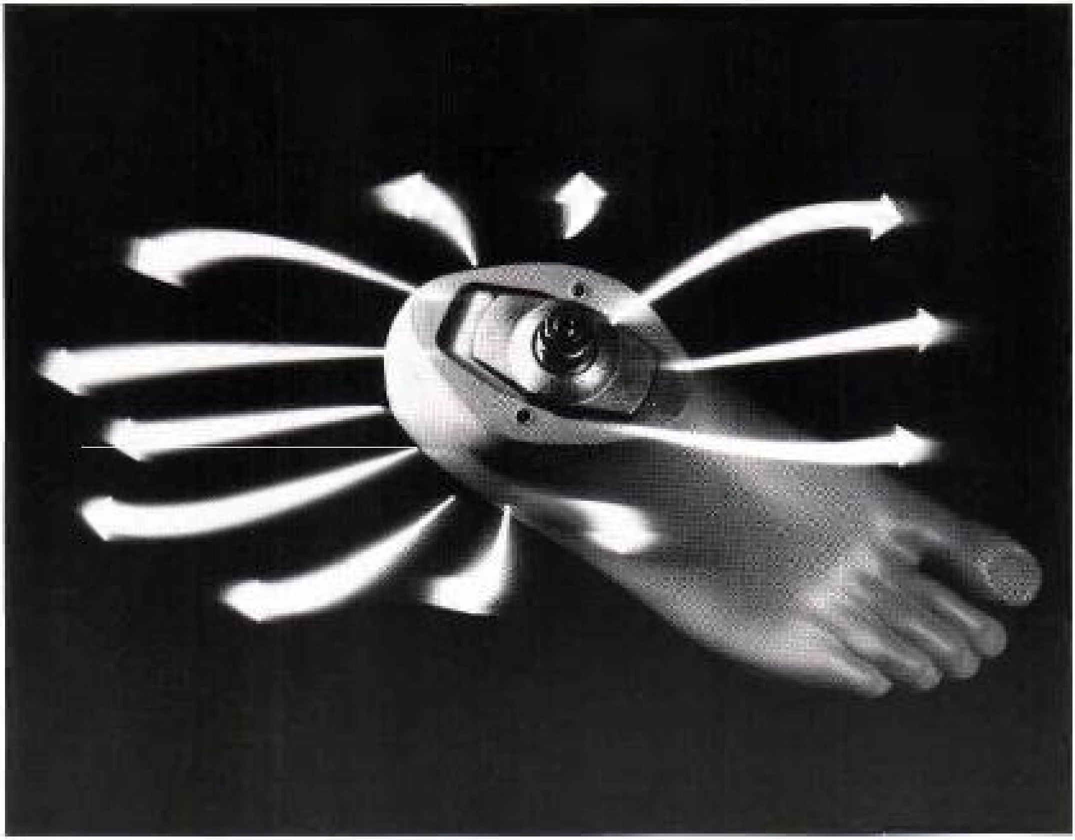
The Journal of the International Society
for Prosthetics and Orthotics

Prosthetics and Orthotics International

April 1993, Vol. 17, No. 1

1M1 – OTTO BOCK Multiaxial Foot

The foot for more comfort and elasticity with optimal cosmesis.



This new prosthetic foot combines multiaxial joint movements with the positive qualities of the OTTO BOCK Dynamic Foot.

- Dampened heel strike through plantar flexion
- Elastic forefoot as in the Dynamic Foot
- Pronation and supination possible in any position of the joint
- Low degree of transverse rotation due to joint bearing elasticity
- For patients who are moderately active

The Multiaxial Foot is supplied with a mounted modular adaptor.
Alignment recommendations:
- Mid foot positioned 10–15 mm anterior to the weight bearing line
- Heel heights of 10–15 mm

Otto Bock®

OTTO BOCK
ORTHOPÄDISCHE INDUSTRIE GmbH & Co
Industriestraße · P.O. Box 12 60
D-3408 DUDERSTADT
Telephone (0 55 27) 8 48-0
Telefax (0 55 27) 7 23 30

Prosthetics and Orthotics International

Prosthetics and Orthotics International is published three times yearly by the International Society for Prosthetics and Orthotics (ISPO), Borgervaenget 5,2100 Copenhagen Ø, Denmark, (Tel. (31) 20 72 60). Subscription rate is GBP65 per annum, single numbers GBP22. The journal is provided free to Members of ISPO. Remittances should be made payable to ISPO.

Editorial correspondence, advertisement bookings and enquiries should be directed to Prosthetics and Orthotics International, 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 4049).

ISSN 0309–3646

Produced by the National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, Glasgow

Printed by David J. Clark Limited, Glasgow

The Journal of the International Society for Prosthetics and Orthotics

April 1993, Vol. 17, No. 1

Contents

ISPO

Elected Members of Executive Board:

M. L. Stills (President)	USA
S. Sawamura (President Elect)	Japan
P. Christiansen (Vice President)	Denmark
J. Vaucher (Vice President)	Switzerland
H. Arendzen	Netherlands
D. N. Condie	UK
T. Keokarn	Thailand
H. Shangali	Tanzania
W. H. Eisma (Immediate Past President)	Netherlands
J. Steen Jensen (Hon. Treasurer)	Denmark
N. A. Jacobs (Hon. Secretary)	UK

Standing Committee Chairmen and Task Officers

B. Ebskov (Finance)	Denmark
W. H. Eisma (Protocol and Nominations)	Netherlands
S. Sawamura (Congress)	Japan
J. Hughes (Education)	UK
S. Heim (Education in Developing Countries)	Germany
W. Neumann (Certification)	USA
H. Arendzen (Membership, Publications)	Netherlands
D. N. Condie (Standards, Research)	UK
P. Christiansen (Professional Register)	Denmark

Consultants to the Executive Board

H. C. Chadderton (Consumer)	Canada
J. F. T. Bredie (IVO)	Netherlands
J. Van Rolleghem (Interbor)	Belgium
J. N. Wilson (WOC)	UK
T. Lagerwall (RI/ICTA)	Sweden

International Consultants to the Executive Board

C. Marincek	Central and Eastern Europe
J. Craig and R. de Saez	Central and South America
O. E. Feldman	Russia

Chairmen of National Member Societies

Australia	G. Carter
Austria	W. Ott
Belgium	E. Schoolmeester
Canada	G. Martel
Carribean	J. Martina
China	Zhongzhe Wu
Denmark	P. Christiansen
Finland	L. Nummelin
Germany	R. Baumgartner
Hong Kong	K. Y. Lee
India	M. K. Goel
Israel	M. Azaria
Japan	S. Sawamura
Korea	J. S. Shin
Netherlands	J. H. B. Geertzen
New Zealand	A. W. Beasley
Norway	O. Johansen
Pakistan	N. M. Akhtar
Sweden	A. Stenström
Switzerland	J. Vaucher
UK	R. A. Cooper
USA	M. Schuch

Past Presidents

K. Jansen (1974–1977)	Denmark
G. Murdoch (1977–1980)	UK
A. Staros (1980–1982)	USA
E. Lyquist (1982–1983)	Denmark
E. G. Marquardt (1983–1986)	Germany
J. Hughes (1986-1989)	UK
W. H. Eisma (1989–1992)	Netherlands

Secretary

Aase Larsson	Denmark

Editorial

The financial statement for 1992 shows a high profit, which was mainly achieved through the successful VII World Congress in Chicago competently conducted by Dudley Childress and the American group, whose hard work also assured the scientific success of the programme.

The sound economic position of the society was further strengthened by the income from the Amputation Up-date Course in Groningen, Netherlands. It allows the Society to carry the costs of setting up international courses in the Developing World in collaboration with other international bodies during the coming years.

Even without these contributions the accounts showed a profit for the third continuous year. During the past triennium the Society has put a high emphasis on involvement with other international organisations working in the field of amputation and prosthetics, severe disablement and orthotics; both in clinical practice as well as in areas of education.

The Society is grateful to the many volunteers whom, without reimbursement, have worked with the daily operations and kept the wheels running, in particular at the National Centre in Glasgow, the Rehabilitation Department in Groningen, and around the Copenhagen office. In spite of increasing activities our Secretary, Aase Larsson, in Copenhagen has managed to accomplish the extra work load with only minor need for external labour assistance.

The Society acknowledges the continuous support from the War Amputations of Canada, and in particular SAHVA in Copenhagen, which will in the future, as it has done in the past, provide free office facilities. SAHVA has also supported ISPO economically for many years. We know that the financial infusion from SAHVA has been a stimulus for the extension of our international activities. In our current economical situation we shall abstain from further requests to SAHVA, but are happy to learn that they will continue as a Sponsoring Member of ISPO.

Because of the considerable costs involved, the Society refrained from issuing the triennial membership list to all our members, but only printed a limited stock for which requests can be made to the Copenhagen office. We currently seek alternative solutions such as to provide up-dated membership lists to the National Member Societies on a yearly basis.

The journal, Prosthetics and Orthotics International, has again operated with a deficit because of the high number of printed pages. Further efforts will be made by our Honorary Secretary, who also undertakes the Production Editorship, to further increase the number of advertisements and find less expensive ways of distribution.

During the past triennium the membership fee has not been adjusted to inflation. We aim at keeping unchanged fees for the coming period, for the 28 high income countries according to the World Bank ranking amounting to 450 DKK, and for members from other geographical areas 225 DKK. We will still aim at running the daily work of the Society and its international engagements within the revenue from our membership. A conservative investment policy should also in the future ensure that some more costly activities like conferences and courses, could be carried through without eating into the capital.

J. Steen Jensen
Honorary Treasurer

Prosthetics and Orthotics International, 1993, 17, 2–5

ISPO Statement of Accounts, 1992

Auditor's Report

We have audited the enclosed Financial Statements for the year 1992.

The audit has been performed in accordance with approved auditing standards and has included such procedures as I have considered necessary.

The financial statements have been prepared in accordance with statutory requirements, and the constitutions of the Society and generally accepted accounting principles. In our opinion the financial statements give a true and fair view of the state of the associations affairs as of December 31, 1992 and of the result for the year then ended.

Copenhagen, March 1, 1993.
Revisorgruppen Danmark.

Søren Wonsild Glud
State Authorised Public Accountant

Accounting Policies
Securities
Bonds and shares have been valued at the lower of cost or market.

Office equipment
Computer and office equipment have been stated at cost less depreciation, computed straight line over 5 years.

Accrual concept
The accrual concept of accounting has been used in these financial statements.

Income Statement for the Year 1992

SUMMARY	1992	1991
Society membership fees (note 1)	1.151.504	991.546
Sponsorship (note 2)	55.870	137.696
Meetings with other organisations (note 3)	(195.948)	(209.681)
Conferences, courses etc (note 4)	1.635.875	31.106
World Congress 1989	–	–
Prosthetics and Orthotics International (note 5)	(72.504)	(266.641)
Professional register	–	(3.953)
Publications (note 6)	18.748	(12.979)
Activity result	2.593.545	667.094
Administration expenses (note 7)	(912.793)	(833.988)
Primary result	1.680.752	(166.894)
Interest (note 8)	410.133	316.321
Dividend (note 8)	1.504	1.504
Maturity yield (note 8)	7.388	5.135
Change in value of securities (note 8)	(6.392)	64.284
Financial income	412.633	387.244
Net income	DKK 2.093.385	220.350

Balance sheet as of December 31, 1992

ASSETS		1992	1991
Cash		2.994.181	964.215
Accrued interest		85.925	68.468
Advertising receivable		66.778	64.202
Prepayment, World Congress		94.433	142.616
Advance funding of World Congress 1980		87.437	87.437
Other		31.296	–
Receivables		365.869	362.723
Securities (note 9)		2.855.379	2.863.660
Office equipment (note 8)		–	–
Total assets	DKK	6.215.429	4.190.598

LIABILITIES AND CAPITAL			
Accrued expenses		93.239	69.898
Accrued printing cost		160.000	—
Prepaid membership fees		3.350	34.100
Prepaid advertising income		9.387	6.174
Prepaid subscription income		72.464	96.600
Prepaid conference fees		–	200.222
Short-term liabilities		338.440	406.994
Provision World Congress 1980		87.437	87.437
Equity capital January 1, 1992		3.696.167	3.475.817
Net result		2.093.385	220.350
Equity capital December 31, 1992		5.789.552	3.696.167
Liabilities and Capital	DKK	6.215.429	4.190.598

Contingent liabilities (note 10)

Notes to the Financial Statements

1. Society membership fees

Membership fees consist of payments from 2604 listed members, excluding 45 honorary members.

		1992	1991
2. Sponsorship			
Contributions from: the War Amputations of Canada		30.870	37.696
SAHVA		25.000	100.000
	DKK	55.870	137.696

3. Meetings with other organisations

American Orthotics and Prosthetics Association		–	62.259
OT Berlin		–	54.449
Asian conference Japan		–	28.040
American Academy of Orthotists and Prosthetists		53.578	26.750
Education Commission		23.853	11.159
World Orthopaedic Concern		10.229	9.493
Interbor Brussels		5.906	8.013
UN Vienna		–	6.674
Other		26.261	–
ACCOPRA		–	2.845
Paraplegia		14.670	–
WHO Geneva		9.211	–
RI Nairobi		52.240	–
	DKK	195.948	209.682

4. Conferences, courses etc

Amputation surgery		–	(31.106)
Groningen		114.642	
Chicago		1.521.233	–
	DKK	1.635.875	(31.106)

5. Prosthetics and Orthotics International

Advertising		246.559	189.069
Subscriptions		209.691	225.675
		456.250	414.744
Printing and mailing		(494.539)	(603.820)
Production editor		(29.086)	(52.640)
Meeting expenses		(5.129)	(24.925)
		(528.754)	(681.385)
Net result	DKK	(72.504)	(266.641)

6. Publications

Booksales		28.120	22.755
CAD CAM report		–	(19.614)
Amputation Surgery Consensus		(9.372)	(16.120)
Total cost	DKK	18.748	(12.979)

7. Administrative expenses

Executive board and office:			
Travel and hotel cost		304.512	253.705
Meeting expenses		6.480	20.213
International Committee		–	11.078
Directory printing		58.500	–
	DKK	369.492	284.996

Secretariat, Copenhagen:			
Staff salaries		272.783	307.696
Labour tax		12.360	12.968
Data service		5.522	6.386
Meeting expenses		13.436	16.181
Bank expenses		1.339	6.863
Postage		86.930	52.403
Telephone		4.713	5.437
Stationery		16.174	36.232
Office supplies		2.481	6.904
Auditing		38.000	38.526
Bookkeeping		22.848	19.022
Sundry		15.872	6.570
Knud Jansen medals		50.843	—
Society promotion		—	9.490
Depreciation		—	24.314
		543.301	548,992
Administrative expenses, total	DKK	912.793	833.988

8. Office equipment

Computer equipment, at cost		95.347	95.347
Office equipment, at cost		26.220	26.220
		121.567	121.567
Depreciation January 1		(121,567)	(97,254)
Depreciation December 31		0	(24,313)
		(121.567)	(121.567)
	DKK	0	0

9. Securities

		Nominal value	Original cost	Year end value	Interest/ dividend
Bonds					
9% Kred. Danmark 2007		3.036.000	2.832.820	2.832.819	278.483
Shares					
Den Danske Bank		9.400	30.891	22.560	1.504
Total	DKK	3.045.400	2.863.711	2.855.379	279.987

10. Contingent liabilities

The association is involved in a court trial in connection with the World Congress 1980. The association might be liable to additional cost in this connection. The outcome is at present uncertain.

Prosthetics and Orthotics International, 1990, 14, 9-21

Biomechanics and shape of the above-knee socket considered in light of the ischial containment concept

C. H. PRITHAM

Durr-Fillauer Medical Inc., Chattanooga, Tennessee

Abstract

In recent years considerable interest has been generated in the United States and abroad about new style above-knee prosthetic sockets, variously referred to as Narrow M-L, NASNA, CAT-CAM and SCAT-CAM. More than a little confusion has attended the process. Moreover, the impression has been created that they are not governed by the basic biomechanical rules identified by Radcliffe as affecting the quadrilateral socket. Attention has come to be focused on the role of ischial containment and the term Ischial Containment (IC) socket is enjoying widespread use. This paper reviews many of the critical features of such sockets with the goal of first demonstrating that many of these features are dictated by the requirements of ischial containment, and second that the principles set forth by Radcliffe are fully applicable. The paper concludes with a brief discussion of the alignment principles associated with Long's Line.

Introduction

In 1974 Ivan Long became involved in a project :o evaluate radiographically the femoral alignment of above-knee amputees (Long, 1975; 1985; Mayfield et al, 1977). This has come to have profound effects not only on Long's practice but also on the practice of many prosthetists. In the process considerable confusion has caused many of the issues involved to be obscured; and somehow or another, the perception that the new style sockets are different from quadrilateral style sockets and unaffected by the principles of above-knee prosthetics as explained by Radcliffe (1955; 1970; 1977) has crept into popular consciousness. Recently, however, some semblance of order has begun to emerge (Pritham, 1988; Schuch, 1988) and attention has come to be focused on the role of the ischium. It is the author's contention that most if not all of the major factors influencing the shape of the newer sockets can be explained in terms of the principle of ischial containment. Further, it is the author's belief that this principle is fully compatible with Radcliffe's biomechanical analysis of the function of the quadrilateral socket and that the varying socket configurations are not at odds but rather are separate but related entities in a continuum labeled "above-knee sockets."

The goal of this article is to explore and clarify the issues involved. A wide variety of claims for the new socket configuration have been made. While there is a certain body of anecdotal subjective evidence to support some of them, the author is not aware of large scale objective scientific studies to substantiate any of the claims. However, for purposes of advancing the argument many of these contentions are accepted as given.

All correspondence to Mr. C. H. Pritham, C.P.O., Technical Coordinator, Durr-Fillauer Medical, Inc., P.O. Box 5189, Chattanooga, Tennessee, 37406, USA.

The problem

To understand properly the problem it is perhaps best to turn to Long's own statement (1985).

"Most above-knee amputees walk with a wide base and lurch to the amputated side. Only 100 per cent concentration can change that pattern. We looked at 100 x-rays of above-knee amputees standing in their prostheses and found 92 out of 100 to have a difference in angle of the femur. In 91 of 92, the difference was towards abduction."

"Abduction was caused by the quadrilateral socket being entirely too large in the M-L dimension and too tight in the A-P. The ischium sits on top of the seat at best, and a couple of inches above it in most fittings. The x-rays show the lateral wall to be several inches away from the femur except at the most distal point. When the femur exerts force against the lateral wall in weightbearing, the quadrilateral socket moves laterally immediately, because the ischium has no effect on stopping this shift. With the more narrow socket and increased A-P, the ischium is inside the socket, preventing lateral shifting of the socket during weight-bearing."

Of course, socket fit was not the only factor considered by Long and his co-workers. Apparently the initial focus of their investigations was not socket shape but alignment of the prosthesis (Long, 1975). Alignment will be considered separately at the end of this paper.

Mayfield (Mayfield et al, 1977) described the findings in an initial group of 38 amputees (presumably a sub-group of the 92 mentioned above by Long). Seventy-nine per cent of them were in abduction or neutral, and 13 per cent were in less adduction than the sound side. Only 8 per cent were in adduction equal to or greater than the sound side. Twenty of the 38 were refitted with revised techniques and an improvement in femoral alignment and gait. Another group of 13 new patients were fitted utilizing the new techniques and similar results were achieved.

In short, in the majority of cases examined by Long the prosthesis was ineffective in maintaining the proper relationship between the femur and the socket.

The solution

The solution as stated above is to prevent the proximal socket from shifting laterally by using ischial containment (also called bony or skeletal lock). To understand this solution properly it is perhaps best to start with Radcliffe's principle of lateral stabilization (1955). This may be summarized as follows:

1. The weight of the amputee's body, acting through the centre of gravity, tends to cause the pelvis to dip towards the sound side during stance phase on the amputated side.
2. This converts the pelvis into a lever with the supporting point, lateral of the ischium, acting as the fulcrum (Fig. 1).
3. The tendency of the pelvis to dip is resisted by the gluteus medius exerting a counteracting moment to the pelvic lever.

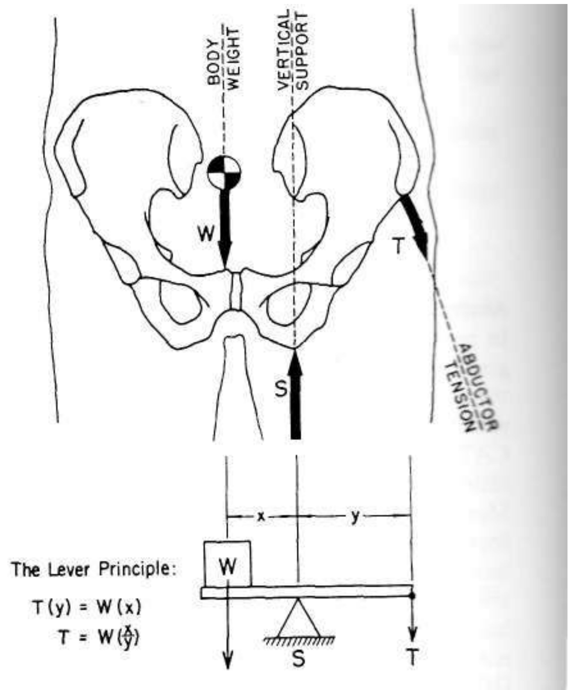


Fig.1. Moments acting about the support point in the frontal plane in lateral stabilization of the pelvis (Radcliffe, 1955).

4. For the gluteus medius to work at maximal physiological efficiency it must be maintained close to its normal rest length.
5. This is achieved when the femur is at its normal position of adduction.
6. The lateral wall of the socket must be shaped to maintain this position, anticipate the outward movement of the femur under load, and to distribute the pressure comfortably.
7. As a result of these forces acting against the shaft of the femur laterally, a counterpressure is created by the medial brim of the socket pressing against the stump so that "pressure in the crotch or medial area is then predominantly lateral rather than vertical" (Radcliffe, 1955). That is to say, a compressive force is exerted by the medial wall against the medial proximal tissues of the limb (Fig. 2).
8. This in turn creates a shearing force in the soft tissues trapped between the medial brim of the socket and the medial structures of the pelvis (Fig. 3).

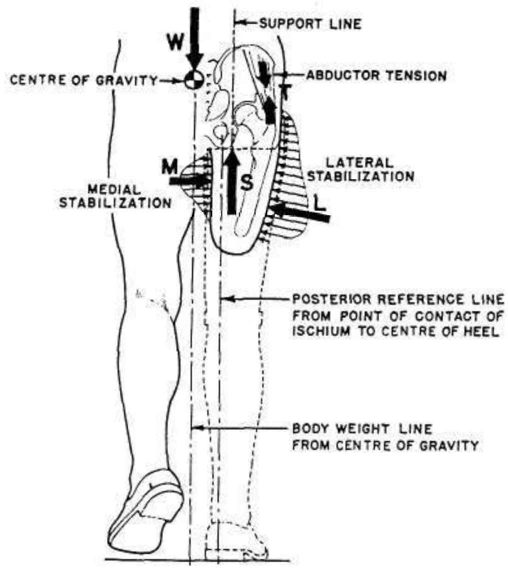


Fig. 2. Lateral stabilization of the pelvis (Radcliffe, 1970).

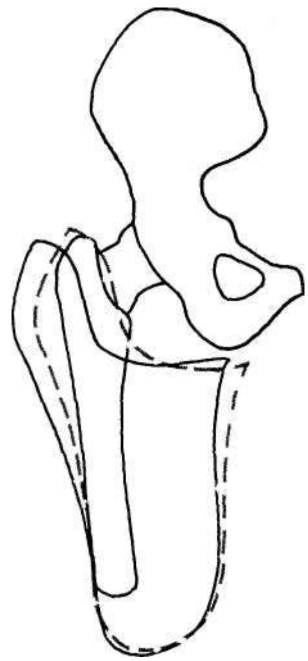


Fig. 3. Socket displaced laterally (solid) from its original resting position (dotted). This can result in a shearing force being exerted against the tissues between the medial bones of the pelvis and the medial brim of the socket.

The basic principle, as described by Radcliffe, is contained in numbers one through seven above. Point number eight is an addition to the basic principle added in response to comments like that of Long previously quoted.

Haberman (1963) performed a very similar analysis and attributed the shearing force to the medial displacement of the prosthetic support point (ischial tuberosity, about which the stump rotated on the prosthesis) relative to the physiological centre of rotation, the hip joint. To reduce this shearing to a minimum he advocated maintaining the support point as far laterally as possible in order to align it as closely as possible with the physiological centre of rotation. How this was to be accomplished is not apparent from Haberman's paper, although presumably it could be done by reducing the amount of ischial weightbearing and increasing the amount on the gluteus maximus.

Radcliffe, (1955) by way of contrast, was considerably more sanguine about the consequences of exerting laterally directed pressures in the perineum, although he did say "Flattening the medial wall of the socket is one means of ensuring a comfortable distribution of pressure in the adductor region" and "Providing efficient medial-lateral stabilization will also minimize medial shifting of the ischial tuberosity which might result in painful skin abrasion in this important weightbearing area" (1970).

Radcliffe also pointed out that the closer the support point is to the centre of gravity the less the moment tending to cause pelvic dip and the more efficient the countermoment of the gluteus medius on the femur. He apparently seemed to have considered any concomitant increase in shear forces as a small price to pay and well within the manageable limits.

The reasons for this sangiuneness are perhaps worth considering. Radcliffe's work was part of a larger effort initiated in response to the needs of World War II amputees, who at that time were for the most part young and healthy. It is to be presumed that much of his practical experience was gained with such amputees. Working with this group who had firmer tissues and stronger muscles than those that prevail with today's more typical patient, may well have masked problems that are more prevalent in today's practice. Another contributing factor that cannot be dismissed outright is Radcliffe's assertion that many of the problems described by Long and others may well be the results of poorly fitting sockets (Radcliffe, 1989), i.e. not made according to the principles outlined by the University of California team.

Leaving this last point aside, it may be presumed that the laterally directed shearing force in the perineal area and the inability of the soft tissues to withstand it causes discomfort and contributes to malalignment.

The solution that has emerged, and that was clearly apparent to Long, is to extend the medial brim upward so that pressure is brought to bear against the ramus. (Fig. 4). to quote Radcliffe (1989) "the medial counterpressure on the pubic (ischial) ramus clearly is capable of providing medial counterpressure which supplements the medial pressure on the adductor musculature. Since the socket slopes downward and inward along the entire medial brim this contour is faired into the medial wall of the socket which gives the impression of exaggeration of the medial counterpressure in the upper one-third of the socket."

This is the principle of ischial containment and many of the determining features of the newer designs derive from the desire to make ischial containment possible. It would seem logical to consider these features in a point-by-point fashion proceeding around the periphery of the socket.

Medial-lateral dimensions

The medial brim of the IC socket is an oblique sloping surface, upon which the ischium occupies a somewhat tenuous perch. To quote Radcliffe (1989).

"In taking advantage of the weightbearing potential on the medial aspect of the ramus the prosthetist is creating a situation much like weightbearing on the seat of a racing bicycle. To prevent the ramus from sliding laterally and downward into the socket the prosthetist must exaggerate the counterpressure from the lateral side. This has been done by a reduction in the M-L dimension particularly in the area just distal to the head of the trochanter."

Hence the emphasis on the M-L dimension of the IC socket. However, it has become clear at only a relatively late stage that the dimensions at more than one level are involved (Fig. 5).

Proximally the socket in the area at about the

Fig. 4. Medial forces borne by bones of the pelvis and soft tissues, the principle of ischial containment.

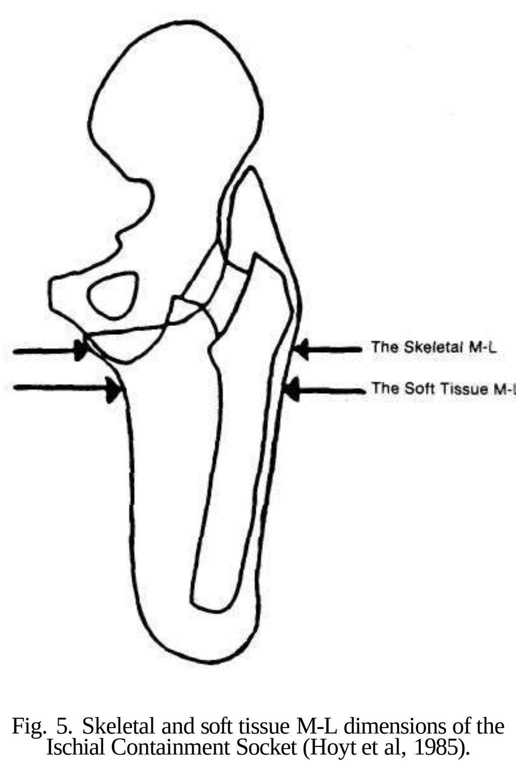


Fig. 5. Skeletal and soft tissue M-L dimensions of the Ischial Containment Socket (Hoyt et al, 1985).

level of the medial brim must be wide enough to accommodate the bones of the pelvis and the greater trochanter. Given that the ischial and pubic ramus pass obliquely (in the direction of internal rotation) from the ischial tuberosity to the pubic symphysis, it would seem logical that the M-L dimension in this area must be at least as large, if not larger, than the equivalent dimension of the quadrilateral socket.

At a level about 4cm distal to the ischial tuberosity, the M-L dimension is considerably reduced. As Radcliffe points out this is in order to bring pressure to bear on the femur on the area distal to the greater trochanter. It may also serve to load the tissues medially and thus play a role in creating the laterally directed counterforce necessary for lateral stabilization of the socket.

The dimension at the level of the ischial tuberosity is variously referred to in current texts as the Ischial Tuberosity (IT) M-L or the skeletal M-L (Hoyt et al, 1987; Prosthetic Consultants, 1987). The more distal diameter is called the Soft Tissue M-L or the Distal Ischial Tuberosity (DIT) M-L, and is either derived from the values given in Long's chart relating it to the circumference distal to the ischial tuberosity (Long, 1985) or is very closely related to these values, in most techniques.

Much of the confusion and the unfortunate sobriquet "Narrow M-L" would seem to have grown up over this latter dimension. A failure to appreciate the role of ischial containment and the need for different M-L dimensions at different levels coupled with a desire to emulate a poorly understood technique has led to more than one improperly fitting AK socket. Focusing on the lateral gapping in a quadrilateral socket and reducing the M-L dimensions in response would seem to be treating the symptom rather than the cause of the ailment.

Anterior-posterior dimension

For any particular fitting the volume of the patient's stump is a given (constant) regardless of the shape of the socket that the prosthetist wishes to fit. To quote Radcliffe (1989) again: "The soft tissues must be accommodated. Therefore, the A-P dimension is correspondingly increased as compared to the quadrilateral socket." Hence it can be seen that the major dimensions of the IC Socket are dictated by the imperatives of ischial containment. Other, secondary, rationales for a wide A-P dimension have been presented. It has been postulated that the greater A-P dimensions of the IC Socket better accommodate the major muscle groups of the thigh, permitting them to function more effectively (Long, 1985; Sabolich; 1985). Second it has been suggested but never proven that a concentration of pressure in the Scarpa's Triangle has a deleterious effect on circulation in the distal tissues (Sabolich, 1985).

With regard to the first point, Radcliffe (1977) clearly understood the necessity of allowing sufficient room for functioning muscle groups. "Regions of firm musculature such as along the rectus femoris muscle are channeled to avoid excessive pressure as required". "The socket contours are determined by reference to the information on stump muscle development recorded during the examination (Radcliffe, 1955). With these statements in mind there would seem to be no contradiction in principle between the quadrilateral socket and the IC Socket. Rather it would seem to boil down to a difference of opinion between advocates of both about which does the better job.

The second point is considerably more problematic. It seems self-evident that if any fundamental problem (such as adverse effects on circulation resulting from pressure in the Scarpa's Triangle) were to exist with the quadrilateral socket, there would have been considerable hue and cry and the design would have fallen into disfavour very early on. Yet the basic socket design has been in widespread international use for more than 25 years. Writing in 1964, Hall, stated: "Properly applied pressure is well tolerated by neurovascular structures. This is an interesting concept for orthopaedic surgeons, who have been painfully aware of the results of unrelieved plaster-of-Paris cast pressure over neurovascular tracts. Surprisingly, these vessels and nerves will tolerate firm pressure over extended periods of time if it is applied properly, while the same degree of pressure over a functioning muscle will prove to be intolerable. As considerable force must be applied over a sufficient area in the socket wall to stabilize the stump, and since those areas overlying contacting muscle bellies are not feasible, the ability to utilize zones superficial to neurovascular structures becomes most important." No convincing evidence has been advanced, even at this date, to challenge this assertion.

Contrary to the apparent opinions of some, Radcliffe never advocated application of all of the anterior counterpressure in the Scarpa's Triangle. What he did say was: "Distributed over the upper portion of the *entire anterior wall* (present author's emphasis) of the socket, such anterior counterpressure easily can prevent the ischium from sliding into the socket and can prevent the discomfort that would result in the crotch area." (Radcliffe, 1955). Clearly it was his intent that forces be distributed over the widest possible area, while taking due notice of the nature of the tissues involved and their response to pressure. "Since, by and large, the portion of the stump in contact with the region of the anterior brim is soft tissue, some compression of the stump is necessary."

Interestingly enough in recent months at least one of the most vocal advocates of Skeletal Contoured Adducter Trochanteric-Controlled Alignment Method (SCAT-CAM) fitting techniques, Sabolich, has begun using more contouring in the Scarpa's triangle than was formerly his practice. This is being done to improve anterior-posterior control and rotary stability. While this necessarily results in some reduction in the A-P diameter, the intent is most emphatically not to reduce the diameter to the same value that would be achieved in a quadrilateral socket. It is perhaps best thought of as channeling or contouring and not as a reduction in diameter. Sabolich remarks that quite often it is accompanied by an increase in the depth of the rectus channel laterally.

Regardless of amputation level or socket style, the underlying principles remain the same. Force should be distributed over the widest possible area with due recognition of the volumetric relationships to be effected, functioning muscle groups, and the response of tissues to the load. Confronted with conflicting claims from advocates of differing socket designs about which more effectively fulfills the same purpose, and in the absence of objective evidence to support one position or the other, it would seem necessary to give equal weight to both positions. Ultimately- the only necessary justification, and indeed the only compelling one, for a wide A-P in the IC Socket is the necessity of preserving the proper volume to accommodate the limb.

Medial brim

The desire to distribute at least a portion of the laterally directed thrust of the proximal socket to the ischium has major implications for the shape of the medial brim. The medial border of the ischium is to be loaded, while at the same time the adductor longus tendon and the pubic ramus, which are not pressure tolerant, are not to be loaded. Hence, the medial brim is high enough posteriorly to bear against the ischial ramus and dips lower as it passes anteriorly to clear the pubic ramus and adductor longus tendon (Fig. 6). Since it is desired to distribute pressure as evenly as possible, the brim parallels the course of the ischium as it goes from posterior to anterior and is therefore internally rotated when viewed in the transverse plane. These are the general criteria for shape of the medial brim. Specific details vary with fitting philosophy and with patient characteristics.

The height of the medial brim and the amount of ischium encompassed would seem to

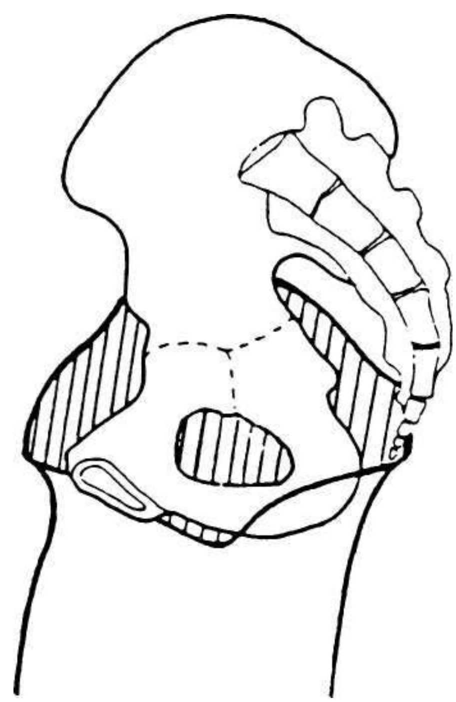


Fig. 6. Medial view of the Ischial Containment Socket in the sagittal plane showing relationship between the proximal edge of the medial wall and the bones of the pelvis (Hoyt et al, 1985).

be influenced primarily by the prosthetist's fitting philosophy. Above-knee sockets can be characterized by the amount of ischial containment from none (quadrilateral) to "maximal" (Pritham, 1988). Advocates of SCAT-CAM style sockets, at the maximum end of the scale, believe that it is both possible and desirable to bring the medial brim as far proximal as possible. Those individuals who believe in the broader group of moderate ischial containment socket designs are less emphatic about the need for height.

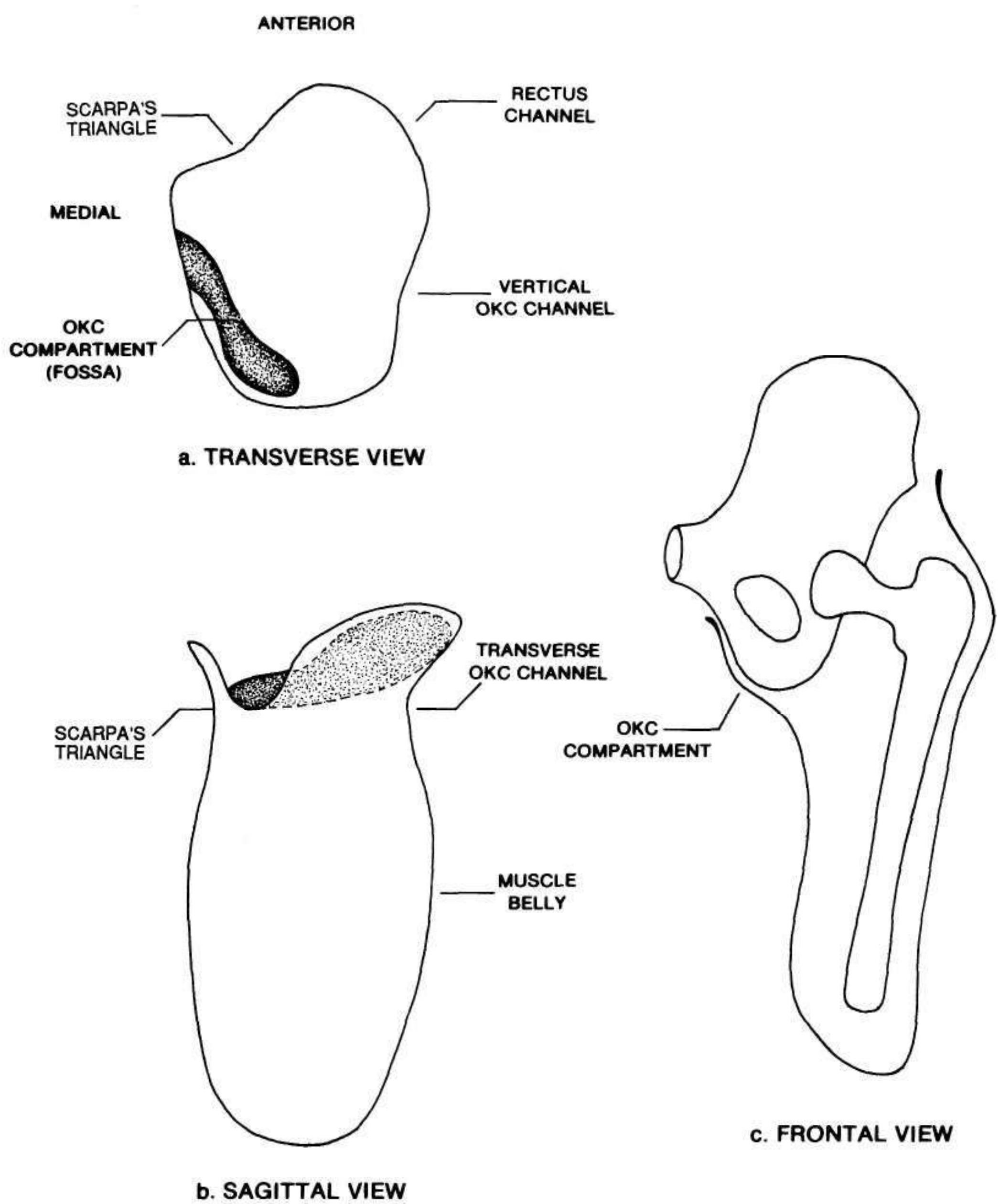


Fig. 7. The SCAT-CAM socket as described by Sabolich. Redrawn from an original by him to clarify some points. According to him some contours are exaggerated for emphasis, but not by much.

Those prosthetists that advocate higher brims are generally of the opinion that the increased height is made possible by flexible thermoplastic sockets. The general scheme is that the more flexible the brim, the more comfortable the patient, and the higher the brim can be. However, the question must be asked, what really is the role of the brim under these circumstances? Are the higher brims with their increased flexibility actually firm enough to be distributing an appreciable compressive load to the patient's tissues? Or, is the brim so flexible that it is acting only as a "shear distributor" to reduce the shear forces that are built up around the edge of the device in the transitional zone between the rigid socket wall and the relatively soft flesh of the patient? This latter concept is one that was developed by Murphy (1971) and Bennett (1971) in a series of theoretical articles published a decade ago. The practical implications of these articles and their potential impact upon prosthetic/orthotic design have never been fully appreciated.

In many instances the medial brim is not a flat oblique surface but rather is corrugated or channeled in cross-section as it goes from posterior to anterior. This is done to increase the amount of the ischium bearing against the brim and thus decrease the unit pressure. The amount of channeling that is needed would seem to be determined primarily by tissue properties. The softer the tissue, the more the load borne by the ischium and the more prone is the patient to discomfort. In an attempt to relieve this discomfort, the point of contact between the brim and ischium is relieved. When done correctly this results in a channel. The softer the tissue the more the brim is convoluted in cross-section and corrugated. This forms a concave inner surface. The firmer the tissue, the more the load that is borne by the soft tissue, the less that is borne by the ischium, and the flatter the brim can be in cross-section. The extreme of this case would be the patient who can bear all of the laterally directed load on the soft tissues without any reliance on the ischium. It would seem logical to consider a quadrilateral socket for such a patient. Nevertheless, it could be argued that comfort for such a patient, particularly one engaged in stressful athletic activities, could be enhanced by including the ischium in the socket.

Sabolich (1985) has described the channeling in the medial brim as an OKC (Oklahoma City) fossa. Most recently the fossa has been deepened and accentuated in the shape to become the OKC Compartment (Fig. 7.). "This Compartment ideally contains all the tuberosity and most of all the ramus except for the exiting symphysis pubis. As in the original article (Sabolich, 1985), the ramus is in a better location to include both in a compartment which makes the best possible use of medial superior containment both vertically and horizontally. This compartment is specifically contoured for these bones. This is the tough part."

Anterior brim

The impression has been created that the anterior brim of an IC Socket is lower than the anterior brim of a quadrilateral socket. In reviewing the literature, however, it is difficult to see how this impression has come about. The height of the anterior brim was not addressed in Long's (1985) article but was described in the Chicago Workshop (Pritham, 1988) as following the inguinal crease. Shamp recommends that the anterior brim be at the same level as the posterior brim (Prosthetic Consultants, 1987). The UCLA-CAT-CAM manual prescribes a brim just proximal to the inguinal crease (Hoyt et al, 1987). The consensus of the Chicago Worship was that generally it should follow the inguinal crease.

Radcliffe (1955) stated: "If fitted properly, the anterior brim usually can be brought up to the level of the inguinal crease without producing discomfort when the wearer is seated. The actual height of the anterior brim varies with the individual and is limited by contact with bony prominences."

It can be seen then that in height at least there is no real difference between the anterior wall of an IC Socket and a quadrilateral socket.

Lateral wall

Most descriptions of IC Style sockets describe them as extending quite high above the greater trochanter and with a great deal of contouring around that bony prominence (Hoyt et al, 1987; Long, 1985; Pritham, 1988; Prosthetic Consultants; 1987). This can perhaps best be explained as an offshoot of the demands of ischial containment. As has been previously discussed, one of the primary functions of the

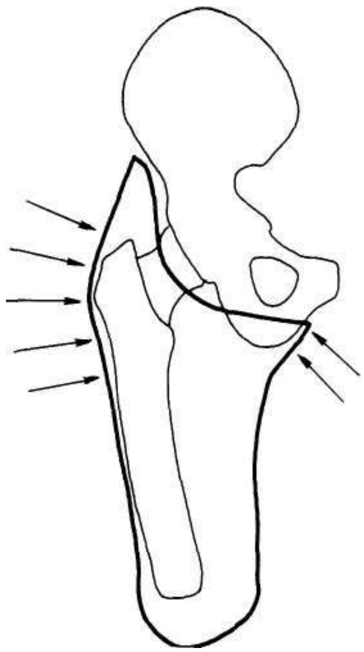


Fig. 8. Counterpressure generated by the lateral wall.

lateral wall is to generate the counterpressure necessary to maintain the ischium on the sloping medial brim (Fig. 8). The height and contouring of the lateral wall about the greater trochanter can be seen as necessary to distribute the load over a wide area and in an equitable fashion so that all the force is not concentrated on the most prominent lateral projection of the greater trochanter.

The other prominent feature of the various IC style sockets, when viewed in the transverse plane, is the extreme obliquity of the area posterior to the greater trochanter (termed the "wallet hollow" area by some) when compared to the comparable area of the quadrilateral socket. This is partly due to the demands of the counterpressure mechanisms and different fitting philosophy just discussed above. It can also be the result of trying to accommodate patients who are not as muscular and firm in this region as some. In many quadrilateral fittings it is necessary to create the same sort of contour just to preserve total contact. Radcliffe in his oral comments at the Miami meeting mentioned the necessity of this when working with older less physically fit patients than the young veterans he had experience with. This portion of his comments does not appear in any of the written accounts of his remarks (Radcliffe, 1989a; Radcliffe, 1989b; Schuch, 1988).

Whatever the socket style, firm pressure and contouring in this region posterior to the greater trochanter does more than generate the previously cited counterpressure. By compressing the gluteal muscle it helps create gluteal weightbearing, and by locking in around the greater trochanter it plays a role in providing rotary stability in the transverse plane. This contour is extended distally into the depths of the socket and, as will be seen, fulfills other roles at these levels.

Posterior brim

The posterior brim of the IC designs is described as being located 4 cm or so proximal to the ischial tuberosity so that the ischium is inside the socket. (It is doubtless this greater height of the posterior brim, as compared to the quadrilateral socket, that creates the impression of a low anterior brim). While it has been claimed that fitting the ischial tuberosity inside enhances a number of biomechanical functions (Prosthetic Consultants, 1987; Sabolich, 1985) the simplest explanation for the posterior brim's greater height is that it is a side effect of ischial containment and the increased height of the medial brim.

Function in the sagittal plane during gait

Radcliffe identified two separate force patterns (Fig. 9) that were exerted on the socket by the

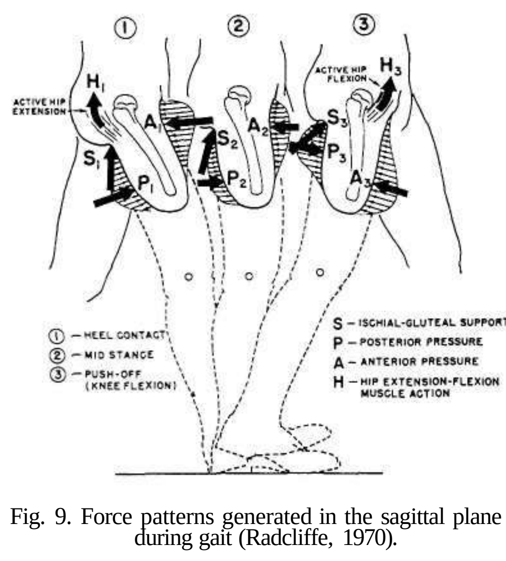


Fig. 9. Force patterns generated in the sagittal plane during gait (Radcliffe, 1970).

stump and which were the results of using the hip musculature to stabilize the knee in the early part of the stance phase and to initiate knee flexion in the later part. The first instance, knee stabilization, creates a situation where force is concentrated on the anterior proximal brim and the distal posterior portion of the socket. It is considered essential by Radcliffe (1970) to fit the anterior brim as high as possible into the inguinal crease so as to use the maximum effective stump length in this situation. With regard to the demands placed on the distal portion of the socket, he said "At the same time, the fitting must anticipate the movement of the femur stump within the soft tissue as the femur first presses posteriorly to maintain knee stability then moves anteriorly to initiate knee flexion in the swing phase". Such socket modification as the previously mentioned flattening of the area posterior to the shaft of the femur and the OKC Channel (Sabolich, 1985) can be seen as attempts to provide for effective transmission of force from the femur to the prosthesis postero-distally in order to stabilize the knee.

The force pattern is essentially reversed later in stance phase during the initiation of knee flexion. It should be borne in mind, however, that the forces required to initiate knee flexion are considerably less than those required to stabilize the knee in early stance phase. For this reason it will be appreciated that the functional demands placed on socket design are less. Undoubtedly this is what Shamp had in mind, when he said of the OKC Channel "Our experience is that the anterior channel is not necessary and may only serve to diminish the volume of the socket." (Prosthetic Consultants, 1987). Sabolich apparently has come to much the same conclusion for in a telephone conversation with the author in September 1988 he stated that it was currently his practice to remove considerable material from the area posterior of the femur and essentially none from the anterior region.

Proximally, much the same situation prevails. It may be argued that enclosing the posterior portion of the ischial tuberosity inside the socket enhances function in the saggittal plane. However, when the functional demands involved, i.e. those related to initiation of knee flexion in late stance phase, are considered, it can be appreciated that it really is not necessary. So, the prime criterion for extending the posterior brim of the socket proximal of the ischial tuberosity remains that of ischial containment. It is interesting to note, that while Radcliffe did not dwell on the work of Schnur, as did Lehneis (1985); he was aware of it, mentioned it in passing, and applied the principles in socket design. In 1955 he said that "conditions which create a great deal of discomfort can be prevented by shaping the bearing surface in such a way that the seat slopes toward the inside of the socket to render it more comfortable. Sloping increases the radius of the edge of the ischial seat and lessens the burning sensation of the skin in this region" (Radcliffe, 1955).

In a somewhat related matter Sabolich describes an indented horizontal channel immediatley distal to the ischial tuberosity. This channel, which he terms the Transverse OKC Channel, touches the ischial tuberosity tangentially and presses against the hamstring tendons. Distal to the channel the socket wall flares outward to accommodate the muscle bellies of the hamstring group. This channel continues the contours of the medial wall posterior and laterally to where it blends in the contours around the femur. Sabolich contends that the transverse OKC Channel enhances A-P Control, while the hamstring relief distally improves the function of those muscles.

Weightbearing

Of weightbearing in the quadrilateral socket Radcliffe (1970) has stated: "In the ischial-gluteal-bearing type of above-knee socket it is assumed that the contact against the ischial tuberositiy is the major source of vertical support. In addition, perhaps one third (33 per cent) of the vertical support is provided by firm contact pressure acting upward on the gluteus maximus. Other areas of the socket, such as the anterior brim also contribute to the vertical support in varying amounts, depending upon the individual fitting".

If "major" is interpreted to mean more than 50 per cent it can be concluded that something in the nature of 83 per cent (50 per cent ischial weightbearing plus 33 per cent gluteal) or more of the patient's weight is borne by ischial-gluteal weightbearing with the remaining 17 per cent or less borne by the anterior brim and other areas. The question is, how does this differ in the IC Socket?

As has been stated by Sabolich (1985) one of the goals of CAT-CAM fittings has been to increase the amount of weight borne by the femur, and that is at least one of the justifications he cites for increasing the adduction angle. This is in contrast to the more commonly stated goal of striving to fit the amputee in a postition of normal adduction, inclined eight degrees or so, from the vertical. At eight degrees, or even if the limb is adducted to the maximum possible, the femur is still so near the vertical that the large majority of the force exerted against it is directed horizontally. Thus, force exerted by the lateral wall creates the previously described lateral counter-pressure necessary to maintain the ischium on the medial brim and relatively little of the force is exerted in the vertical plane to provide weightbearing. The weightbearing potential of the femur is further limited by the cross-section of the femoral shaft and head. It might be mentioned in passing that studies have been conducted, by Gottschalk (1989), of Dallas Texas, that suggest that the femur in an IC socket is no more likely to be adducted than it is in a quadrilateral socket.

It is an article of faith by prosthetists that if the soft tissues of the stump are properly compressed and contained in a socket that weight can be borne by the tissues (hydrostatic weightbearing). It has been one of the goals of Sabolich (1985) and others to employ this concept in fitting the newer style sockets. The concept has been the subject of a study by Redhead (1979), who labeled it Total Surface Bearing and who reviewed his work in this area at the Miami Meeting (Schuch, 1988). Unfortunately, the concept was roundly condemned by Radcliffe and other engineers present at that meeting and, in light of the controversy, it would seem that no definitive statement about the role of soft tissue weight-bearing in IC Socket can be made.

In remarks made in Miami, Radcliffe (1989 b) suggested that the ischial ramus as well as the tuberosity was bearing weight in the IC Socket. When this was discussed in Chicago (Pritham, 1988) it was pointed out that the medial brim was so oblique and nearly vertical that only a small component of the force exerted by it would be in a vertical direction and thus the contribution of the ramus to weightbearing was questioned.

The matter of weight distribution in the socket is of more than academic interest. It may well be that the various proponents of IC fitting techniques, with their emphasis on weight-bearing on structures other than the ischial tuberosity, have succeeded in shifting the support point laterally. As was pointed out in the discussion of the principle of lateral stabilization, the closer the support point of the socket is aligned with the physiological hip joint axis, the less shear will be created by the medial brim. Redhead (1979) in his discussion of the Total Surface Bearing Socket made much the same point.

In the end however, it would seem that no more conclusive statement about weight-bearing in the IC Socket can be offered than that made by Radcliffe about the quadrilateral socket. It seems likely that something more than 50 per cent and less than 100 per cent of the weight is borne by the ischial tuberosity in the IC Socket, and that, in descending order, weight is also borne by the gluteus, the femur, and the anterior brim.

Alignment

In all the furore and debate over socket design one fundamental fact is often overlooked. Long's original objective was to study alignment of the prosthesis, not socket configuration. In a recent private communication he states—

"The original radiographical study in 1974 was to study femoral alignment — not socket shape. This study proved how little we knew about proper alignment of the above-knee prosthesis and led to the use of Long's Line for improved adduction angle. These x-rays were all with Quad sockets.

The need for a different socket shape became apparent. Not to achieve adduction, we could achieve adduction in the Quad socket, but we then had lateral gapping and discomfort.

I have never claimed that the socket shape gives you proper adduction. It does help to maintain it."

From this work in 1974 was spawned the concept of Long's Line (Long, 1975). This states that the normal femoral adduction angle can be approximated by positioning the end of the femur under the femoral head (the centre

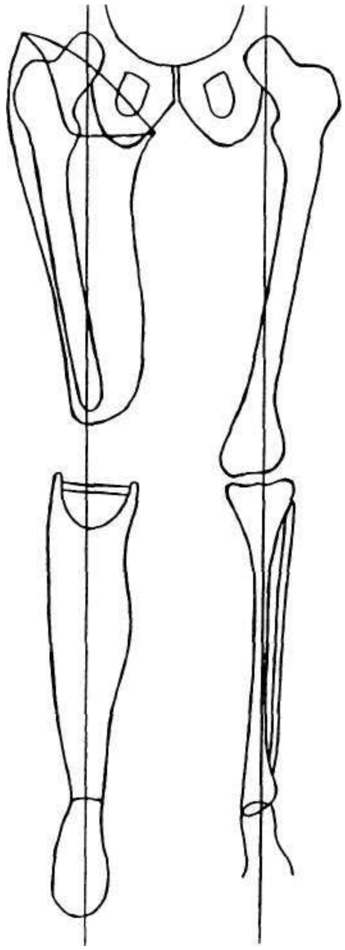


Fig. 10. Long's Line from: fabricating the Long's line above-knee prosthesis. (Long, 1985.)

of the head is approximated by bisecting the M-L dimension of the socket proximally). Further it maintains that a vertical line through these two points should be used for alignment in the frontal plane so that the knee is displaced laterally and the foot is centred on the line (Fig 10). The principle would seem to be that this comparative outset not only provides incentive for the amputee to adduct the femur as he strives to bring the foot in under him for proper balance, but it also provides clearance between the two legs thus permitting increased adduction.

The line described by the centres of the hip, knee, and ankle is of course the mechanical axis of the lower limb and was first described in the last century. Radcliffe (1955) alludes to alignment systems that centre the M-L dimension of the brim over the foot. What would seem to be original to Long is the concept of locating the femoral end on the line as well as the femoral head to determine the adduction angle.

From the foregoing, and from the work of Gottschalk et al. (1989), it would seem that quite quite possibly the operative factor influencing adduction angle in the frontal plane is alignment rather than socket design. Gottschalk would of course give primacy to efficient adductor muscles, while others would give the nod to socket design). Changes in socket configuration initially were made to ameliorate deficiencies in fit that emerged as a result of realigning the prosthesis, and to assist in maintaining the desired position. Eventually the process of reconfiguring the socket came to eclipse the matter of alignment. This brings us full circle and to consideration of ischial containment concepts.

Conclusion

The fundamental biomechanical principles governing behaviour of a prosthesis remain the same, independent of socket style. What differs is the strategy for dealing with these principles. An alteration in one or more basic features of a socket design affects others, and in a chain reaction, one socket configuration is inevitably transformed into another. The goal of this paper has been to demonstrate that once the decision to employ ischial containment in the AK socket has been made, the quadrilateral socket is inevitably changed into something different yet related. While different in shape and application of pressure, the two are related in that they both obey the principles of AK prosthetic behaviour, as described Radcliffe. In exploring this thesis, a variety of the crucial criteria describing an IC Socket have been discussed, but no attempt has been made to be exhaustive or all encompassing. It is hoped that this exercise will serve to put events of the past few years in perspective and clarify some of the issues involved.

It should be amply evident that a wide variety of issues remain unresolved. What is the support point in the IC Socket? What is the weightbearing distribution? Can the controversy over hydrostatic weightbearing be resolved? Can the questions raised by Dr.

Gottschalk's work be resolved? Can the claims made by advocates of IC style Sockets be verified? For whom is the IC Socket indicated? Contraindicated? What patient best benefits from which height and style brim? And last, but not least, can a readily applicable and teachable technique be developed so that the benefits of the IC Socket be made available equitably? These and a host of related questions would seem to give scope for investigators for quite some time to come.

As has been previously discussed, a good many of the claims made for the IC style sockets, while accepted as true for purposes of :his article, remain unsubstantiated by objective scientific investigation. There is sufficient experience, however, from a good many practitioners to support the claim that it is possible to fit a patient comfortably and functionally with such sockets. This body of evidence also shows that it is considerably more difficult to fit a patient with an IC socket than it is to fit him/her with a quadrilateral socket, and that considerably more experience is necessary in order to learn how to do it properly. The ultimate issue that must be resolved is whether or not the results justify the increased effort and aggravation.

Acknowledgements

Finally, the author would be remiss if he did not acknowledge with gratitude the contributions of the following people who reviewed the manuscript in its gestation: Carlton Fillauer, C.P.O., Gunter Gehl, C.P., Christopher Hoyt, C.P., Ivan Long, C.P., John Sabolich, C.P.O., Michael Schuch, C.P.O., and Daniel Shamp, C.P.O.. A particular note of gratitude is due to John Michael, C.P.O and A. Bennet Wilson, Jr., who reviewed and edited the article several times and made many helpful suggestions.

REFERENCES

BENNET, L. (1971). Transferring load to flesh — Part II. Analysis of compresssive stress. *Bull. Prosthet. Res.* 10(16)45-63.

GOTTSCHALK, F., KoUROSH, S., ROBERTS, J., STILLS. M., MCCLELLAN, B. (1989). Does socket configuration influence the position of the femur in above-knee amputation? AAOP Annual Meeting, February 3, Orlando, Florida.

HABERMAN, H. (1963). Above-knee prosthetic techniques in Germany. *Orthop. Prosthet. Appl. J.* 17,15-26.

HALL, C. B. (1964). Prosthetic socket shape as related to anatomy in lower extremity amputees. *Clin. Orthop.* 37, 32-46.

HOYT, C, LITTIG, D., LUNDT, J., STAATS, T. (1987). The UCLA CAT-CAM above-knee socket. Third Edition — Los Angeles: UCLA Prosthetics Education and Research Program.

LEHNEIS, H. (1985). Beyond the quadrilateral. *Clin. Prosthet. Orthot.* 9, 6-8.

LONG, I. A. (1975). Allowing normal adduction of femur in above-knee amputations. *Orthot. Prosthet.* 29, (4) 53-54.

LONG, I. A. (1985). Normal shape-normal alignment (NSNA) above-knee prosthesis. *Clin. Prosthet. Orthot.* 9, (4) 9-14.

MURPHY, E. F. (1971). Transferring load to flesh — Part I. Concepts. *Bull. Prosthet. Res.* 10 (16) 38-44.

MAYFIELD, G. W., SCANLON, J., LONG, I. (1977). A new look to and through the above-knee socket. *Orthop. Trans.* 1,(1)95.

PRITHAM, C. H. (1988). Workshop on teaching materials for above-knee socket variants, *J. Prosthet. Orthot.* 1, (1) 50-67.

PROSTHETIC CONSULTANTS. (1987). Manual for the use of the Shamp brim for the narrow M-L above-knee prosthetic socket — Sterling, Ohio; Ohio Willow Wood.

RADCLIFFE, C. W. (1955). Functional considerations in the fitting of above-knee prostheses. *Artificial Limbs.* 2, (1) 35-60.

RADCLIFFE, C. W. (1970). Biomechanics of above-knee prostheses. In: Prosthetic and Orthotic Practice edited by G. Murdoch, London: Edward Arnold. 191-198.

RADCLIFFE, C. W. (1977). The Knud Jansen Lecture, above-knee prosthetics. *Prosthet. Orthot. Int.* 1, 146-160.

RADCLIFFE, C. W. (1989 a). A short history of the quadrilateral above-knee socket In: Report of ISPO workshops. Edited by R. Donovan, C. Pritham, A. B. Wilson Jr. Copenhagen: ISPO. 4-12.

RADCLIFFE, C. W. (1989 b). Comments on new concepts for above-knee sockets In: Report of ISPO workshops. Edited by R. Donovan, C. Pritham, A. B. Wilson Jr. Copenhagen: ISPO. 31-37.

REDHEAD, R. G. (1979). Total surface bearing self-suspending above-knee socket. *Prosthet. Orthot. Int.* 3,126-136.

SABOLICH, J. (1985). Contoured adduction trochanteric-controlled alignment method (CAT-CAM). *Clin. Prosthet. Orthot.* 9 (4), 15-26.

SCHUCH, C. M. (1988). Report from: international workshop on above-knee fitting and alignment techniques. *Clini. Prosthet. Orthot.* 12 (2), 81-98.

Test instrument for predicting the effect of rigid braces in cases with low back pain

S. W. WILLNER

Department of Orthopaedics, Malmo General Hospital, Lund University, Malmo, Sweden

Abstract

The difficulty of predicting the acceptance and the result of wearing rigid braces has been identified before and is reported in the literature. Therefore a test instrument has been developed and tested. The intention is that the test instrument can imitate a rigid brace. Furthermore, different attributes of the rigid brace can be altered. Thus the range of the lordosis, the level of maximal dorsal support and the amount of abdominal support can be altered. By changing these parameters the maximal pain relief is sought. A good correlation between the result in the test instrument and the rigid brace manufactured according to the information from the former was seen (93%). No false negative results were seen. Thus, if no acceptance or pain relief was seen in the test instrument no pain relief could be expected in a rigid brace.

Another purpose of this test instrument is to simplify the manufacture of the brace and to transfer easily the information gained from the test instrument to the brace with the aid of a so called measuring device.

Introduction

Outer spinal supports are among the most frequently prescribed types of orthoses. There is no doubt that in many cases with different types of low back pain (LBP) they have a good effect.

However, sometimes braces are prescribed on nonspecific indications and patients do not experience the expected relief. It is therefore very important to make sure that the indication for wearing the brace is correct before it is manufactured.

The failure or brace treatment in **LBP** is caused mainly by a lack of knowledge of the pathomechanism of the spinal disorder and how the brace influences the pain.

In a Swedish study (Willner, 1985) the acceptance of wearing a rigid brace (Flexaform brace) varied between different diagnoses of the LBP. But on the average only 51% of all patients included in this study with low back pain accepted wearing a rigid brace and reported pain relief. That means that about half of the cases were not affected by a rigid brace and consequently had been prescribed braces with no effect.

To be able to predict the result of treatment with a rigid brace before it is manufactured and delivered to the patient, a special test instrument has been developed and tested. The main purpose of this test instrument is to imitate a rigid brace and to see whether this type of orthosis can be accepted by the patient and, if so, how it should be fitted. This instrument estimates the degree of the lordosis and the level of the maximal dorsal support to achieve optimal pain relief and acceptance. If no pain relief is achieved—there is, according to the author's experience, no indication for prescribing a rigid brace.

Another purpose of this test instrument is to make the fitting of the brace easier and more accurate by using a special measuring device for transferring the information from the test instrument directly to the brace module **to** be **fitted.**

The aim of this paper is to present this test instrument and describe its use in a group of patients with **LBP.**

All correspondence to be addressed to Dr. S. W. Willner, Department of Orthopaedics, Malmo General Hospital, S-214 O1, Malmo, Sweden.

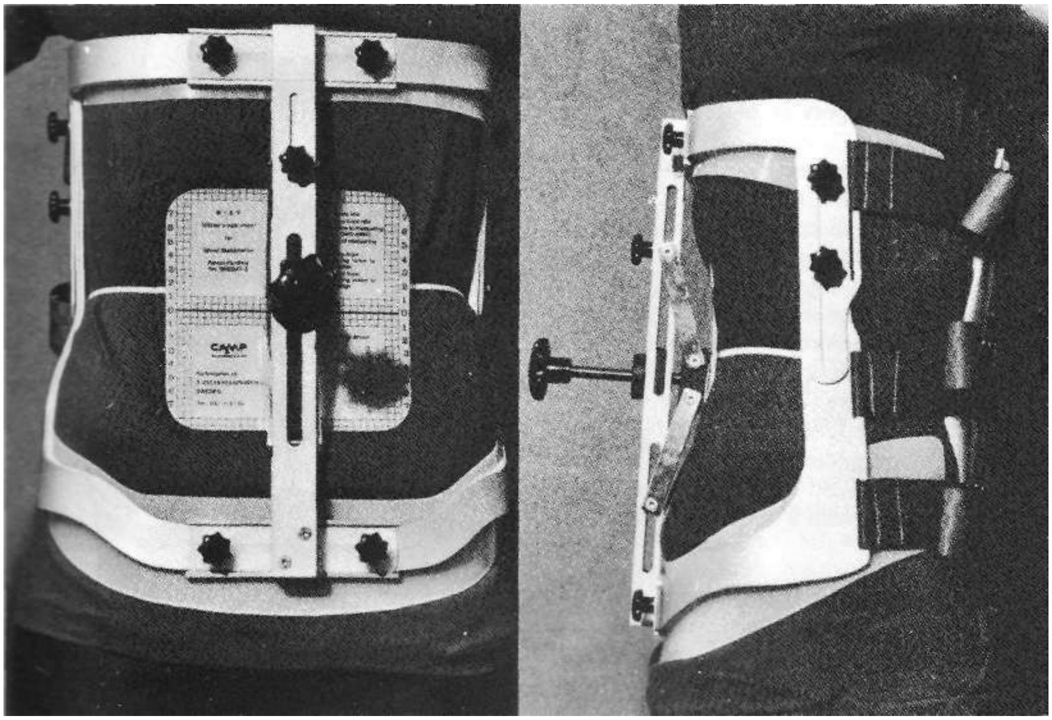


Fig. 1. Test instrument. Left, posterior view. Right, lateral view.

Method

The instrument consists of an aluminium frame. adjustable in width and length, which is to be applied to the back of the patient (Fig. 1).

At the back of the frame there is a back stay with an adjustable back support. This back support is adjustable in height and the lumbar lordosis can be increased or decreased by using an adjustment screw. At the front there is an abdominal support with six pull straps and adjustable fix locks.

A measuring device for the instrument is seen in. Figure 2. This instrument can be adjusted in height, has a locking control and a measuring screw. The device allows the transfer of the observed information to the brace module and simplifies the construction of the brace.

The dorsal frame is adjusted to fit the trunk. For this purpose, the curve of the frame must be placed just above the iliac crest to permit the transfer of information between the trunk of the patient, the test instrument and the brace to be fitted. The height of the instrument must correspond to the planned height of the brace to be fitted, that is, the upper edge of the frame

Should be just below the lower border of the scapula. Thereafter an abdominal support is

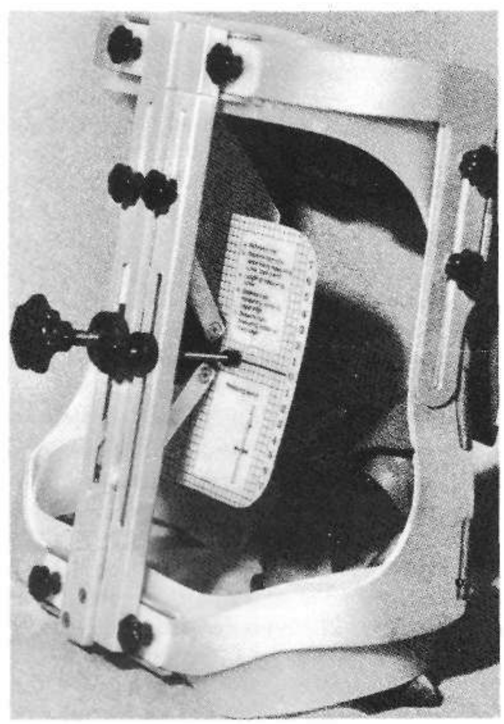


Fig. 2. The measuring device fixed on the test instrument.

applied and fixed to the frame with fix locks. The strips must be tightened to stabilize the instrument to the trunk. The lower border of the abdominal support is placed above the pubis.

The back support is now loosened and placed at the level at which the patient needs the maximal support. Thereafter the main dorsal screw (initially unscrewed as far back as possible) is adjusted to increase the range of the lordosis and the immobilization of the spine, until the patient reports optimal pain relief.

If the patient in this position can now move his back away from the back support, an abdominal pad is added under the abdominal support until a complete stabilization in the test instrument is attained. Two different sizes of pads are available (with a thickness of one and two cm).

The observed positions and the range of the back and abdominal support can be transferred to the brace by a measuring device.

However, to be able to transfer the information from the test instrument to the brace to be fabricated via the measuring stick, the following defined lines must be taken into consideration (Fig. 3).

(a) The reference line—the line joining the upper palpable corners of the iliac crest.

(b) The null line—the horizontal line on the back support marking the level of the maximal dorsal support.

(c) The central line—the horizontal line marked on the abdominal pad, the level for the maximal thickness of the pad.

With the aid of the measuring stick the following parameters are registered: (Fig. 2).

(1) The height of the brace to be fabricated.

(2) The level of the null line in relation to the upper and lower edges of the brace.

(3) The range of lordosis. This is measured by screwing the screw of the measuring stick until the top of the screw touches the back support at the level of the null line. The distance from the top of the screw to the measuring stick is recorded.

The level of the maximal thickness of the abdominal pad (if any) is decided by measuring the distance between the central line on the back of the pad and the lower edge of the abdominal plate, which should be placed over the pubis (Fig. 3). Before taking the test instrument off, the null line on the back support

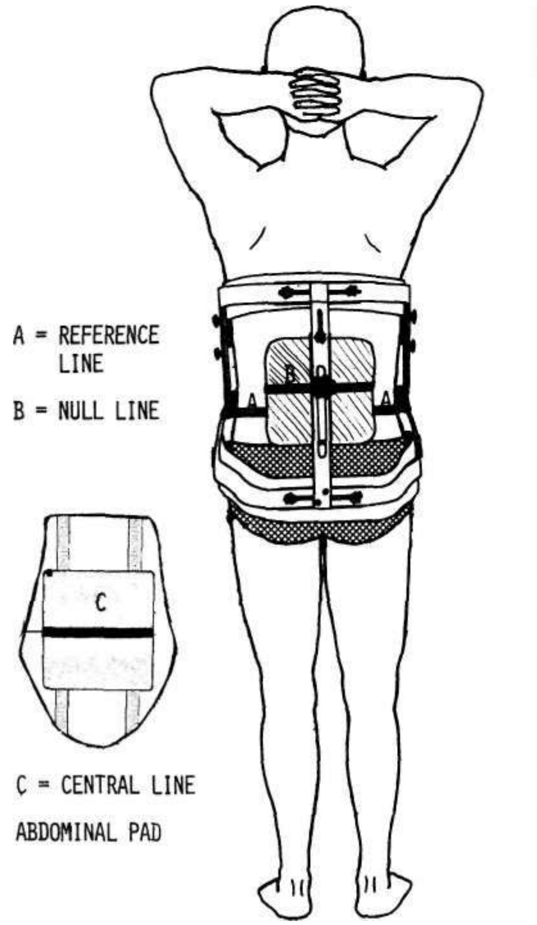


Fig. 3. Reference lines to allow the information observed in the test instrument to be transferred to the brace to be manufactured.

in relation to the reference line of the patient is measured expressed in cm above or below this reference line (Fig. 3).

All this information is recorded on a special form. At the workshop this information and another measuring device are sufficient for the manufacture of a well fitted brace. The reference line is easily identified on the brace module and is marked (Fig. 4, top). A hole in the mid line of the brace is drilled at the level of the null line.

Thereafter the measuring stick, adjusted according to the information on the form, is placed vertically on the module. The upper and lower borders of the brace are now easily decided. If the measuring screw comes through the brace and protrudes on the inner side, a pad of the same thickness must be made and placed in the brace (Fig. 4, bottom). If an abdominal

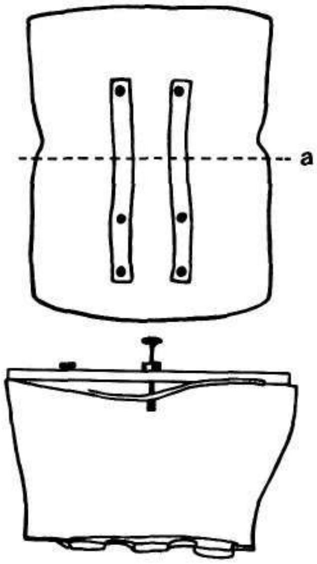


Fig. 4. Top, marking of the reference line (a) on the prefabricated module of a rigid brace. A hole is drilled at the level of the null line. Bottom, measuring device is placed on the module and the screw passes through the hole at the level of the null line.

pad has been seen used in the test instrument, a similar pad or bend in the anterior reinforcement is made with its maximal thickness or bending at the level of the central line.

The examination with this test instrument can be made at the doctor's surgery, the physiotherapy department, or if suitable, at the workshop. The test should go on for at least half an hour, during which time the patient tests the status in the standing, walking, bending forward and sitting positions. If there is any doubt concerning the test results these should pe repeated twice or even three times.

Patient tests

Between 1986-87, 88 consecutive cases which had been referred to the Spinal Unit in Malmo with the question "Indication for a rigid brace?" were investigated with the test instrument.

In this test material 42 cases were females and 46 males. The mean age at the time of the test was 43.6 years (range 20-70) in the men and mean age 44.4 (range 20-68) in the women.

Of these 88 cases 59 had received a rigid brace either before this test in 27 cases, or after the test in 32 cases. Of these 59 cases 12 had a spondylolisthesis, 5 spinal stenosis (all operated on and with failures) and 42 unspecified LBP with negative myelographies.

Table 1. Correlation between the result of the test instrument and of the rigid brace

		Result of brace treatment (Number of cases)	
		Good	Poor
Result in test instrument	Good	40	4
	Poor	0	15

Results

Of the 59 cases with prescribed rigid braces either before or after the tests, 40 cases (68%) had positive results in the test instrument as well as in the rigid brace (Table 1). In 15 cases (25%) negative results were seen, i.e. no pain relief was seen either in the test instrument or in the brace. In 4 cases (7%), all in the unspecified LBP group, a false positive finding was seen, i.e. a pain relief in the test instrument while a corresponding pain relief could not be achieved in the brace. No false negative findings were seen in the test instrument, i.e. if no pain relief could be noticed in the test instrument, no pain relief was seen in a rigid brace. As a consequence of this no braces were prescribed in the remaining 29 cases with unspecified LBP.

In 12 cases with spondylolisthesis there was a 100% positive correlation between the results in the test instrument and in the rigid braces which were fabricated according to the information found in the test.

In 5 cases with spinal stenosis operated upon, all with failures postoperatively, no positive results could be seen, either in the brace, or in the test instrument.

A correct prediction of the results of wearing rigid braces was made in 93% when using the test instrument, implying that this is a more accurate method than using different types of clinical estimations only (Winner, 1985) (Table 2).

Table 2. Comparison between the results in rigid braces either with or without using a test instrument.

Result in rigid brace (%)		Good	Poor
With test instrument	Spondylolisthesis	100	0
	LBP unspecified	90	10
Without test instrument	Spondylolisthesis	80	20
	LBP unspecified	35	65

Discussion

According to the literature, the observed frequency of accepting and wearing a brace varies. Ahlgren and Hansen (1978) observed that 75% of the patients with LBP wore their soft braces regularly. McKenzie and Lipscomb (1979) found an acceptance of corset wearing of only 50%. It was also seen that the utilization of brace wearing increased noticeably with increasing age of the patients. Magnusson and Nachemson (1985) reported that of those patients who had been prescribed a soft brace, 16% in the age group under 50 years were permanent brace wearers. Of patients between 60-69 years of age 50% wore their braces permanently and in those over 70 years of age 70%.

Concerning the acceptance of the rigid brace a variation was seen (Willner, 1985). This was especially observed in patients with unspecified LBP. About two thirds of these patients did not report any pain relief, or if they did, could not stand the brace, for example, because of its rigidity or unacceptable abdominal pressure. On the other hand, in cases of spondylolisthesis a high frequency of pain relief and acceptance was seen —85%. Even in cases with spinal stenosis verified by myelography a flexion brace gave pain relief in about 70% of the patients. In the group consisting of unspecified LBP only 15% of the patients experienced complete pain relief in rigid braces, i.e. many rigid braces were prescribed unnecessarily and in 20% only a partial pain relief in a rigid brace was achieved.

This shows that it is difficult to predict the effect of a rigid brace only by clinical estimation, especially in unspecific LBP.

Based on these observations the test instrument described was developed.

By changing the controlled parameters maximal pain relief was aimed at. With this test instrument it is possible to establish: 1) whether wearing a brace will be acceptable to the patient 2) and if so, how the brace should be contoured to give an optimal result.

In the present study this test instrument was studied in 59 cases, in which comparison could be made with a rigid brace already provided. In 93% of all these cases a corrleation was seen between the result of the test instrument and that in the rigid brace, positive as well as negative results.

This showed that if the patient did not experience any pain relief in the test instrument no pain relief could be expected in the rigid brace. That was the reason why braces were not prescribed in 29 of the 88 cases with negative results in the test instrument. In this study a low frequency of false positive findings was seen in the test instrument (7%). On the other hand no false negative findings were observed.

It was noticed that pain relief was experienced related to a very individually specific degree of the lordosis and when the maximal pressure of the dorsal plate was applied at a very specific level. With only a very small change in the degree of the lordosis or the level of maximal pressure of the dorsal plate, the pain returned and the acceptance of wearing the braces deteriorated.

Another reason for developing this test instrument was to be able to simplify the manufacture of the rigid brace. The information gained from the test instrument can easily be transferred to the brace by a measuring device. With this device the height of the brace, the degree of the lordosis and the level of the dorsal support are registered. Also the range of the abdominal support can be estimated.

Conclusion

A test instrument was developed which imitates a rigid brace. This instrument can, with a high degree of accuracy, predict whether a rigid brace will give pain relief in patients with LBP and also show how the brace should be manufactured to give optimal pain relief.

REFERENCES

AHLGREN, S. A., HANSEN, T. (1978). The use of lumbosacral corsets prescribed for low back pain. *Prosthet. Orthot. Int. 2,*101-104.

MAGNUSSON, A., NACHEMSON, A. (1985). Tygkorsetten som handikapp- och behandlingshjalpmedel-en enkatundersokning om dess anvandning. Handikappsinstitutet. Best-nr 6217.

MCKENZIE, A. R., LIPSCOMB, P. R. (1979). Corsets on and off (abstract). *J. Bone Joint Surg.* **61B,** 384.

WILLNER, S. W. (1985). Effects of a rigid brace on back pain. *Acta Orthop. Scand.* 56, 40-42.

Shock absorbing material on the shoes of long leg braces for paraplegic walking

F. BIERLING-S0RENSEN, H. RYDE, *F. BOJSEN-M0LLER AND **E. LYQUIST

Centre for Spinal Cord Injured, Department TH, Rigshospitalet, University of Copenhagen
* *Laboratory for Functional Anatomy, Department of Anatomy, University of Copenhagen.*
** *The Society and Home for Disabled, Department of Prosthetics and Orthotics, Copenhagen.*

Abstract

A study was designed to evaluate if shock absorbing material (ethyl vinyl acetate (EVA)) on the shoes of long leg braces could decrease the accelerations and consequent shock forces transmitted through the leg and brace during paraplegic walking. Six male paraplegics (26-55 years old) took part, four using a "swing-to" and two a "swing-through" technique when walking. Recordings comprised accelerometry of leg and brace, force platform measurement, and still photography of the trajectories of the leg segments. Each experimental condition was tested three times with a coefficient of variation (CV) for the measurements ranging from 5-22%. Compared to hard heels, shoes equipped with 20mm EVA soles decreased the acceleration amplitude in the first 10 msec as well as at maximum for shoe-to-ground contact. With the accelerometer at the malleolus reduction of the amplitude averaged 22% and 12% respectively, and 35% and 21% respectively with the accelerometer on the caliper (p: 0.03-0.1). In a second trial the two "swing-through" walkers had new shoes made with a 10mm thick EVA heel built in. After 3 months of walking with these shoes tests were carried out with the accelerometer attached to the malleolus both when the new and the former shoes were put on the calipers. CV for these measurements were 15-24%. It was found that the new shoes decreased the amplitudes by up to 62% and 26% on average (all $p<0.01$). The experimental subjects indicated that the EVA soles/heels gave a more comfortable and silent walk, e.g. the "bump" transmitted up through the body to the head diminished. In future, shock absorbing material should be built into the heels of shoes provided to long leg braces for paraplegic walking.

Introduction

Shock absorbing soles in shoes have been shown to reduce the shock transmitted up through the legs in walking and running (Bojsen-M0ller 1983, Light et al. 1980, Wosk & Voloshin 1985), and furthermore such soles have been found to be of advantage in relation to low back pain, and foot fatigue and stiffness (Dyer 1983, Wosk & Voloshin 1985), and may improve comfort and provide pain relief (Clark etal. 1989).

Paraplegics when walking with long leg braces have a heavy shoe to ground contact. Due to their spinal cord lesion they cannot feel the heel strike in the heel pad and they have no muscular function which can reduce the shock waves up through the body. In addition they have an abnormally low bone mineral content in the long bones of the lower extremities (Biering-S0rensen et al. 1988). These conditions imply that paraplegics are potentially more vulnerable to the heel-strike than normal persons.

The purpose of this study was to evaluate if the shock absorbing material EVA (ethyl vinyl acetate) attached to the shoes of long leg braces can decrease the acceleration and consequent shock forces transmitted up through the leg and long leg braces during paraplegic walking.

Participants and methods

Participants

A total of 6 spinal cord injured patients participated in the study (Tablel). They were all fully rehabilitated and trained to use long leg braces and forearm crutches. Participant No. 1

All correspondence to be addressed to Dr. F. Bierling-S0rensen, Centre for Spinal Cord Injured, Fysiurgisk Hospital, Havnevej 25, DK-3100, Hornbaek.

Table 1. Basic data for the participants.

Participant no.	Age at lesion (years)	Duration since lesion	Cause of lesion	Neurological incomplete	motor level complete	Weight (kg)	Height (cm)
1	18	29 years	Traffic accident	Th 4		62	178
2	21	19 years	Traffic accident	Th 7	Thlo	67	175
3	37	14 months	Fall		Thlo	68	180
4	36	20 years	Traffic accident		Thl2	78	181
5	24	20 months	Falling tree		Thl2	85	180
6	22	17 years	Gun shot	Thl2	L4	50	170

had some spasticity but medical treatment was not needed. The other subjects had flaccid paresis/paralysis of the lower limbs. None of the subjects had other lower limb problems which influenced their paraplegic walking.

Participants No. 1 and 6 used "swing-through" technique while the others used "swing-to" technique when walking. At the test sessions participant No. 3 walked in parallel bars, while the others used their crutches.

Measurements

To detect the accelerations in the legs and long leg braces a Philips PR 9367/20 unidirectional linear accelerations transducer was used. The accelerometer was connected through a 5m shielded cable to a Philips carrier frequency amplifier PR 9340. The cable was held during the experiment by an assistant to decrease movement artifacts and to eliminate interference with the paraplegic walking.

The signals from the amplifier were recorded on paper by a Siemens Mingograf 800 jetrecorder with a paper-speed of 5cm per sec.

Using the paper recordings the accelerations were described for every step by the maximum amplitude measured within the first 10 msec, i.e. corresponding to the heel-strike, and the overall maximum amplitude for the complete shoe to ground contact.

In one patient (No. 6) light emitting diodes (LED) (Bojsen-M0ller 1983) and still photography were used in combination with accelerometry and a force-time recording from a force-platform (AMTI^R). Both feet were placed on the platform while the crutches were outside. The LEDs were positioned on the leg brace at mid shank, at the heel and at the forefoot. The diodes were fed by a 50 Hz signal from which, however, one impulse was omitted each second. The 50 Hz signal was further registered on the oscillogram together with the signal from the accelerometer with the missing flash forming an exact time link between the recordings and the photography (Fig. 4, left and top right).

Procedures

Sole trial - EVA soles of 20mm thickness were taped to the subjects normal shoes.

The accelerometer was first taped to the medial malleolus of the right leg in a holder of plaster (Fig. 1) with a thin shell which fitted the malleolus to create the best possible contact to the skeletal system without using invasive techniques.

The participants were allowed 5 min. to get

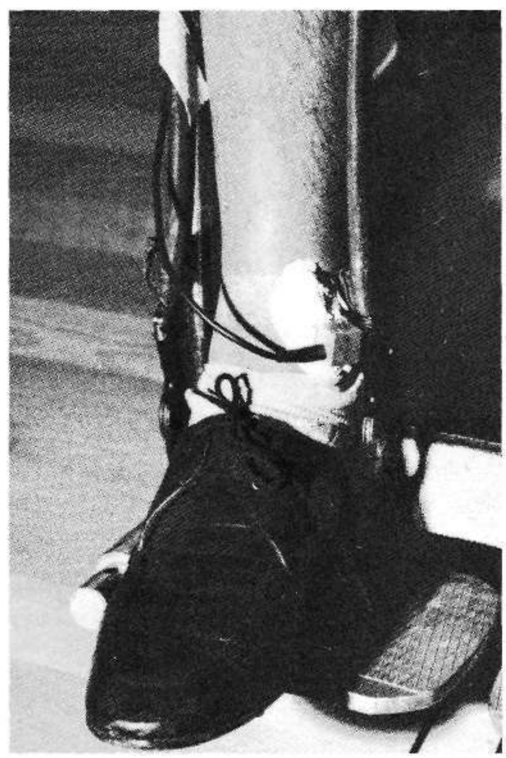


Fig. 1. Accelerometer fixed in a plaster holder and taped to the medial malleolus of the right leg, to create bony contact.

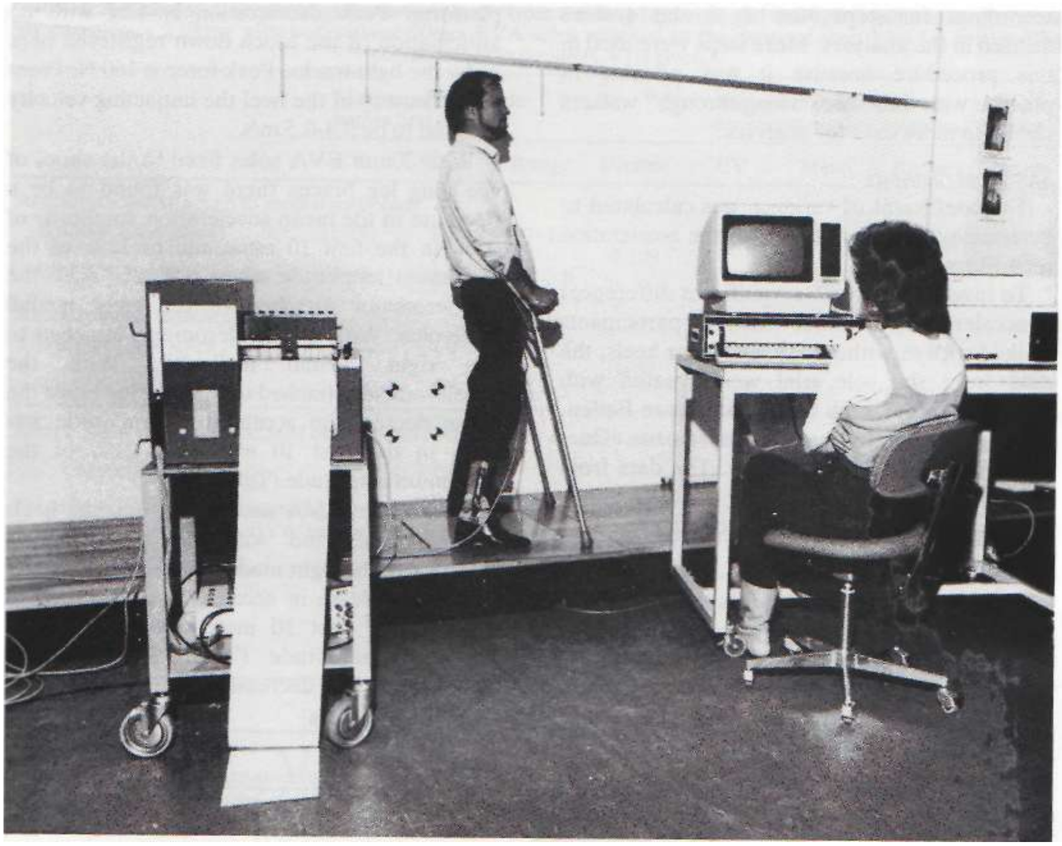


Fig. 2. Paraplegic walking with long leg braces and forearm crutches along a wooden walkway during an experiment with accelerometer attached to the medial malleolus.

used to the equipment. They were then asked to take four consecutive steps three times along a wooden walkway (Fig. 2) with the EVA soles on their shoes, and then three times four steps without the EVA soles.

Afterwards the accelerometer was taped to the long leg brace at the level of the right medial malleolus and the same walking procedure with and without the EVA soles was carried out.

In each of these experiments the accelerometer recording from step No. 2 was used for the analyses. Step No. 2, and not No. 1, was used to ensure that the participant had come into his usual gait pattern.

Heel trial - for participants Nos. 1 and 6, shoes were produced with 10mm EVA sandwiched into the heels (Fig. 3). The participants used these shoes for 2 to 3 months before they were re-tested with the accelerometer taped to the right medial malleolus. First they were tested four times with their new shoes with the 10mm sandwiched into the heels. Afterwards four times with the shoes they used previously.

The test procedure was otherwise the same as described above, except that the accelerometer

Fig. 3. Shoe with 10mm ethyl vinyl acetate sandwiched into the heel.

recordings for steps Nos. 2, 3 and 4 were utilized in the analyses. More steps were used in this procedure because it was possible in practice with these two "swing-through" walkers to obtain more data for analyses.

Statistical methods

The coefficient of variance was calculated to determine the reproducibility of the acceleration recordings.

To investigate possible significant differences in acceleration amplitude when the participants walked with or without EVA soles or heels, the data from the sole trial were treated with Wilcoxon signed-rank test (Kraft & van Eeden, 1968) on the differences of the means. One-tailed p-values were calculated. The data from the heel trials were tested by Mann-Whitney rank sum tests for each person separately.

Results

Figure 4 (bottom right) shows the parameters measured during foot to ground contact for participant No. 6, while walking with long leg braces and forearm crutches on the force platform. Peak deceleration is 4-5g with no anticipation of the touch down registered here or by the light tracks. Peak force is 160 N. From the trajectory of the heel the impacting velocity is found to be 0.4-0.5 m/s.

With 20mm EVA soles fixed to the shoes of the long leg braces there was found to be a decrease in the mean acceleration amplitude of 22% in the first 10 msec and of 12% of the maximum amplitude when recorded with the accelerometer attached to the right medial malleolus. With the accelerometer attached to the right medial malleolus. With the accelerometer attached to the long leg brace the mean decrease in acceleration amplitude was 35% in the first 10 msec and 21% of the maximum amplitude (Table 2).

With 10mm EVA sandwiched into the heels of the shoes and with the accelerometer attached to the right medial malleolus there was a mean decrease in acceleration amplitude of 62% in the first 10 msec and 26% of the maximum amplitude (Table 3). For both participants the decreases were found to be

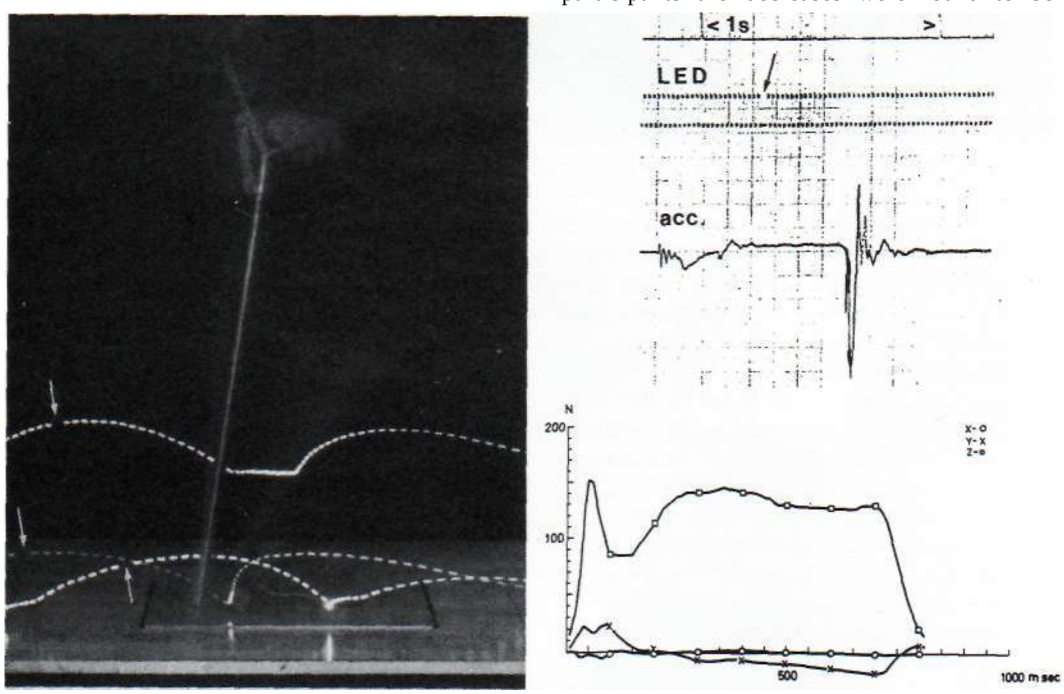


Fig. 4. Left, still photo of paraplegic walking on a walk way with built in force-platform. Light emitting diodes from three light tracks showing the trajectory of the shank, the heel, and the forefoot. Arrows indicate the missing signal and thereby the time link between the three tracks. Top right, oscillogram showing the 50Hz signal for the light emitting diodes (LED) and the accelerometry (ace.) from the same step as Fig. 4, left. Arrow indicates the missing signal. The record is significant in the absence of any anticipation of the impact. Bottom right, force-time curve from force-platform. Same step as in Fig. 4, left and top right. The z-curve (vertical force) shows the impact, but lacks a hump for the push off.

Table 2. Accelerometer-recordings (measured in G) from the long leg brace and the right medial malleolus from six paraplegics walking with and without 20mm EVA-soles attached to the shoes of their long leg braces. The coefficient of variation (CV) is given.

	Maximum amplitude within first 10msec.				Maximum amplitude for complete shoe to ground contact			
	CV	Mean	Range	P-value	CV	Mean	Range	P-value
Accelerometer attached to medial malleolus								
Without EVA soles	11.1%	1.2	0.7–1.6	0.109	22.3%	3.7	1.3–6.0	0.109
With EVA soles	19.9%	1.0	0.3–2.8		21.2%	3.2	1.0–4.8	
Accelerometer attached to caliper								
Without EVA soles	5.6%	4.9	3.6–6.5	0.031	4.8%	5.0	3.6–6.5	0.078
With EVA soles	20.9%	3.2	0.2–5.5		14.2%	4.0	2.0–5.5	

Each CV and mean is calculated on the basis of three trials for each participant, i.e. 18 measurements.

Table 3. Accelerometer-recordings (measured in G) from the right medial malleolus from two paraplegics walking with and without 10mm EVA sandwiched into the heels of the shoes of their long braces. The coefficient of variation (CV) is given.

	Maximum amplitude within first 10msec.			Maximum amplitude for complete shoe to ground contact		
	CV	Mean	Range	CV	Mean	Range
Without 10mm EVA	24.7%*	1.9	0.6–3.2	17.6%	2.4	1.6–3.2
With 10mm EVA	18.3%	0.7	0.4–1.0	15.3%	1.8	1.2–2.2

Each CV and mean is calculated on the basis of 12 steps for each participant, i.e. 24 measurements.
*excluding one outlier: CV=16.0%.

significant (in all instances p<0.01}.

In addition to the recorded accelerometer signals the subjects indicated that the EVA soles and heels gave a more comfortable and silent walk. The "bump" up through the body to the head was said to be diminished.

Discussion

The walking patterns used by paraplegics expose the heels and legs to an impact which they feel is uncomfortable and which may be harmful. The present investigation indicated that the paraplegic leg when walking on hard surfaces in the "swing-to" as well as in the "swing-through" technique is exposed to 3-4g at each touch down. However, placing the accelerometer on the skin although with a snug fit around the prominent malleolus rather than directly to the skeleton introduces an uncertainty and the deceleration may be even greater than that measured. This deceleration will produce a skeletal load which must be considered excessive especially for their fragile bones.

The lack of anticipation of the impact is noteworthy. Normally adjustments of muscle activity, joint position, and velocity of the heel are seen in the last 10-20 msec before heel contact. The paraplegics seem unable to perceive the shocks and to protect themselves against them. The reduction by 33% of the peak load by sandwiching a 10mm thick sheet of EVA foam into the heels of the shoes is one important result of this study.

A somewhat lesser reduction of the accelerations in the sole trial compared with the heel study was found. This might partly be due to the fact that the soles were externally taped to the shoes making them 20mm thicker in the soles. This can well have changed the pattern of walking, while in the heel study the normal walking pattern was possible.

Considering the sole trial with the major reductions in accelerations registered at the caliper it is noticeable how large were the reductions found in the accelerations recorded from the medial malleolus in the EVA heel study. Thus the results indicate that an even larger reduction in accelerations up through the long leg braces might be obtained by building EVA into the heels.

In addition to the significant reductions in accelerations the participating paraplegics claimed that the EVA soles/heels gave them a more comfortable walk.

Therefore the authors suggest it is justified to propose that all shoes for long leg braces for paraplegic walking in the future should have shock absorbing material built into the heels.

REFERENCES

BIERING-S0RENSEN, F., BOHR, H., SCHAADT, O. (1988). Bone mineral content of the lumbar spine and lower extremities years after spinal cord lesion. *Paraplegia*, 26,293-301.

BOJSEN-M0LLER, F. (1983). Biomechanical effects of shock absorbing heels in walking. In: Biomechanical aspects of sport shoes and playing surfaces. Edited by B. Nigg, B. Keer—Calgary: University of Calgary p. 73-76.

CLARK, J. E., SCOTT, S. G., MINGLE, M. (1989). Viscoelastic shoe insoles: Their use in aerobic dancing. *Arch. Phys. Med. and Rehabil.* 70, 37-40.

DYER, C. D. (1985). Visco-elastic insoles in long distance walking. *Bri. Osteopathol. J.* 15, 79-82.

KRAFT, C. H., VAN EEDEN, C. (1968). A non-parametric introduction to statistics. New York: Macmillan Co., p. 219-223.

LIGHT, L. H., MCLELLAN, G. E., KLENERMAN, L. (1980). Skeletal transients on heel strike in normal walking with different footwear. *J. Biomech.* 13, 477-80.

WOSK, J., VOLOSHIN, A. S. (1985). Low back pain: Conservative treatment with artificial shock absorbers. *Arch. Phys. Med. Rehabil.* 66,145-48.

The role of the contralateral limb in below-knee amputee gait

G. R. B. HURLEY[1], R. McKENNEY[1], M. ROBINSON[2], M. ZADRAVEC[3] and M. R. PIERRYNOWSKI[4]

Kinesiology Laboratory, School of Physical and Health Education, University of Toronto, Canada

Abstract

Very little quantitative biomechanical research has been carried out evaluating issues relevant to prosthetic management. The literature available suggests that amputees may demonstrate an asymmetrical gait pattern. Furthermore, studies suggest that the forces occurring during amputee gait may be unequally distributed between the contralateral and prosthetic lower limbs. This study investigates the role of the contralateral limb in amputee gait by determining lower limb joint reaction forces and symmetry of motion in an amputee and non-amputee population. Seven adult below-knee amputees and four non-amputees participated in the study. Testing involved collection of kinematic coordinate data employing a WATSMART video system and ground reaction force data using a Kistler force plate. The degree of lower limb symmetry was determined using bilateral angle-angle diagrams and a chain encoding technique. Ankle, knee and hip joint reaction forces were estimated in order to evaluate the forces acting across the joints of the amputee's contralateral limb. The amputees demonstrated a lesser degree of lower limb symmetry than the non-amputees. This asymmetrical movement was attributed to the inherent variability of the actions of the prosthetic lower limb. The forces acting across the joints of the contralateral limb were not significantly higher than that of the non-amputee. This suggests that, providing the adult amputee has a good prosthetic fit, there will not be increased forces across the joints of the contralateral limb and consequently no predisposition for the long-term wearer to develop premature degenerative arthritis.

All correspondence to be addressed to Mr. G. Hurley, Maritime Orthopedic, 274 Halifax Street, P.O. Box 2453, Station A Moncton, New Brunswick, Canada, E1C 8J3

1. Prosthetic Orthotic Department, The Rehabilitation Centre, Ottawa, Ontario.
2. Prosthetic Orthotic Department, Chedoke-McMaster Hospital, Hamilton, Ontario.
3. Clynch Prosthetics and Orthotics Laboratories Ltd., Calgary, Alberta.
4. Children's Seashore House Gait Laboratory, Elwyn Institute, Philadelphia, Pennsylvania, USA.

Introduction

There appears to be an increase in the number of below-knee amputees in our population due to ageing, accidents, and surgery related to peripheral vascular disease. This increase in the amputee population warrants research that attempts to address issues relevant to prosthetic management. It is clinically valuable to understand the role of the contralateral limb in amputee gait since, if the joint forces in the contralateral limb exceed natural limits, the individual may be predisposed to premature degenerative arthritis (Lewallen et al., 1986). In an attempt to equalize step length, improve balance and ensure knee stability, the prosthetist strives to achieve a symmetrical gait pattern when aligning and fitting an amputee with a prosthesis. Evaluating lower limb symmetry may therefore contribute to a better understanding of the role of the contralateral limb. The purpose of this study is to investigate the role of the contralateral limb in amputee gait by determining lower limb joint reaction forces and symmetry of motion in an amputee and non-amputee population.

Amputee gait has been evaluated both qualitatively (Gonzalez et al., 1974; Urban 1973; Waters et al, 1976; Kegel et al., 1981) and quantitatively in the literature. Quantitative research can be further subdivided into kinematic and kinetic studies. Many studies have presented descriptions of amputee gait based on kinematic measures (Enoka et al., 1982; Hannah and Morrison, 1984; Zuniga et al., 1972). James and Oberg (1973) and Murray et al. (1981), in similar studies of above-knee amputees, found significant differences in velocity, cadence, gait cycle, and stride length when their study group was compared to normal subjects. These authors noted that the step length of the prosthetic leg tended to be longer than that of the contralateral leg. Breakey (1976), in studies of below-knee amputees, reported that the stance phase of gait was longer in the normal limb and shorter in the amputated limb. Robinson et al. (1977) obtained time distance and accelerometer data from below-knee amputees. The subjects took longer steps more quickly on their prosthetic side and the resulting gait was described as asymmetric. Hershler and Milner (1980) also found asymmetry between the unaffected and the amputated side when looking at the variation of hip angle and knee angle throughout all phases of gait in above-knee amputees. Skinner and Effeney (1985) noted that this asymmetry of motion increases the excursion of die centre of mass during each cycle and thereby increases the energy cost of ambulation. These kinematic studies suggest that amputees demonstrate an asymmetric gait pattern. However this observation has not been verified using quantitative methods to determine the degree of symmetry.

Recently, Middleton et al. (1988) used lower limb symmetry as a criterion for evaluating the effects of a rigid ankle-foot orthosis and a hinged ankle-foot orthosis on a spastic diplegic cerebral palsied child. Kinematic and kinetic variables were determined using a video acquisition system and a Kistler force plate. Employing lower limb angle-angle diagrams and a chain encoding technique (Mcllwain and Jensen, 1985), differences in lower limb symmetry while unbraced, and in the braced conditions were determined.

Very little quantitative biomechanical literature is available that evaluates the mechanics of amputee gait utilizing kinetic analyses (Cappozzo et al., 1976; Golbranson, 1980; Lewallen et al., 1985). The majority of this research focuses on evaluating different prosthetic components with regard to amputee gait (Clark and Zernicke, 1981; Hoy et al., 1982; Zernicke et al., 1985). Winter and Sienko (1988) used sagittal plane biomechanical and EMG analyses from eight below-knee amputee trials to demonstrate modified motor patterns from the residual muscles at die hip and knee. Seven of the eight amputee trials were with SACH feet and showed a negligible knee moment of force during early stance (when non-amputees show an extensor moment), and a below normal knee moment of force in late stance. They explain that because of hyperactivity of the hamstrings during early stance there is an excessive knee flexor moment which is cancelled out by co-contracting knee extensors at that time.

Suzuki (1972) used a force plate to measure the three dimensional ground reaction forces on the limb during stance phase. He found the vertical ground reaction forces for the prosthetic and contralateral limbs to be different in subjects with below-knee, above-knee and hip disarticulation amputations. The vertical ground reaction force measured in below-knee amputees for the prosthetic limb was lower in magnitude with a smaller trough than the ground reaction forces measured on the contralateral side. Oberg and Lanshammar (1982) used a SELSPOT motion analysis system and force plate to study amputee gait. Knee moments and gait pattern-force vector diagrams were reported for one above-knee amputee. The authors noted differences between the subject's prosthetic and contralateral sides, however, they were only able to conclude that this type of analysis is valuable in evaluating amputee gait.

Lewallen et al. (1986) have produced the only study evaluating the development of amputee gait in children with respect to potential long term influences. This study compared kinematic and kinetic parameters of a normal and amputee paediatric population (6 amputees, 6 non-amputees) in an attempt to determine whether the loss of a limb segment results in increased forces across the intact joints of the normal limb. Quantitative analysis involved integration of force plate and cine data, and the inverse dynamics approach was utilized to estimate the joint moments in the intact limb. The authors reported that the normal leg in the child amputee displays reduced action and forces in order to achieve a better symmetry with the amputated leg. Furthermore, a tendency for the intact limb to have reduced forces involved in initial weight acceptance on the amputated limb was noted. It was concluded that the intact limb does not develop increased forces in the joints as compared with values for normals. This balance in the child amputee was achieved through slower walking velocity, decreased step length, and increased double support and stance phases as compared

TABLE 1 — Subject Demographics (A — amputee, S — non-amputee).

Subject	Age (yrs)	Height (M)	Weight (Kg)	Socket Design	Foot Component	Gait[2]	Amputation Year	Amputation Reason
A1	42	1.75	84.0	PTB[1]	Seattle[L]	Good	1973	Traumatic
A2	32	1.83	86.5	PTB	Flex[L]	Good	1984	Traumatic
A3	32	1.68	70.0	PTB	Seattle[R]	Good	1986	Congenital
A4	32	1.68	64.5	PTB	Seattle[R]	Fair	1987	Traumatic
A5	43	1.67	73.0	PTB	Seattle[R]	Excellent	1957	Traumatic
A6	42	1.81	98.0	PTB	Flex[R]	Fair	1986	Vascular
A7	26	1.70	60.0	PTB[1]	Seattle[L]	Good	1966	Traumatic

Subject	Age (years)	Height (M)	Weight (kg)
S1	26	1.77	71.0
S2	24	1.78	77.4
S3	27	1.80	81.1
S4	24	1.63	62.7

PTB = patellar-tendon bearing
(R) = right
(L) = left
(1) thigh corset & external hinges
(2) clinical subjective gait analysis

to his normal counterpart. The researchers concluded that, providing the child amputee has a good prosthetic fit, there will be no increased forces across the joints of the intact limb and consequently no predisposition towards premature degenerative arthritis. The conclusion drawn from this investigation is suspect since no statistical technique was employed when comparing joint forces between the amputee and non-amputee groups. These results are also limited since only one stride per subject was analyzed. Considering the supposed asymmetrical nature of amputee gait multi-stride analyses are warranted.

Methodology

Seven below-knee amputees and four non-amputees participated in the study. Information describing the subjects is presented in Table 1. None of the seven amputees was undergoing clinical prosthetic management at the time of resting. Prior to testing, prosthetic fit was checked and the amputee was questioned regarding his/ her evaluation of prosthetic fit. All of the amputees reported that they were satisfied with their present prosthesis. During testing, the amputee's gait was clinically evaluated and characterized as either poor, fair, good, or excellent. All amputee subjects were younger than 45 years of age since the ramifications of long-term wear were of interest. None of the subjects had other medical conditions which could potentially affect their performance during testing. Four non-amputee subjects were selected to obtain data representative of non- amputee gait. The subjects were voluntary participants and informed consent was obtained prior to testing.

Data collection

Testing involved collection of kinematic coordinate data using a WATSMART video system and ground reaction force data employing a Kistler force plate. Anthropometric measurements of each subject were taken in order that segment inertial parameters could be estimated. Force plate and kinematic coordinate data was collected for the left lower limb of the non-amputee subjects and the contralateral limb of the amputee subjects. Kinematic coordinate data was collected for the right lower limb of the non-amputee subjects and the prosthetic lower limb of the amputee subjects. An independent three segment link system was used to model the motion of the contralateral/left lower limb during ambulation. Since only kinematic data was being collected on the prosthetic/right lower limb an independent two segment link system was used to model the motion of this side.

A four camera WATSMART kinematic data acquisition system was used to acquire the two-dimensional positions of five anatomical landmarks on the left/contralateral lower limb and three anatomical landmarks on the right/ prosthetic side (50 hertz sampling rate). Eight one centimetre diameter disks containing 3 infra-red emitting diodes (IREDs) were placed over the anatomical landmarks. These landmarks approximated the positions of the anatomical joint centres of the hip, knee and ankle, and the proximal and distal ends of the foot segment on the left/contralateral lower extremity. IREDs were placed over the anatomical joint centres of the hip, knee and ankle of the non-amputee's right lower limb. The amputee's prosthetic lower limb was treated in a similar manner with the anatomical joint centres of the hip and knee located and a third IRED placed distally bisecting the longitudinal axis of the prosthetic shank in the sagittal plane. The cameras were placed perpendicular to the sagittal motion of the lower limbs. The subject was familiarized with the testing area in order to promote natural performances during data collection. Walking trials lasted 6-7 seconds and approximately 20 trials per subject were carried out.

Estimation of segment inertial parameters (mass and moment of inertia about the transverse proximal axis) were determined mathematically using regression equations (Jensen and Wilson, 1988). These regression equations employ selected anthropometric measurements (Hanavan, 1964) as predictor variables. Segment inertial parameters for the thigh, leg and foot of the amputee's contralateral limb and non-amputee's left limb were calculated in this manner.

Subsequent kinematic and kinetic analyses of coordinate data records were performed using the Waterloo Biomechanical Motion Analysis Software Package. The first eight trials in which the subject successfully contacted the force plate were used for analysis. Each walking trial was composed of one complete stride and both left and right lower limbs were analyzed. All joint coordinate data were filtered through a low pass recursive second order Butterworth digital filter using 5 hertz cutoff frequencies (Pezzack, 1977). Waterloo Programme input parameters were selected and employed in the established manner (Winter, 1979).

Statistical analysis

A chain encoding technique was used to quantify the degree of lower limb symmetry displayed by the subjects during ambulation (Mcllwain and Jensen, 1985). This technique may be used to determine the degree of congruity between any two XY patterns. Each XY data set is converted into a chain encoded data set. The chain encoded data consists of a numeric array of single digits (0-7) describing the direction followed by straight lines connecting the original XY data points plotted on a square aspect ratio XY graph. Cosine cross-correlation analysis is used to determine the degree of congruity between the two chain encoded data sets. The cross-relation function derived from these two generated chains, referred to as the recognition coefficient, was used to quantify the degree of lower limb symmetry.

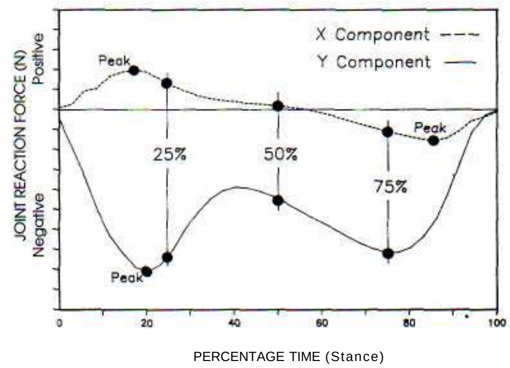


Fig. 1. Joint reaction force dependent variables.

A 2 X 8 (subject X walking trial) partially repeated two-way analysis of variance (ANOVA) design was used to determine if the joint reaction forces acting on the contralateral limb of the amputee were significantly greater than those on the non-amputee during ambulation. Nine values of ankle, knee and hip joint reaction force, during the support phase of walking, were selected as dependent variables. All joint reaction force variables were normalized with respect to the subject's body mass. The nine dependent variables (depicted in Figure 1) per joint are as follows:

1) maximum positive horizontal joint reaction force,
2) maximum negative horizontal joint reaction force,
3) maximum vertical joint reaction force,
4) horizontal joint reaction force at 25% of stance,
5) horizontal joint reaction force at 50% of stance,
6) horizontal joint reaction force at 75% of stance,
7) vertical joint reaction force at 25% of stance,
8) vertical joint reaction force at 50% of stance,

and 9) vertical joint reaction force at 75% of stance.

This design strategy was employed so that peak values as well as the forces occurring during the natural progression through stance could be evaluated. In all 27 ANOVAs, walking trial was the within factor and subject was the grouping factor. The within factor was employed in order to

TABLE 2 — Chain encoding results of angle-angle plots for non-amputee walking trials.

Subject	S2 Right	S3 Right	S4 Right	S1 Left	S2 Left	S3 Left	S4 Left
S1 Right	0.851	0.847	0.767	0.871	0.831	0.836	0.762
S2 Right		0.868	0.818	0.837	0.878	0.875	0.835
S3 Right			0.792	0.840	0.869	0.897	0.776
S4 Right				0.739	0.850	0.787	0.879
S1 Left					0.831	0.828	0.748
S2 Left						0.860	0.827
S3 Left							0.800

determine if any testing effects (ie fatigue, practice) were present. Subjects were divided into two groups — amputee and non-amputee. Statistical analysis was used to compare the joint reaction forces occurring in the amputee's contralateral limb and the non-amputee's during ambulation.

Discussion

This design strategy was employed so that peak values as well as the forces occurring during the natural progression through stance could be evaluated. In each of the 27 ANOVAs, walking trial was the within groups factor and subject was the between groups factor (Winer, 1971). The within factor (walking trial) was employed in order to determine if any testing effects (ie fatigue, practice) were present. Subjects were divided into two groups — amputee and non-amputee. Statistical analysis was used to compare the joint reaction forces occurring in the amputee's contralateral limb during ambulation to the non-amputee's.

Normal gait is characterized by symmetrical movements of the lower limbs throughout the gait cycle. By adopting such a gait pattern an energy efficient mode of ambulation is obtained (Skinner and Effeney, 1985). Furthermore the forces during weightbearing are distributed equally between both lower limbs. As the degree of lower limb symmetry decreases, it is possible that the forces during weightbearing may become unbalanced between the hip, knee, and ankle joints of both lower limbs. This study attempts to quantify the degree of lower limb symmetry since it has been proposed that amputees demonstrate an asymmetrical gait pattern.

Lower limb symmetry

Angle-angle diagrams traditionally depict two selected lower limb joint angle variations plotted against each other for corresponding instants of time (Mcllwain and Jensen, 1985; Hershler and Milner, 1980). For the purposes of this study, it was felt that absolute angular displacements of the thigh and leg segments best depicted the action of the lower limbs during ambulation. The absolute angular displacement of a limb segment is the inclination of this segment relative to the ground. Of the eight available strides, each subject's fourth walking trial was evaluated with regard to lower limb symmetry. Bilateral leg/thigh angle-angle diagrams were utilized in evaluating the degree of symmetry between the lower limbs. An estimate of congruity or similarity in shape between any two angle-angle configurations was obtained by chain encoding each pattern and then determining the cross-relation function from the two generated chains. This recognition coefficient (C) served as the criterion for intercurve comparisons (degree of symmetry). The recognition coefficient can vary from 0.0 to 1.0,

TABLE 3 — Chain encoding results of angle-angle plots for amputee walking trials: Contralateral versus prosthetic side (c-contralateral, p-prosthetic).

Subject	A1–P	A2–P	A3–P	A4–P	A5–P	A6–P	A7–P
A1–C	0.806	0.768	0.825	0.711	0.788	0.763	0.791
A2–C	0.855	0.798	0.844	0.789	0.849	0.816	0.797
A3–C	0.805	0.695	0.856	0.796	0.718	0.681	0.797
A4–C	0.836	0.801	0.828	0.735	0.828	0.803	0.806
A5–C	0.863	0.846	0.813	0.753	0.858	0.837	0.821
A6–C	0.861	0.796	0.852	0.773	0.821	0.768	0.850
A7–C	0.799	0.714	0.855	0.758	0.752	0.712	0.792

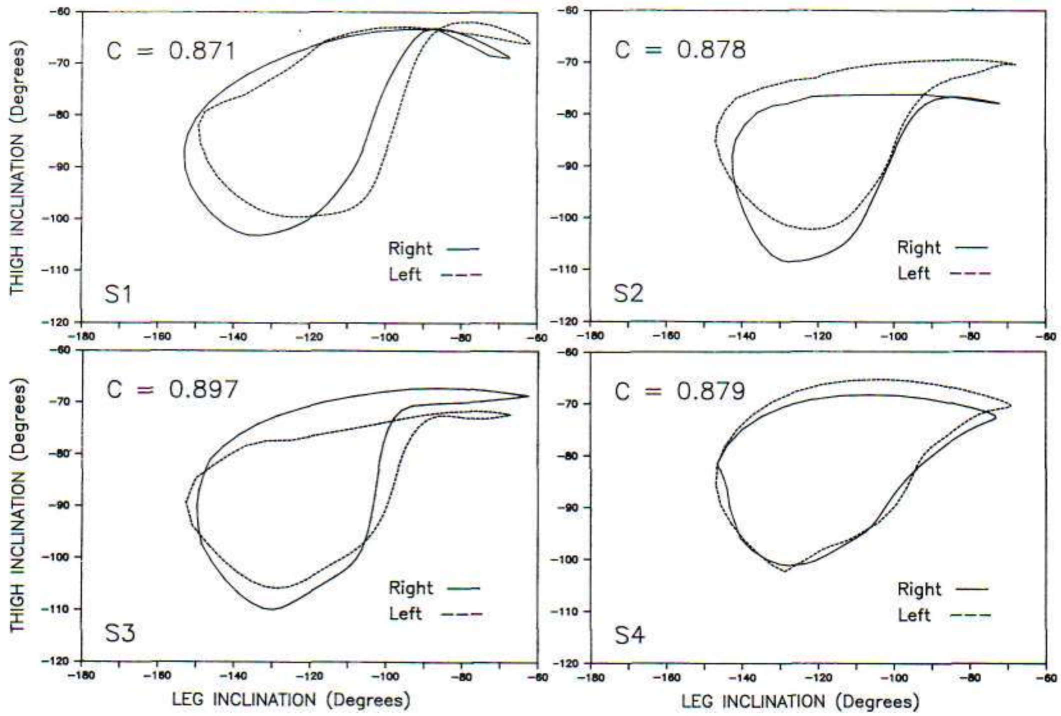


Fig. 2. Bilateral angle-angle plots of leg and thigh for one stride by non-amputees SI, S2, S3 and S4.

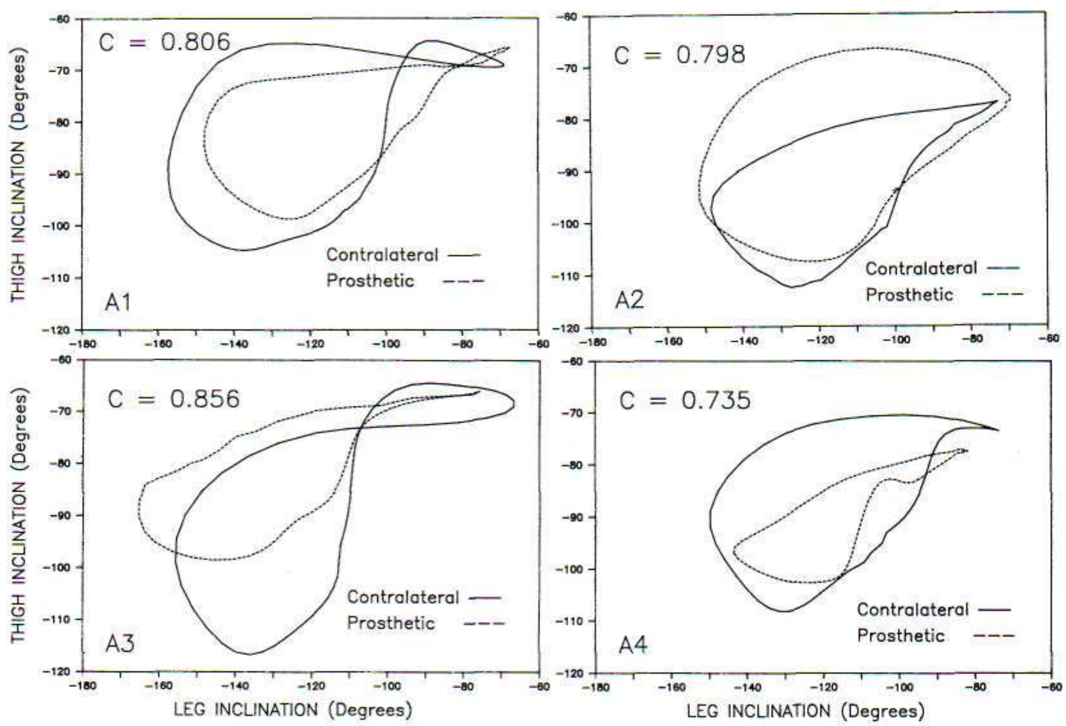


Fig. 3. Bilateral angle-angle plots of leg and thigh for one stride by amputees Al, A2, A3 and A4.

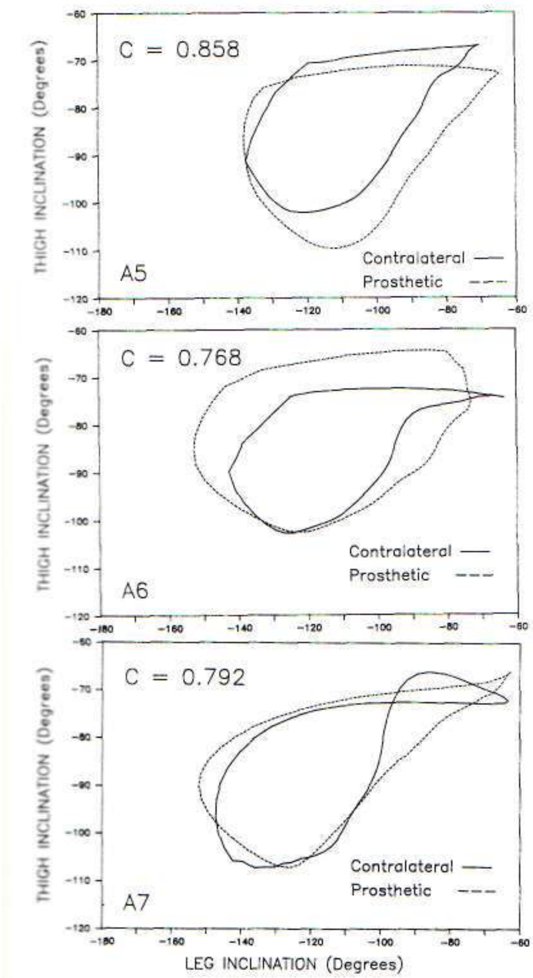


Fig. 4. Bilateral angle-angle plots of leg and thigh for one stride by amputees A5, A6 and A7.

with a value of 1.0 indicating perfect congruity and a value of 0.0 indicating absence of congruity between patterns, (Mcllwain and Jensen, 1985). This chain encoding technique was employed in determining the degree of lower limb symmetry for both the amputee and non-amputee groups.

Tables 2 and 3 present the results of chain encoding both right/left and prosthetic/ contralateral angle-angle plots for amputee and non-amputee subjects respectively (emboldened numbers). The non-emboldened values indicate between — subject variability for both groups. Figures 2 to 4 illustrate bilateral angle-angle diagrams where SI through S4 represent the non-amputee group and Al through A7 represent the amputee group. Non-amputees exhibited the highest degree of lower limb symmetry (mean 0.881, s.d. 0.011), whereas amputees demonstrated a lower degree of lower limb symmetry (mean 0.802, s.d. 0.044). Recognition coefficients, indicating degree of lower limb symmetry, ranged from 0.871 to 0.897 for non-amputees and from 0.735 to 0.858 for amputees. These results indicate that no amputee displayed a degree of lower limb symmetry equal to that of any non-amputee. The amputee demonstrating the highest recognition coefficient (A5, C = 0.858) has been a long time prosthetic wearer and was observed during testing to be an excellent walker. Amputee A4 displayed the lowest degree of lower limb symmetry (C = 0.735). Evaluation of bilateral angle-angle plots (Figure 3, subject A4) indicates limited movement on the prosthetic side suggesting a stiff-legged gait. It is interesting to note that this subject became an amputee quite recently (Table 1).

After evaluating all sound limb angle-angle diagrams for both the amputee and non-amputee groups it appears there exists a resemblance in the shape of the patterns between subjects. Conversely, the angle-angle diagrams depicting prosthetic side motion demonstrate a much more varied pattern between subjects. Tables 4 and 5 present recognition coefficient values evaluating between subject variability for contralateral and prosthetic sides, respectively. Between subjects, the contralateral limb exhibits a higher degree of lower limb symmetry (mean 0.833, s.d. 0.032) than the prosthetic side (mean 0.799, s.d. 0.054). The movements of the prosthetic lower limb may be characterized as more variable between subjects. These results indicate that non-amputees walk more symmetrically than amputees since movements of the prosthetic side do not mirror the sound or contralateral counterpart as well as a sound limb would.

Joint kinetics

Figure 1 illustrates a typical pattern of the horizontal and vertical components of the resultant force acting on any lower limb joint during normal level walking (Winter, 1987). During walking, the peak positive horizontal component of joint reaction force on the lower limb corresponds to weight acceptance and is initially forward in direction until approximately midstance. The negative horizontal component of joint reaction force occurs from approximately

TABLE 4 — Chain encoding results of angle-angle plots for amputee walking trials: Contralateral sides only.

Subject	A2	A3	A4	A5	A6	A7
A1	0.813	0.789	0.853	0.834	0.852	0.870
A2		0.809	0.844	0.868	0.847	0.816
A3			0.776	0.771	0.811	0.875
A4				0.848	0.842	0.840
A5					0.869	0.863
A6						0.820

Mean = 0.833 S.D. = 0.032

midstance to toe off with the peak corresponding to maximal forces acting during push off. The vertical component of the joint reaction force remains negative in direction (downward) throughout stance.

The results of the twenty-seven 2 x 8 (subject X walking trial) partially repeated two-way analyses of variance may be categorized into two areas: subject main effects and trial main effects. No significant trial main effect was displayed in any of the ANOVAs. This indicates that the likelihood of any error attributed to a testing effect, such as fatigue or practice, is negligible. No interactions between subject and trial were found.

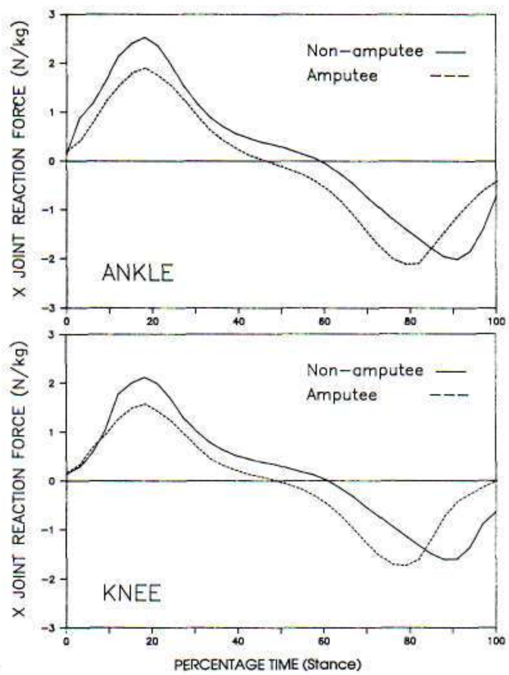


Fig. 5. Top, ankle horizontal joint reaction force, during stance, for an amputee and non-amputee subject. Bottom, knee horizontal joint reaction force, during stance, for an amputee and non-amputee subject.

Four significant subject (amputee/non-amputee) main effects were displayed in relation to the knee and ankle joint ANOVAs. No significant subject main effects were present for any ANOVA involving a hip joint reaction force dependent variable. The non-amputee demonstrated significantly higher peak positive horizontal components of joint reaction forces than the amputees in the study at both the ankle ($F(1,9) = 8.19$, $p < 0.05$) and knee ($F(1,9) = 10.26$, $p< 0.05$) joints. These effects were also demonstrated with regard to the values of ankle ($F(1,9) = 10.29$, $p < 0.05$) and knee ($F(1,9) = 7.13$, $p < 0.05$) horizontal component of joint reaction force occurring at 25% of stance. Figure 5 displays the ankle and knee horizontal components of joint reaction force occurring during stance for an amputee and non-amputee subject. Considering that peak horizontal joint reacton force occurs very close to the value corresponding to 25% of stance, it is understandable that significant effects were present for both of these dependent measures. The amputees experienced significantly lower ankle and knee horizontal components of joint

TABLE 5 — Chain encoding results of angle-angle plots for amputee walking trials: Prosthetic sides only.

Subject	A2	A3	A4	A5	A6	A7
A1	0.798	0.812	0.765	0.846	0.802	0.875
A2		0.738	0.721	0.853	0.861	0.832
A3			0.806	0.766	0.745	0.834
A4				0.711	0.709	0.767
A5					0.857	0.874
A6						0.809

Mean = 0.799 S.D. = 0.054

TABLE 6 — Walking Velocity (ms^{-1}).

Subject	Trial 1	Trial 2	Trial 3	Trial 4	Trial 5	Trial 6	Trial 7	Trial 8	Mean
S1	1.505	1.372	1.498	1.474	1.435	1.396	1.417	1.428	1.44
S2	1.367	1.434	1.400	1.378	1.413	1.415	1.451	1.437	1.41
S3	1.364	1.434	1.524	1.451	1.537	1.535	1.529	1.446	1.48
S4	1.713	1.595	1.935	1.486	1.937	1.784	1.850	1.477	1.72
Non-amputees									1.51
A1	1.223	1.335	1.445	1.196	1.652	1.237	1.260	1.332	1.34
A2	1.442	1.524	1.525	1.551	1.595	1.648	1.510	1.448	1.53
A3	1.408	1.547	1.760	1.639	1.441	1.523	1.495	1.443	1.53
A4	1.118	1.046	1.154	1.219	1.275	1.208	1.171	1.151	1.17
A5	1.293	1.251	1.288	1.270	1.232	1.268	1.143	1.238	1.25
A6	1.169	1.099	1.071	1.147	1.160	1.041	1.023	1.169	1.11
A7	1.338	1.306	1.410	1.351	1.282	1.215	1.243	1.253	1.30
Amputees									1.32

reaction force on their contralateral side, at weight acceptance, than non-amputees. This may be due to the less active push-off inherent to prosthetic componentry on the amputee's prosthetic side as compared to his non-amputee counterpart. Decreased mass on the prosthetic side, relative to an intact lower extremity, might also contribute to the lower horizontal component of joint reaction force displayed by the amputees during weight acceptance.

The results indicate that the amputees evaluated in this study did not experience increased forces across the joints of their contralateral limbs as compared to a group of non-amputees. These results are in agreement with the findings of Lewallen and colleagues 1986). These researchers also reported that the child amputee accomplished this balance by walking slower than his non-amputee counterpart. Table 6 presents the walking velocity for all subject trials analyzed. Differences in walking velocity between the amputee (mean = 1.32 ms^{-1}) and non-amputee (mean = 1.5 ms^{-1}) groups were present. Furthermore, 5 of the 7 amputees average walking velocity over 8 trials was lower than the slowest non-amputee (8 trial average). It appears that the adult amputee may employ a slower walking velocity, than his non-amputee counterpart, in order to decrease the forces acting across the joints of his contralateral **limb.**

Conclusions

It has been proposed that amputees demonstrate an asymmetrical gait pattern with regard to lower limb movement. This statement is supported in this study since the amputees demonstrated a lesser degree of lower extremity symmetry than the non-amputees. It is proposed that this asymmetrical movement may be attributed to the inherent variability of the actions of the prosthetic lower limb. Although amputees may demonstrate an asymmetrical gait pattern, it appears that the forces acting across the joints of the contralateral limb are not significantly higher than that of a non-amputee. This being the case, providing the adult amputee has a good prosthetic fit, there will be no increased forces across the joints of the contralateral limb and consequently no predisposition for the long-term wearer to develop premature degenerative arthritis.

Acknowledgements

The authors would like to thank the Model and Instrument Development Company, Seattle, Washington, U.S.A. who graciously donated a Seattle foot to all participating amputee subjects in consideration of their time and contribution. The subjects were not aware of this donation until they had completed testing.

REFERENCES

BREAKEY, J. (1976). Gait of unilateral below-knee amputees. *Orthot. and Proset.* 30, 17—24.

CAPPOZZO, A., FIGURA, F, LEO, T, MACTHETTI, M. (1976). Biomechanical evaluation of above-knee prostheses. *Biomechanics* V—A, 366—372.

CLARK, L. A., ZERNICKE, R. F (1981). Balance in lower limb child amputees. *Proset. Orthot. Int.* 5, 11—18.

ENOKA, R. M., MILLER, D. I., BURGESS, E. M. (1982). Below-knee amputee running gait. *Am. J. Phys. Med.* 62, 66-84.

GOLBRANSON, F. L. (1980). The use of gait analysis to study gait patterns of the lower-limb amputee. *Bull. Pros. Res.* 17, 96-97.

GONZALEZ, E. G., CORCORAN, P. J., RODOLFO, L. R. (1974). Energy expenditure in B/K amputees: correlation with stump length. *Arch. Phys. Med. Rehabil.* **55,** 111-119.

HANAVAN, E. P. (1964). A mathematical model of the human body. Technical Report, Aerospace Medical Research Laboratory TR-64-102, Wright-Patterson Air Force Base, Ohio, 1964.

HANNAH, R. E., MORRISON, J. B. (1984). Prosthetic alignment: effect of gait of persons with below-knee amputations. *Arch. Phys. Med. Rehabil.* 65, 159-162.

HERSHLER, C, MILNER, M. (1980). Angle-angle diagrams in above-knee amputee and cerebral palsy gait. *Am. J. Phys. Med.* 59, 165-183.

HOY, M. G., WHITING, W. C, ZERNICKE, R. F. (1982). Stride kinematics and knee joint kinetics of child amputee gait. *Arch. Phys. Med. Rehabil.* 63,74—82.

JAMES, U., OBERG, K. (1973). Prosthetic gait pattern in unilateral above-knee amputees. *Scand. J. Rehabil. Med.* 5, 35-50.

JENSEN, R. K., WILSON, B. D. (1988). Production of segment inertias using curvilinear regression. In: Proceedings of the 5th Biennial Conference and Human Locomotion Symposium of the Canadian Society for Biomechanics. Ed. C. Cotton — Ottawa, Ontario, Canada, 90-91.

KEGEL, B., BURGESS, B. M., STARR, T. W, DALY, W. K. (1981). Effects of isometric muscle training on residual limb volume strength, and gait of below-limb amputees. *Phys. Therap.* 62, 1419—26.

LEWALLEN, R., DYCK, G., QUANBURY, A., Ross, K., LETTS, M. (1986). Gait Kinematics in Below-Knee Child Amputees: a Force Plate Analysis. /. *Pediatr. Orthop.* 6, 291-298.

LEWALLEN, R., QUANBURY, A. C, Ross, K., LETTS, R. M. (1985). A biomechanical study of normal and amputee gait. In: Biomechanics IX—A. Proceedings of the 9th International Congress on Biomechanics. Ed. D. Winter — Champaign I11.: Human Kinetics, 587-592.

MCILWAIN, J. S., JENSEN, R. K. (1985). Congruity of normal gait patterns over a three-year period. In: Biomechanics IX—A. Proceedings of the 9th International Congress on Biomechanics. Ed. D. Winter — Champaign I11.: Human Kinetics, 411-416.

MIDDLETON, E. A., HURLEY, G. R. B., MCILWAIN, J. S. (1988). The role of rigid and hinged polypropylene ankle-foot-orthoses in the management of cerebral palsy: a case study. *Proset. Orthot. Int.* 12,129—135.

MURRAY, M. P., SEPIC, S. B., GARDNER, G. M., MOLLINGER, L. A. (1981). Gait patterns of above-knee amputees using constant friction knee components. *Bull. Pros. Res.* 17, 35—45.

OBERG, K., LANSHAMMAR, H. (1982). An investigation of kinematic and kinetic variables for the description of prosthetic gait using the ENOCH system. *Proset. Orthot. Int.* 6, 43-47.

PEZZACK, J. C, NORMAN, R. W, WINTER, D. A. (1977). An assessment of derivative determining techniques used for motion analysis. J. *Biomech.* 10, 377—382.

ROBINSON, J. L, SMIDT, G. L., ORORA, J. S. (1977). Accelerographic, temporal and distance gait factors in below-knee amputees. *Phys. Ther.* 57, 898—904.

SKINNER, H. B., EFFENEY, D. J. (1985). Gait Analysis in Amputees. *Am. J. Phys. Med.* 64, 82—89.

SUZUKI, K. (1972). Force plate study on the artificial limb gait. /. *Jap. Orthop. Assoc.* 46, 43—55.

URBAN, J. (1973). Effect of physical training in healthy male unilateral A/K amputees. *Scand. J. Rehabil. Med* 5, 88-101.

WATERS, R., PERRY, J., ANTONELLI, D., HISLOP, H. (1976). Energy cost of walking amputees: the influence of level of amputation. *J. Bone Joint Surg.* **58A,** 42-46.

WINER, B. J. (1971). Statistical principles in experimental design. 2nd Edition. New York: McGraw-Hill, p. 538.

WINTER, D. A. (1987). The Biomechanics and Motor Control of Human Gait. Waterloo, Ontario: University of Waterloo Press.

WINTER, D. A. (1979). Biomechanics of Human Movement. New York: Wiley.

WINTER, D. A., SIENKO, S. E. (1988). Biomechanics of below-knee amputee gait. *J. Biomech.* 21,361—367.

ZERNICKE, R. E, HOY, M. G., WHITING, W. C. (1985). Ground reaction forces and centre of pressure patterns in the gait of children with amputation: Preliminary report. *Arch. Phys. Med. Rehabil.* 66, 736-741.

ZUNIGA, E. N., LEAVITT, L. A., CALVERT, J. C, CANZONERI, J., PETERSON, C. R. (1972). Gait patterns in above-knee amputees. *Arch. Phys. Med. Rehabil.* 53, 373-382.

Prosthetics and Orthotics International, 1993, 17, 9–13

Incidence and prognosis of dysvascular amputations in Okayama Prefecture (Japan)

H. NAGASHIMA*, H. INOUE* and H. TAKECHI**

**Central Rehabilitation Service, Okayama University Hospital, Okayama, Japan*
***Kibikogen Rehabilitation Centre, Okayama, Japan*

Abstract

This survey analysed the clinical characteristics of subjects who first underwent major amputation of lower limbs necessitated by dysvascular disease during the 5 year period from 1984 to 1988. All were residents of Okayama Prefecture, Japan, and have been issued with a Physically Disabled Person's Certificate.

In total, 114 dysvascular amputees, representing 58.2% of all lower limb amputations performed in the resident population during the study period, were surveyed. The underlying disagnosis was arteriosclerotic obstruction in 64.9% of the subjects, diabetic gangrene in 22.8%, acute embolism in 7.0% and Buerger's disease in 5.3%. The yearly incidence of new dysvascular amputees per 100,000 people was estimated to be 1.2 among the general population and 5.7 among those aged over 65 years.

At three years after primary amputation, the survival rate was 52.3% in arteriosclerotic obstruction, and 66.7% in diabetic gangrene. Secondary amputation was performed in 17.0% of the entire group. The concurrent incidence of hemiplegic stroke was 19.8%

Among 36 amputees due to arteriosclerotic obstruction, who survived 3 years postoperatively, 10 (27.8%) were fitted with prosthetic limbs.

All correspondence to be addressed to Dr. H. Nagashima, Central Rehabilitation Service, Okayama University Hospital, 2-5-1 Shikata-cho, Okayama-shi, Japan.

Introduction

Although numerous reports have investigated the epidemiology and prognosis of amputations necessitated by dysvascular disease, only a few have surveyed the status of amputations in a defined region. The true, overall picture cannot be extrapolated from any experiences at isolated centres or statistical data on subjects with prosthetic limbs. The results would of course also be influenced by the patient criteria, such as whether or not finger amputations were included. Denmark, the only country which has established a national registration system, has demonstrated an annual increase in the number of dysvascular amputations. The present study, designed to derive an understanding of the status of amputations performed in Okayama Prefecture, having a population of roughly 2 million inhabitants, was carried out to support the reassessment of integrated rehabilitation strategy.

Subjects and methods

At the start, amputees were identified through the resources of the Welfare Office, which maintained generally accurate data, derived in the process of issuing Physically Disabled Person's Certificates, to patients with amputations performed in the area. As the possession of this certificate conveys significant economic benefits, virtually all amputees are expected to apply for and be certified within a few months postoperatively.

Table 1. Number of new amputations proximal to the transmetacarpal or transmetatarsal level.

Years	Total	Upper-limb	Lower-limb
1979–83	225	64	161
1984–88	226	30	196

The total number of amputees from various causes was 226, undergoing primary procedures during the 5 year period from January 1984 to December 1988. Data for upper limb were collected for amputations proximal to the transmetacarpal level, and those for the lower limb were collected for amputations proximal to the transmetatarsal level. Among the 226 subjects, 30 (13.3%) underwent amputation of the upper limb and 196 (86.7%) amputation of the lower limb (Table 1).

Among 196 lower limb amputees, 114 were due to dysvascular diseases. The main purpose of this survey was the analysis of these dysvascular amputees. Follow-up data for the 114 dysvascular amputees, for the period up to November 1991, were obtained through personal letters with attached questionnaires, hospital records or census registration offices. This allowed for the prognoses of 106 subjects (93%) to be investigated retrospectively. The interval between the primary amputation procedure and follow-up ranged from 3 to 8 years. The population of Okayama Prefecture was 1.91 million inhabitants in 1984 and 1.93 million in 1988.

Results

Incidence

The 114 new dysvascular amputations which took place from 1984 to 1988 were all in the lower limbs, 1.5 times the number for the previous 5 year period. In particular, there was an increase in amputations necessitated by arteriosclerotic obstruction (ASO) and gangrene associated with diabetes mellitus (Table 2).

The underlying diagnosis in the 114 amputees was ASO in 64.9%, diabetic gangrene in

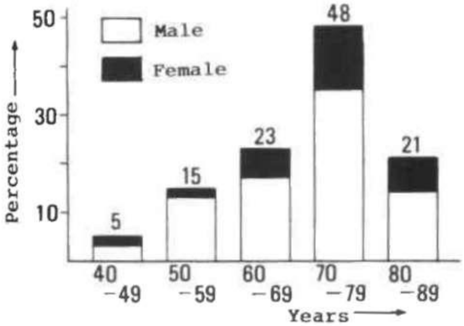


Fig. 1. Age and sex distribution in 112 dysvascular amputees (the figure does not contain information about 2 males aged 24 and 91 years).

22.8%, acute embolism in 7.0% and Buerger's disease in 5.3%. In ASO, acute embolism and Buerger's disease, the concomitance of diabetes mellitus was not obvious. The age distribution ranged from 26 to 91 years. By age, subjects in their 70's constituted the largest group, with those over 65 years accounting for 67.5% of the total. Males outnumbered females, the overall ratio being 3:1 and for the 80's age group 2:1 (Fig. 1).

For each disease category the mean age at amputation was 74.5 years (47-91) for ASO, 61.4 years (46-75) for diabetic gangrene, 65.0 years (26-84) for acute embolism, and 57.7 years (45-74) for Buerger's disease. The level of amputation was selected to be above the knee in 71.6% of ASO and only one (3.8%) of diabetic gangrene. On the other hand, amputation below the knee was performed in 23.0% of ASO and 88.5% of diabetic gangrene (Fig. 2).

Survival

Survival rates at 3 years after primary amputation was 52.3% in ASO and 66.7% in diabetic gangrene (Table 3).

The 3 year survival rates by age group in ASO were 51.4% for the 70's and 35.0% for the 80's (Table 4).

Secondary amputation, i.e. ipsilateral reamputation or contralateral amputation, was performed in 15 ASO cases (21.7%), and in 1

Table 2. Causes of new lower-limb amputation from the transmetatarsal level proximally.

Years	Total (n)	Dysvascular (%)	Trauma (%)	Tumour (%)	Miscellaneous (%)
1979–83	161	45.3	25.5	21.7	7.5
1984–88	196	58.2	26.2	9.7	6.1

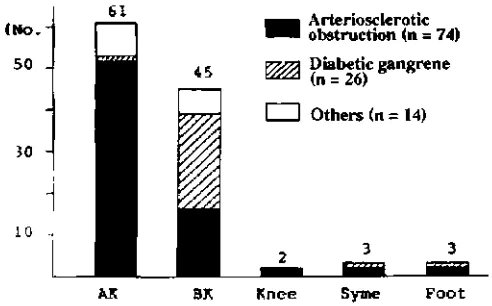


Fig. 2. Level of primary amputation, 1984-1988.

case each of diabetic gangrene and acute embolism. There were 9 bilateral amputees, 8 of which had ASO (Table 5).

Complications of hemiplegic stroke occurred in 17 ASO cases (24.6%), 1 case of diabetic gangrene and 3 cases of acute embolism.

Prosthetic appliances

Prosthetic appliances were distributed according to the Welfare Law to 12 of 69 ASO cases (17.4%) and 13 of 24 diabetic gangrene cases (54.2%). The distribution to 3 year survivors was 10 of 35 ASO cases (28.6%) and 10 of 16 diabetic gangrene cases (62.5%).

Discussion

Reports of the epidemiology and prognosis of amputees in a given area are relatively rare. Reports from Denmark (Danish Amputation Register, DAR) provide accurate statistics on a national scale, but other countries have only issued partial statistical data. In Scandinavia, the annual incidence of new amputations is 30-40 per 100,000 inhabitants, with dysvascular amputations accounting for about 90% (Ebskov, 1986; Larsson and Risberg, 1988; Pohjolainen *et al.*, 1989). Corresponding figures are somewhat lower in the United States (Bradway *et al.*, 1984) and United Kingdom (Sethia *et al.*, 1986; Murdoch *et al.*, 1988), and appear to be even still lower in Asia (Hla Pe, 1988; I-Nan Lien, 1989). In Japan, epidemiological surveys on amputees are practically nonexistent. This report presented the results for the single region of Okayama Prefecture. Our survey findings indicated that the mean annual incidence of new, major lower limb amputations is about 2 per 100,000 inhabitants, with dysvascular amputations accounting for about 60% of the total. In terms of the population aged over 60 years, the respective values were 6.4 per 100,000 inhabitants and 78.1%. These figures are forecast to continue to rise hereafter, in line with the increasing trend throughout the world.

The ratio of ASO to diabetic gangrene was 3:1. Although amputations necessitated by diabetes mellitus are tending to decline in Denmark (Ebskov, 1988), the results of this survey demonstrated that amputations necessitated by ASO as well as those for diabetic gangrene are comparably increasing. As the prevalence of aortic stenosis is generally claimed to be lower in Orientals than in Western people (Gore *et al.*, 1960), this disparate trend may be ascribed to differences in genetic characteristics and diet. However, recent years have witnessed a dramatic increase in protein and fat intake in Japan, combined with remarkable aging of the population. Although the current incidence of dysvascular amputation is 1/30 that of Scandinavia, the difference will most likely become smaller in the near future.

Table 3. Survival rates (%) after primary amputation.

	1-year	2-year	3-year
Arteriosclerotic obstruction (n=69)	89.9	72.5	52.3
Diabetic gangrene (n=24)	91.7	84.0	66.7

Table 4. Percentage surviving in 69 amputees for arteriosclerotic obstruction within different age groups.

Age (years)	Years after amputation 1	2	3
<60 (n=5)	100	100	100
60-69 (n=9)	100	88.9	66.7
70-79 (n=35)	82.8	71.4	51.4
≥80 (n=20)	90.0	60.0	35.0

Table 5. Incidence of secondary amputation.

	Ipsi-lateral	Contra-lateral	Total
Arteriosclerotic obstruction (n=69)	7	8	15
Diabetic gangrene (n=24)	1	1	2
Acute embolism (n=8)	0	0	0
Buerger's disease (n=5)	1	0	1

The level of primary amputation was above-knee in 71.6% of ASO cases and below-knee in 88.5% of diabetic gangrene cases. There are very few reports investigating the ratio of above-knee amputations in ASO in a given region. Based on a survey of Copenhagen, Denmark, in the 1970's, Jensen *et al.* (1982) reported that the ratios of above-knee, through-knee and below-knee amputations in ASO were 35%, 23% and 42%, respectively. The relatively fewer above-knee amputations of diabetic gangrene cases in the present study was in sharp contrast to the Scandinavian study (Ebskov, 1983; Christensen *et al.*, 1988). This difference may reflect the lower incidence of severe diabetes mellitus in Japan.

Three year survival rates were about half in ASO and close to 70% in diabetic gangrene. Twenty in ASO cases (29%) survived for at least 3 years without secondary amputation or hemiplegic stroke. These figures are far better than the inaccurate estimations that existed before the survey. Moreover, after 4 years survival rates continued to decline in ASO, whereas it appeared to remain generally stable in diabetic gangrene. Pohjolainen *et al.* (1989) investigated survival rates by disease in Southern Finland in the 1980's. They reported that the 2 year survival rates were found to be 38% in ASO and 48% in diabetes mellitus. Their subjects with the term of diabetes mellitus may be amputees caused from diabetic gangrene. The results of the present survey demostrated 2 year survival rates to be 60% in ASO even for patients in their 9th decade, while that for diabetic gangrene was even higher. The incidence of secondary amputation was 21.7% in ASO and 8.3% in diabetic gangrene. Compared with other reports, these figures were moderate for ASO and very low for diabetic gangrene (Ebskov, 1983; Stirnemann *et al.*, 1987; Murdoch *et al.*, 1988; Pohjolainen *et al.*, 1989). The low rate of secondary amputation in diabetic gangrene may be attributed to the fact that Syme's or foot amputation was not frequently employed.

The use of prosthetic appliances in ASO amputees of subjects in the present study was low similar to some reports (Malone *et al.*, 1981; Stirnemann *et al.*, 1987; Beekman and Axtell, 1987; Pohjolainen *et al.*, 1989). In this survey, the above-knee ASO amputees without secondary amputation or hemiplegic stroke, represented 32 cases (46.4%) of all ASO amputees. Thirteen of these amputees survived for more than 3 years, but only three possessed a prosthetic limb according to the results of the questionnaire (Table 6).

Table 6. Ambulatory appliances distributed through the Welfare Law among unilateral above-knee amputees caused from arteriosclerotic obstruction, without secondary amputation or hemiplegia.

	Prosthesis	Wheelchair	None
Total (n=32)	6.3%	50.0%	50.0%
3-year survivors (n=13)	15.4%	53.8%	38.5%

The growth in the aged segment of the population is particularly dramatic in Japan. Statistics of the World Health Organization forecast that the proportion of the population aged over 65 years will exceed 23% in Japan by the year 2020. The incidence of dysvascular amputations, which is particularly high in aged patients, is consequently forecast to further increase hereafter. Japan is thus facing a situation where integrated planning of the treatment and welfare of amputees must undergo serious reconsideration.

REFERENCES

Beekman CE, Axtell LA (1987). Prosthetic use in elderly patients with dysvascular above-knee and through-knee amputations. *Phys Ther* **67**, 1510–1516.

Bradway JK, Malone JM, Racy J, Leal JM, Poole J (1984). Psychological adaptation to amputation: an overview. *Orthot Prosthet* **38(3)**, 46–50.

Christensen KS, Falstie-Jensen N, Christensen ES, Brøchner-Mortensen J (1989). Results of amputation for gangrene in diabetic and non-diabetic patients: selection of amputation level using photoelectric measurements of skin perfusion pressure. *J Bone Joint Surg* **70A**, 1514–1519.

Ebskov B (1983). Choice of level in lower extremity amputation — nationwide survey. *Prosthet Orthot Int* **7**, 58–60.

Ebskov B (1986). The Danish Amputation Register 1972-1984. *Prosthet Orthot Int* **10**, 40–42.

Ebskov B (1988). Trends in lower extremity amputation (Denmark, 1978-1983). In: Amputation surgery and lower limb prosthetics./ edited by Murdoch G, Donovan RG. — London: Blackwell Scientific Publications. p3-8.

Gore I, Robertson WB, Hirst AE, Hadley GG, Koseki Y (1960). Geographic differences in the severity of aortic and coronary arteriosclerosis; United States, Jamaica, W.I., South India and Japan. *Am J Pathol* **36**, 559–574.

Hla Pe (1988). A 15 year survey of Burmese amputees. *Prosthet Orthot Int* **12**, 65–72.

I-Nan Lien (1989). The incidence and trend of limb amputation. In: VI World Congress, International Society for Prosthetics and Orthotics, 12-17 November 1989, Kobe, Japan.

Jensen JS, Mandrup-Poulsen T, Krasnik M (1982). Wound healing complications following major amputations of the lower limb. *Prosthet Orthot Int* **6**, 105–107.

Larsson PA, Risberg B (1988). Amputation due to lower-limb ischemia. *Acta Chir Scand* **154**, 267–270.

Malone JM, Moore W, Leal JM, Childers SJ (1981). Rehabilitation for lower limb amputation. *Arch Surg* **116**, 93–98.

Murdoch G, Condie DN, Gardner D, Ramsay E, Smith A, Stewart CPU, Swanson AJG, Troup IM (1988). The Dundee experience. In: Amputation surgery and lower limb prosthetics./ edited by Murdoch G, Donovan RG. — London: Blackwell Scientific Publications. p440—457.

Pohjolainen T, Alaranta H, Wikstrom J (1989). Primary survival and prosthetic fitting of lower limb amputees. *Prosthet Orthot Int* **13**, 63–69.

Sethia KK, Berry AR, Morrison JD, Collin J, Murie JA, Morris PJ (1986). Changing pattern of lower limb amputation for vascular disease. *Br J Surg* **73**, 701–703.

Stirnemann P, Mlinaric Z, Oesch A, Kirchhof B, Althaus U (1987). Major lower extremity amputation in patients with peripheral arterial insufficiency with special reference to the transgenicular amputation. *J Cardiovasc Surg* **28**, 152–158.

Prosthetics and Orthotics International, 1993, 17, 14–20

Dundee revisited — 25 years of a total amputee service

C. P. U. STEWART and A. S. JAIN

Dundee Limb Fitting Centre, Dundee, Scotland

Abstract

The Dundee Limb Fitting Centre has provided an integrated rehabilitation programme for the amputee since 1965.

A review of 1846 primary amputees is discussed.

During this period a dramatic change in the above-knee/below-knee (AK/BK) ratio has been achieved with 71% BK and 26% AK occurring in 1989.

Over 80% of all amputees, the majority being elderly with peripheral vascular disease, were successfully fitted with a prosthesis.

Final discharge home or to a residential home for the elderly was achieved in 76.2% of cases with 3.6% dying in the Unit.

Bilateral amputation occurred in 18% of cases of whom 48% were BK-BK.

Overall 66% were successfully fitted with a prosthesis.

The results demonstrate the advantages of an integrated approach to the amputation and consequent rehabilitation.

Introduction

Dundee Limb Fitting Centre (DLFC) was established by Professor George Murdoch in 1965 as a specialised unit for the management of the amputee in Tayside, Scotland. Over the years the theme of total patient care by a multi-professional clinical team has evolved. The centre is now an unique institution where the total service for the amputee is provided by the Tayside Health Board for the population of the region. Over the years, because of its reputation and quality of service, the centre has attracted referrals nationally.

All correspondence to be addressed to C. P. U. Stewart, Dundee Limb Fitting Centre, 133 Queen Street, Broughty Ferry, Dundee DD5 1AG, Scotland.

For the most, amputation is considered as the end of the road, but for the patient it is the beginning of a new life. Once a person becomes an amputee his or her life style changes dramatically. Accordingly, it behoves those concerned in the management of these patients to make life both comfortable and worth living. It must be emphasised that amputee management requires a holistic approach. The management process is complex and must cover all aspects provided by professional people from various disciplines.

Over the years it became apparent that numerous surgeons were performing amputations and that more above-knee (AK) amputations were carried out compared to below-knee (BK). In Dundee serious thought was given to improving this situation. Various important points were looked at and two significant factors were identified:

1) to concentrate the experience in the field of amputation surgery;
2) to provide accurate level selection to save as many knee joints as possible.

The following phases of patient management were identified:

1) vascular assessment and level selection;
2) pre-operative management;
3) surgical management;
4) post-operative management;
5) prosthetic management and rehabilitation.

In these areas experience was concentrated to gain more expertise to achieve a high quality service. This expertise was then integrated to provide total patient care. Once a patient is indentified by a clinician as a possible candidate for amputation, he or she is referred to the Tayside Amputation Service. The service from thereon takes over the management and provides the assessment, surgery, pre and post-operative care and rehabilitation.

Assessment of the patient is carried out in the vascular Laboratory at Ninewells Hospital and the surgery is performed in a specialised unit, where nursing and theatre facilities are available, by one of two teams of orthopaedic surgeons. Following, surgery the patient is transferred to the DLFC, usually within a week, for post-operative management, rehabilitation and prosthetic fitting.

Bi-weekly multi-disciplinary ward rounds are carried out with detailed case conferences with full discussion and appropriate goal setting. Close co-operation with family members is encouraged and interdisciplinary discussions are full and frank. Visiting social workers and district nurses ensure that discharge is smooth and stress-free. Clerical staff ensure that out-patient follow-up appointments occur regularly.

On average a BK amputee is discharged from the centre, with a definitive prosthesis, within 49 days of surgery. Patient contact and care is maintained through the Out-Patient Clinic for the rest of the patient's life.

Information relating to amputation surgery and rehabilitation has been collected over the years and is described below. The results show that with a careful, integrated, organised and methodical approach to the care of the amputee, high rates or BK amputation can be achieved with over 76% returning to their home, independent and mobile with a prosthesis.

Method

Some 1805 primary amputees have been studied from 1965 to 1989, the data on level of amputation, revision rate, prosthetic fitting, age, sex, pathology, discharge placement and date of death have been recorded over the years and now are stored on an Olivetti M24 PC with dbase III+ software.

In 1981 a comprehensive review of the stored data was undertaken and the new expanded data sheet implemented. Further, although not reported here, physiotherapy and prosthetic data have been routinely stored for analysis.

Analysis was carried out using dBase III+ and Reflex a database manager package.

Results

1) The average age at the time of first amputation of the amputee has remained remarkably stable at 69.1 years (range 66.6 yr–71.46 yr).
2) The sex ratio was 59.7% male, 40.3% female.
3) Figure 1 is a graph of AK to BK as a percentage of lower limb amputations. The change from a high rate of AK amputation in 1966 to a high rate of BK from 1972 to 1989 is clearly shown.
4) Table 1 shows the levels of amputation in Scotland and in the Tayside region in identified periods. The DLFC admitted 98% of all the amputations performed in

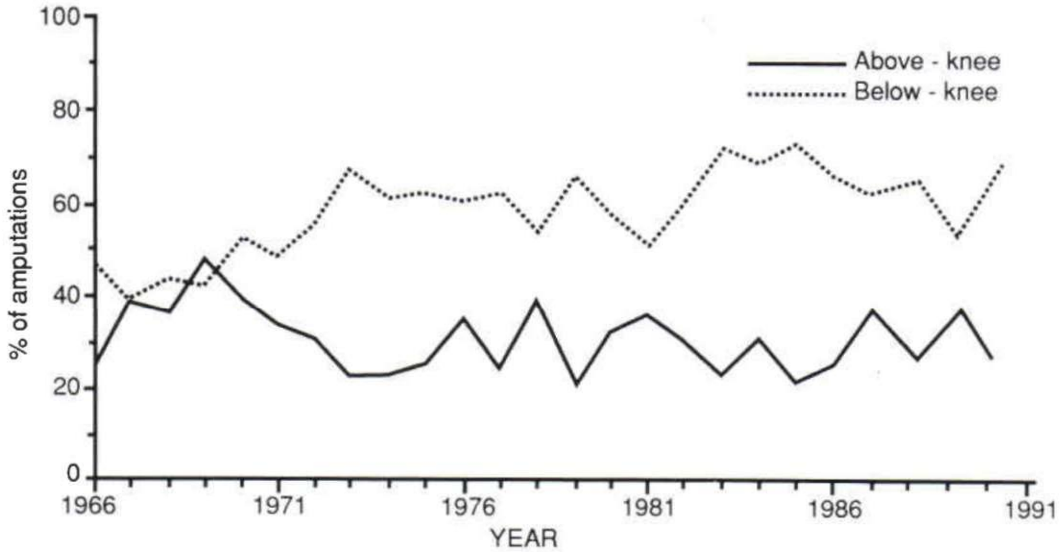


Fig. 1. Above-knee vs below-knee amputations (1966–1990).

Tayside but a few were discharged directly from the hospital where the amputation was carried out and were not admitted to the Centre.

5) Table 2 lists the overall causal condition of the amputation, the predominance of vascular cases is clear.
6) Table 3 lists the incidence of those amputees in the period 1981–89 who have had vascular surgery prior to the amputation. The overall incidence of those having vascular surgery was 33.8% and these patients were more likely to have an AK amputation.
7) Table 4 indicates that prosthetic fitting was achieved in 81% of all the cases with primary amputation. The supply profile is shown in this figure.
8) Table 5 shows the supply of wheelchairs for the period 1981–1989 during which 25% of all cases received a wheelchair. The 1989 figures show the steady increase in the numbers receiving a chair.
9) Table 6 lists the place of discharge in the period 1981–89. Over 76% managed to go either to their own home or to a residential home for the elderly.
10) Table 7 shows for the 25 year period the details of the double amputees who represented 18% of all the patients studied.

Discussion

This study reviews the amputees admitted to DLFC over 25 years and reports the results achieved by a comprehensive National Health Service Amputation Service. The most dramatic change which has occurred has been the complete reversal of the AK:BK ratio (Fig. 1). There were 26% AK amputations in 1989 and 71% BK which in this respect was one of the most satisfactory results of the 25 years (Table 1). In 1965–66 there were 46% AK amputations and only 25% BK, demonstrating the dramatic change in these ratios. This high BK rate is in keeping with that reported by Malone *et al.* (1979) who achieved a ratio of 3.3 BK to 1 AK.

These high rates of BK amputation achieved are due to the integration of expertise with comprehensive multi-disciplinary team approach in patient management. The amputation service starts with use of a fully equipped vascular laboratory providing advice on level selection as reported by McCollum *et al.* (1984).

Recent literature has reported an apparent improvement in the BK:AK ratios (Pohjolainen *et al.*, 1990). Reviewing 175 amputees in Finland he reported 35% AK and 53% BK. Fyfe (1990) in the United Kingdom reported similar figures of 38% AK and 54% BK. Both these papers however, report patients referred specifically for prosthetic fitting and do not truly represent the total amputation rate in their respective countries. Pohjolainen *et al.* (1989) reviewed 16 operation units in Finland and found an overall rate of 33.9% BK and 62.6% AK. This demonstrated a remarkable difference between the amputation rate from the surgical team and those referred for prosthetic fitting. A study from Scotland (Knight and Urquhart, 1989), again displays data collected from Limb Fitting Centres and reports 43% AK and 55% BK.

Table 1. Levels of amputations for Scotland, Tayside and DLFC.

Level	Scotland (1982–1983)	Tayside (all years)	DLFC (all years)	DLFC 1989
Through Hip/Hindquarter	43%	0.9% (16)	1.5%	
Above-knee		31.4% (579)	32.9%	26%
Through-knee		0.5% (10)	5.0%	
Below-knee	55%	62.7% (1158)	56.6%	71%
Symes		4.5% (83)	4.0%	
		100% (1846)	100% (1805)	

Notes

1. In addition to the above there were 1,003 toes amputated in this period.
2. At DLFC there was an 18% failure rate at first level of amputation, local surgery achieved healing in 12%, higher amputation was required for 6%.
3. DLFC figures include some patients from outwith Tayside. Mainly from N. Fife.

Table 2. The overall pathology of lower limb amputations.

Pathology	DLFC (1966–1990)	Scotland (1982–83)
Peripheral vascular disease	59.7%	65%
Diabetes mellitus	26.0%	22%
Trauma	5.3%	6%
Tumour	1%	3%
Others	8%	4%

Note
1. Others include vasculitis, varicose ulceration.

In keeping with published literature there was an overall predominence in the Dundee figures of peripheral vascular disease (85.7% with 59.7% having arthrosclerosis without diabetes and 26% with diabetes mellitus (Table 2). The Scottish figures (Knight and Urquhart, 1989) revealed 87% peripheral vascular disease with 65% arthrosclerosis and 22% diabetes related. Pohjolainen *et al.* (1990) reported 81.2% with vascular disease in those referred for prosthetic fitting. Pohjolainen *et al.* (1989) reported that 88.8% of those on whom an amputation had been performed had vasculatar disease with almost equal arthrosclerosis and diabetic levels of 46.4% and 42.4% respectively.

In a Swedish study by Renstrom (1981) the incidence of diabetes mellitus was found to be 49.5% (193 cases) with Fleurant and Alexander (1980) from the United States reporting 74% having diabetes mellitus.

Over the period, 1981–1989, it was recorded that 53.8% of amputee cases in Dundee had previously had a vascular surgical procedure before amputation (Table 3), (either sympathectomy, femoral popliteal by-pass, thrombectomy or endarterectomy). Men were statistically more likely to have had surgery prior to amputation ($p<0.01$). It was established that the vascular procedure did affect the overall outcome of amputation surgery in respect of level; 64.1% had a BK amputation if no surgery was performed, whereas only 60.3% achieved this level after vascular procedure. This was a significant difference ($0.01<p<0.05$). This reduction in BK level is in keeping with other studies. Sethia and Berry (1986) drew attention to the fact that failed vascular reconstruction might adversely affect the ultimate level of lower limb amputation. Larsson and Risberg (1988) found no such effect. However Falstie-Jensen and Christensen (1990) reviewed 83 lower limb amputations by way of a logistical regression analysis of 18 variables and found that amputation failure occurred where previous vascular surgery had been performed. In addition they also indicated that the older the patient the more likely was a stump failure leading to a higher level of amputation.

In the years of study there was an overall revision rate of 18%. Of these 6% required a revision to another level and the others only required local surgery for example a "Wedge Resection' (Haddon *et al.*, 1987).

An overall prosthetic fitting rate of 81% was achieved (Table 4). There was no significant difference in the fitting rate of AK or BK amputees. These proportions of fitting are in keeping with those of Moffat *et al.* (1981) who achieved 72.7% fitting and Cummings (1974) at 79.7%, but less than Malone *et al.* (1979 and 1981) who reported 100% fitting. Recent figures from Finland report only 26.9% of 577 patients being fitted with a prosthesis (Pohjolainen *et al.*, 1989). Knight and Urquhart (1989) reported that of those fitted in Scotland with a prosthesis only 62% actually used their limb all the time, whereas 9% never used it at all. Some evidence was found that in those with high levels of amputation less use of the limb was made. In the single BK amputee 87% wore

Table 3. Lower limb amputation surgery for vascular cases (1981–1989).

Level	Had vascular surgery		Had no vascular surgery	
Hindquarter	0.2%	(1)	0.3%	(1)
Through hip	1.8%	(7)	0.9%	(3)
Above-knee	34.5%	(133)	25.9%	(85)
Gritti-Stokes	0.6%	(2)	0.6%	(2)
Through-knee	1.6%	(6)	3.0%	(10)
Below-knee	60.3%	(233)	64.1%	(210)
Syme	1.0%	(4)	5.2%	(17)
	100%	(386)	100%	(328)

Notes
1. 54.1% (386/714) of all cases (1981–1989) had a vascular procedure.
2. Of this group it was found that those who had had vascular procedures were slightly more likely to have an above-knee amputation than those who had not had any vascular surgical intervention ($0.01<p<0.05$).

Table 4. Prosthetic supply over the 25 years.

	All	(1803)	AK	(588)	BK	(1017)
Supplied	75%	(1354)	68%	(400)	78%	(797)
Not supplied	12%	(218)	14%	(84)	11%	(113)
Wheelchair only	4%	(77)	8%	(45)	3%	(31)
Wheelchair and prosthesis	6%	(106)	8%	(46)	5%	(55)
Not recorded	3%	(48)	2%	(13)	2%	(21)

Note 81% fitted with prosthesis.

Table 5. The supply of wheelchairs (1981–1989).

		19891–89		1989	
AK	Total Number	226		32	
	Wheelchair only	42	85(38%)	5	14(44%)
	Wheelchair+prosthesis	43		9	
BK	Total Number	452		51	
	Wheelchair only	31	84(19%)	3	19(37%)
	Wheelchair+prosthesis	53		16	

Notes

1. Overall 25% received a wheelchair
   38% above-knee, 19% below-knee: $p<0.01$

For those *only* supplied with a wheelchair above-knees were more likely to receive a wheelchair than below-knee ($p<0.01$).

2. In 1989 40% received a wheelchair
   44% above-knee, 37% below-knee: Not significant

the limb at least half a day every day whereas in a single AK amputee only 70% used the limb for half a day.

In the years 1981–1989, during which time the provision of a wheelchair was recorded on the DLFC data base, it was found that 25% of cases were supplied with a wheelchair (Tables 4 and 5). However, it was interesting to note that although the level of supply was consistent until 1985 the request for wheelchairs for amputees had risen sharply to 40% in 1989. This is thought to be due to the increase in frailty of the patients despite the fact that the age at first amputation has remained remarkably consistent over the period of study at 69.1 years. This last figure of wheelchair supply is in keeping with the 48% reported by Knight and Urquhart (1989) for Scotland.

During the whole period of the study it was found that significantly more AK amputees received a wheelchair than BK amputees ($p<0.01$). By 1989 the last year of the study, this difference was not so significant.

The amputees are discharged from DLFC or average 49 days post-operatively with a definitive prosthesis, no temporary devices being used. The period between delivery of the first definitive device and the need for replacement was about 9 months. During the period 1981–1989, 76.2% were discharged home either to their own home or to a residential/nursing home (Table 6). Only 8.4% were placed in long term hospital care.

Bilateral lower limb amputees represented 18% of all amputees (Table 7). These again were mainly in the peripheral vascular disease group 94%. The second amputation at BK level

Table 6. Discharge placement (1981–1989).

Home	72.8% (527)	76.2%
Residential home	3.6% (26)	
Long term care	8.4% (61)	
Dead (in unit)	3.6% (26)	
Others	11.6% (84)	
Total	100% (724)	

Note
Others include acute hospital transfer, and other hospital for long term rehabilitation.

Table 7. Bilateral amputee levels.

Level	Number
BK – BK	48% (157)
BK – AK	19% (61)
AK – AK	20% (66)
Others	13% (43)
Total	100% (327)

Notes
1. 327 (18%) of all amputees have had a bilateral amputation.
2. 66% of bilateral amputees have prosthetic fitting.
3. 31% of bilateral amputees supplied with wheelchair.
4. 74% of bilateral amputees were discharged home or went to a residential home.
5. 94% of bilateral amputees had vascular disease.

was achieved in 61% (189 cases), resulting in an overall bilateral BK amputation rate (BK-BK), of 48% (157 cases). BK-AK bilaterals were found in 19% (61 cases), and 20% (66 cases) had bilateral AK level. Prosthetic fitting was achieved in 66% (215 cases), the majority being bilateral BK level (131 cases), 83% of the bilateral BK amputees. It was found that in those fitted with a BK and AK prosthesis it was immaterial which was the initial level of amputation. Overall only 31% (101 cases) were supplied with wheelchairs, although more recently all bilateral amputees have had a wheelchair supplied. This contrasts fairly markedly to the report by Van de Ven (1973) reviewing elderly lower limb amputees who indicated all (100%) were supplied with a wheelchair. Discharge to home or welfare acommodation was achieved in 74% of the double lower limb amputees.

Conclusion

The 25 years experience of lower limb primary amputees has been analysed using data collected over this period.

In the 25 years studied, the average age of the amputees at the time of first amputation has remained remarkably steady at 69.1 years in keeping with other studies. The predominence of peripheral vascular disease cases is in keeping with other Western studies. The sex ratio was 57.9% male to 40.3% female.

A comprehensive amputee service based on the total, integrated care of the amputee has resulted in a dramatic yet sustained change in the AK:BK amputation ratio. This has also resulted in high ratios of prosthetic fitting although 25% of the cases required a wheelchair.

It has also been shown that previous vascular surgery does adversely affect the final level of amputation. Those who had previous vascular surgery carried out, displayed a lower below-knee to above-knee ratio than those who had no vascular surgery. Final discharge was achieved in an average of 49 days, with a definitive prosthesis in 82% of cases, either to their own home or a residential home (77.5% of cases).

The results clearly show the value of an integrated amputation and limb fitting service with a high BK level of amputation and high prosthetic fitting rate. The high discharge rate to either the patient's home or to a residential home adds to the advantages of such a scheme with social integration of the majority of amputees back into the community.

In summary the main advantage of an integrated programme are:
1) a high ratio of below-knee to above-knee amputation;
2) a high level of prosthetic fitting;
3) a high level of social integration.

REFFRENCES

Cummings V (1974). Immediate rigid dressings for amputees. *NY State J Med* **74,** 980–983.

Falstie-Jensen N, Christensen RS (1990). A model for prediction of failure in amputation of the lower limb, *Dan Med Bull* **37,** 283–286.

Fleurant FW, Alexander J (1980). Below knee amputation and rehabilitation of amputees. *Surg Gynecol Obstet* **151,** 41–44.

Fyfe NCM (1990). An audit of amputation levels in patients referred for prosthetic rehabilitation. *Prosthet Orthot Int* **14,** 67–70.

Hadden W, Marks R, Murdoch G, Stewart C (1987). Wedge resection of amputation stumps: a valuable procedure *J Bone Joint Surg* **69B,** 306–308.

Knight P, Urquhart J (1989). Outcome of artificial lower limb fitting in Scotland: a survey of new patients referred to the limb fitting centre. Health Services Research Input, Information Statistics Division, community services agency — Edinburgh: Scottish Health Service.

Larsson PA, Risberg B (1988). —Amputations due to lower limb ischemia, *Acta Chir Scand* **154,** 267–270.

Malone JM, Moore W, Goldstone J, Malone S (1979). Therapeutic and economic imput of a modern amputation program. *Bull Prosthet Res* **10(32)** 7–17.

Malone JM, Moore W, Leal JM, Childers SJ (1989). Rehabilitation after lower extremity amputation, *Arch Surg* **116**, 93–98.

Malone JM, Leal JM, Moore WS, Henry RE, Daly MJ, Patton DD, Childers SJ (1987). The "gold standard" for amputation level selection: Xenon — 113 clearance. *J Surg Res* **30**, 449–455.

Moffat LEP, Rainey JB, Macpherson AIS (1987). Major lower limb amp. research. *J R Coll Surg Edinb* **26**, 50–53.

McCollum PT, Spence VA, Walker WF, Swanson AJG, Turner MS, Murdoch G (1984). Experience in the healing rate of lower limb amputations. *J R Coll Surg Edinb* **29**, 358–362.

Pohjolainen T, Alaranta H, Wikstrom J (1989). Primary survival and prosthetic fitting of lower limb amputees. *Prosthet Orthot Int* **13**, 63–69.

Pohjolainen T, Alaranta H, Karkkainen M (1990). Prosthetic use and functional and social outcome following lower limb amputation. *Prosthet Orthot Int* **14**, 75–79.

Renstrom P (1987). The below knee amputee: a clinical study and experimental investigation of circulation, muscle atrophy and strength of the amputated leg. *Goteborg, Sweden: University of Goteborg.*

Sethia KK, Berry AR, Morrison JD (1986). Changing pattern of lower limb amputation for vascular disease. *B J Surg* **73**, 701–703.

Van de ven CMC (1973). A pilot survey of elderly bilateral lower-limb amputees, *Physiotherapy* **59**, 316–320.

INSTITUTIONAL MEMBERS

We are pleased to announce that the following organisation was an Institutional Member of ISPO for the year 1992.

Mostafazan and Janbazan Foundation
Orthotics and Prosthetics Centre
PO Box 14155–6344
Teheran
IRAN

Prosthetics and Orthotics International, 1993, 17, 21–26

Movement of the tibial end in a PTB prosthesis socket: a sagittal X-ray study of the PTB prosthesis

M. LILJA, T. JOHANSSON and T. OBERG

Department of Biomechanics and Orthopaedic Technology, University College of Health Sciences, Jönköping, Sweden

Abstract

To investigate the movement of the tibial end in the sagittal plane in the PTB prosthetic socket during a gait cycle, 7 patients with a median age of 72 years were examined using X-ray technique. The gait cycle was reduced to four different static positions: heel contact, mid-stance, push-off and swing phase. The mean value of tibial movement in the socket in the anteroposterior direction was 2.2 cm, in proximodistal direction 2.8 cm, and the total sagittal movement during the whole gait cycle was 7.5 cm. The results indicate that one factor affecting the magnitude of the movement was the prestretching of soft tissues. All the patients who experienced a good prosthetic fitting had their soft tissues prestretched. The extreme dorsal and proximal positions of the tibial end during the gait cycle was in the swing phase position. The extreme distal position occurred somewhere between mid-stance and push-off. The extreme anterior position of the tibial end was seen during heel contact. This study has shown the magnitude of the movements in a PTB socket during a simulated gait cycle. The study has given hints on factors affecting prosthetic fitting, and further research within this field might provide indications of how to optimise socket shape to give maximal patient comfort.

Introduction

Pain in the amputation stump is a major problem for both the patient and the prosthetist (Mattson, 1980; Whylie, 1991; Troup, 1988). According to Persson and Liedberg (1983) 20% of all amputees have stump pain. Other frequent problems are secondary wound healing and adherent scars (Yaramenko and Andruhova, 1986; Persson and Liedberg, 1983). Yaramenko and Andruhova (1986) found a frequency of 11.5% and Persson and Liedberg (1983) 9% of secondary wound healing in trans-tibial amptees.

When the soft tissues adhere to the skeleton, movement between the tibial end and the prosthetic socket can be a factor causing problems. The PTB socket and the PTB suction socket allow movement of the tibial end during walking. For PTB prostheses Eriksson and Lempberg (1969) showed a movement in the proximodistal direction of 2.25 cm. For PTB suction prostheses the movement in the proximodistal direction was found to be 1.1 cm (Grevsten and Eriksson, 1975). In the authors' opinion there is a connection between pain, adherent scar and the movement of the tibial end in the socket. They also believe that the skeletal movements can be one explanation of stump pain. When the soft tissues adhere to the bone, bone movement will stretch the skin. This stretching can result in laceration and in the worst case, necrosis (Lilja and Johansson, 1992).

It is even more important to decrease the movements in the socket, as it is known that skin damage increases with time in patients with prostheses (Persson and Liedberg, 1983).

The aim of this study was to examine the tibial movements in a PTB prosthesis. In contrast to earlier studies, the simulation of the different phases of the gait cycle in the present

All correspondence to be addressed to Magnus Lilja, B.Sc. C.P.O., Department of Biomechanics and Orthopaedic Technology, University College of Health Sciences, P.O. Box 1038, S-551 11 Jönköping, Sweden.

study is more representative of all the tibial movement during a normal gait cycle with a PTB prosthesis. The investigation was limited to trans-tibial prostheses. In studies of tibial movement in PTB prostheses, X-ray is a well documented method (Newton *et al.*, 1988; Baumgartner *et al.*, 1980; Haslan *et al.*, 1983; Bao-Shan *et al.*, 1977).

Material and method

Seven unilateral trans-tibial amputees with a median age of 72 (61–79) years, were examined. All patients, 2 women and 5 men, had PTB prostheses. The main diagnoses were diabetes mellitus in 5 and arteriosclerosis in 2 of the cases. All amputations were performed with a long posterior flap. Three of the patients had used their prosthesis for less than 6 months and none had used it for more than 1 year. The patients were consecutively selected from a population of trans-tibially amputated patients at the orthopaedic clinic of the county hospital of Jönköping, Sweden. The averge stump length was 14.1 (10–20) cm. The average width over the condyles was 9.5 (9–12) cm. Four out of seven patients perceived a good prosthetic fit. Three of these four patients had the soft tissues prestretched. This prestretching was carried out by the patient himself when putting on the prosthesis. The three patients who did not prestretch the soft tissues perceived the prosthetic fitting as poor.

Experimental setup

The patient was examined in four different positions representing four phases in the gait cycle: heel contact, mid-stance push-off and finally swing phase. To simulate heel contact and push-off the floor was tilted at an angle of 15 degrees. The patient could stand vertically on either the toes or the heel on the tilted floor and then load with the whole body weight. The floor reaction force affected the knee joint with an extending or flexing moment exactly as in push-off and heel contact respectively during normal gait. In mid-stance the patient was standing on a plane floor.

Normally the floor reaction force is about 75% of the body weight in mid-stance, and about 120% of the body weight in push-off and heel contact (Cunningham, 1958), but in the authors' simulation the floor reaction force was equal to the body weight in all these positions. To simulate the swing phase the prosthesis was positioned at an angle of 45 degrees relative to the floor.

X-ray examination

All X-ray pictures were taken in a sagittal projection. To determine the scale a metal measuring device was used. The roentgenological examination was performed according to a standard procedure. To obtain high contrast radiographs, all prostheses were prepared with a thin steel wire, running along the inner surface of the external prosthetic socket. This procedure was undertaken to simplify the measurement of the tibial end positions.

Calculation of movements

To calculate the total movement of the tibial end, the four positions of the tibial end were marked in a co-ordinate system. The four positions were joined by straight lines, forming a polygon. The perimeter of this polygon is almost equivalent to the total movement in the sagittal plane during the gait cycle. This perimeter describes the minimum movement that the tibia can perform in the sagittal plane.

Results

Overview

In all cases a distinct and uniform motion of the tibia in the prosthetic socket was observed during simulated gait. All graphs of the tibial movement showed similar patterns of movement. During the swing phase large movements of the tibial end in a posterior and proximal direction were observed. Then the movement changed into an anterior and distal direction. The pattern of movements and the magnitude of the movements are shown in Figure 1.

Total movement in the sagittal plane

The average total movement in the sagittal plane was 7.5 (5.3-9.6) cm. The movements in the individual patients are shown in Figure 2. The largest movement occurred in the anterodistal direction between push-off and swing phase. This movement was seen in all but one patient, who showed the largest movement between swing phase and heel contact.

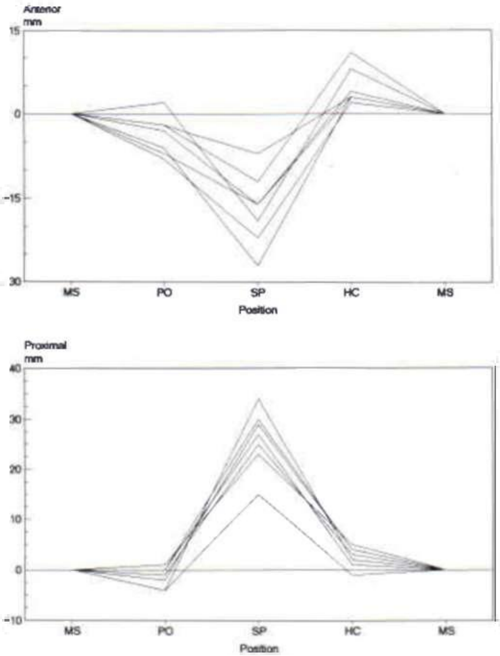


Figure 1. The pattern of movements and the magnitude of the movements of the tibial end in a PTB prosthesis socket in seven patients. HC=heel contact, MS=mid-stance, PO=push-off, SP=swing phase.
Top: Movements in the anteroposterior direction.
Bottom: Movements in the proximodistal direction.

Movement in anteroposterior direction

The average total horizontal movement of the tibial end was 2.2 (1.0-3.0) cm. The most posterior position of the tibia was seen during the swing phase and the most anterior position at heel contact. The largest movements of the tibia in the prosthetic socket occurred when the prosthesis was lifted up and swung forward to heel contact. The cumulative movement of tibia in anteroposterior direction during a whole gait cycle was 4.5 (2.1-6.0) cm.

Movement in proximodistal direction

The movement of the tibial end in the proximodistal direction was larger than the movements in the anteroposterior direction. The mean value was 2.8 (2.0-4.0) cm. The extreme proximal and distal positions vary between swing phase and push-off or swing phase and mid-stance. The mean value of the cumulative total movement in the proximodistal direction during a whole gait cycle was 5.7 (4.2-8.1) cm.

Discussion

Methodological considerations

The four positions were chosen for the following reasons: at heel contact and push-off the floor reaction forces are maximal at the same time as the lever arms to the knee are as long as possible and consequently the moments of force about the knee axis at a maximum. Mid-stance was chosen as a neutral position when the floor reaction force passes through the knee joint. The swing phase finally was chosen because it is the position when the weight of the prosthesis has the maximum effect on the stump at the same time as the triceps surae muscle is relaxed (Grevsten and Ståhlberg, 1975). It was attempted to simulate the direction of the floor reaction forces, especially in relation to the knee joint. The dynamic component of the floor reaction force could not be simulated in the static model. The heel contact and push-off phases in normal gait give a dynamic contribution to the floor reaction force which can reach about 120% of the body weight. Normally, at mid-stance, the floor reaction force is about 75% of the body weight (Cunningham, 1958; Lamoreux, 1985). In all positions the assumption was made that the floor reaction force was 100% of body weight. In spite of that, the model is regarded as acceptable as an approximation of slow gait with fairly moderate dynamic contributions to the floor reaction forces. Furthermore, the patient himself decided how much pressure he could tolerate by leaning the body backwards or forwards. The load level in this study was as close as possible to the tolerance limit of the patient, which means that the patients compensated for load differences.

Total movements

The movements of the tibia in a prosthetic socket have been studied in a few earlier studies (Eriksson and Lempberg, 1969; Grevsten and Eriksson, 1975). In this study the pattern of the tibial movement was uniform for all the subjects during the gait cycle. This motion pattern is regarded as valid for the majority of PTB prostheses with a soft socket. However, the magnitude of movement differed from one patient to another. Prestretching of the soft tissues seemed to be a factor affecting the magnitude of movement. In all cases except one, the largest movement was observed

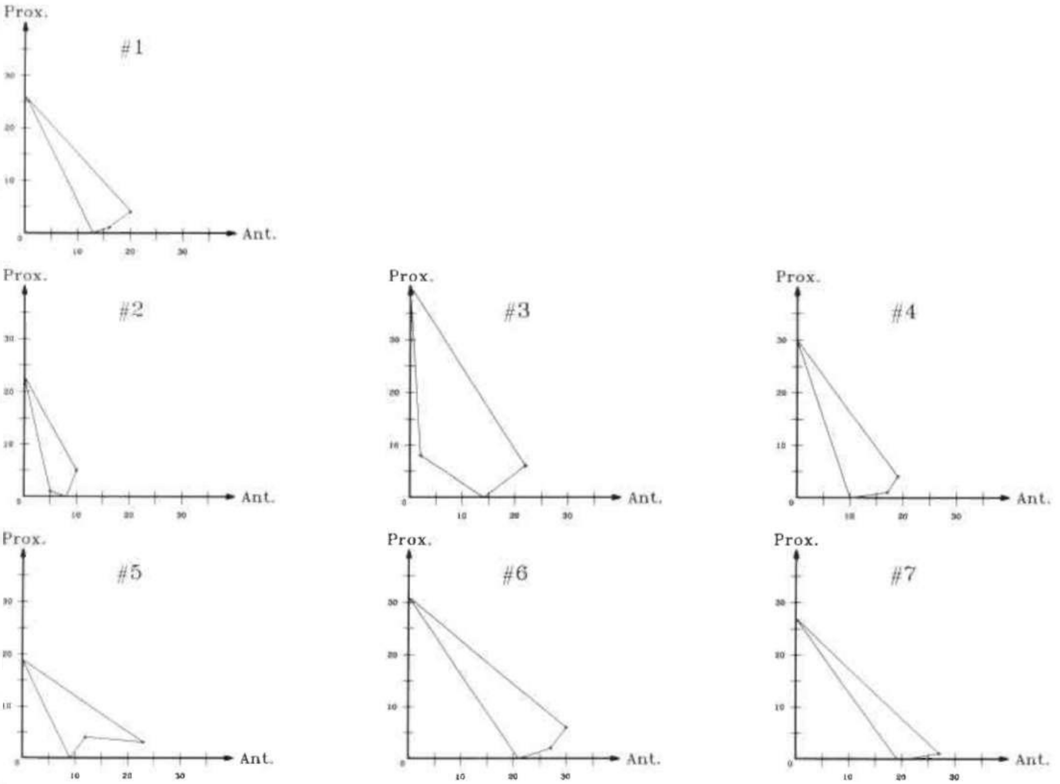


Figure 2. Movements of individual patients. # indicates case number. The movements in anteroposterior direction are indicated on the X-axis, those in proximodistal direction on the Y-axis.

between the swing phase and heel contact. In cases where the soft tissues were prestretched the total movement was reduced. Standing in the swing phase produces maximal moments of forces about the knee axis caused by the weight of the prosthesis. In this position the lever arm is as long as possible, the triceps surae muscle is relaxed (Grevsten and Ståhlberg, 1975), and the tibia is in a maximal dorsal and proximal position in the prosthetic socket. The most distal positioning of the tibial end occurs at some time between push-off and mid-stance. The most anterior position was, as expected, at heel contact.

Clinical implications

Two extreme cases can be theoretically considered: a) the prosthesis is anchored to the bone, and all energy and all movement is transferred to the prosthesis, and b) the movement of the tibia in the soft tissues is so large, that no energy and no movement is transferred. In reality there is a situation somewhere between these two extremes.

In the first case there is no energy loss. This is, of course, the ideal situation. In the second case all movement energy is lost in the soft tissues of the stump, i.e. the effort made by the patient will give no output as gait movements. This can be seen as the worst case of bad prosthetic fitting. The larger the movements of the bone and in the soft tissue, the larger the energy loss. The movement of the tibial end describes a polygonal figure (Fig. 2). The surface of this polygon will, in some respect, reflect the magnitude of the energy loss. The exact relationship between the movements of the tibial end and the energy loss cannot be easily described due to the non-linear viscoelastic properties of the soft tissues (Fung, 1987).

In an earlier study on PTB suction prostheses (Grevsten and Eriksson, 1975) the movements were twice as large in the proximodistal direction but similar in the anteroposterior direction as those found in this present study. It is believed that this can be explained by an inadequate suspension in the PTB suction prosthesis (Grevsten, 1978). The total amount of movement of the distal tibial end in the sagittal plane was as large as 7.5 cm. According to an earlier study of 22 patients provided with PTB prostheses (Öberg *et al.*, 1992) the median distance that the patients walked every day was 1,550 m. The stride-length was about 0.5 m. Thus the patients took about 3,100 steps per day. If one looks at the tibial movements inside the PTB socket it can be seen that the tibial end moves about 7.5 cm extra for each step. During one day the tibial end will move approximately 232.5 m extra inside the PTB prosthesis.

Lower limb amputees are generally old people with a mean age of 60–80 years in the vascular cases. They often have other diseases as well, and many of them have small residual capacity. Whatever the magnitude of the energy loss, it will influence the clinical outcome and performance of the patient. The body has a preference for the least energy consuming strategy, and energy loss may be the critical factor in deciding if the patient will be a walker or wheelchair case.

The movement of the tibial end in the soft tissues also greatly influences proprioception and kinesthesis. The position of the limb is mainly registered by proprioceptors in or nearby the joints. If the joint movement is not followed by a corresponding movement of the prosthesis, it is impossible to know the exact position of the foot, and consequently there will be an inaccurate gait pattern. In the ideal case all movement is transferred to the prosthesis, in the worst case no joint movement is transferred.

If numbers are considered the clinical results of the horizontal movement can be seen. One of the patients had a horizontal tibial end movement of 3.0 cm when he moves from swing phase to heel contact. The length of his lower limb was 42 cm from the knee down to the heel. The length of the amputated tibia was 14 cm. The difference of the actual position of the prosthetic foot and the position expected by proprioception and kinesthesis was 9 cm. Clinically, for every step the patient takes he has to produce a larger motion in the knee and the hip to compensate for the movements of the tibia inside the socket. This motion will lead to extra learning problems in prosthetic training although probably the patients will learn relatively quickly the new position of the prosthetic foot.

If the skin sticks to the walls of the prosthetic socket, all movement takes place in the soft tissue, and no friction energy is released in the contact between the skin and socket. If however there are adherences between the tibial end, the soft tissues and the skin, the movement of the tibial end are transferred directly to the skin, with friction and danger of friction wounds (Lilja and Johansson, 1992). Sometimes it can be observed that the outer skin layer of the stump is removed. This could be an effect of the friction between the stump and the socket. Thus even in this respect, movements of the tibial end in the socket can be clinically important.

At present knowledge concerning soft tissue mechanics of the amputation stump is scarce. The result of this study strongly indicates a need for further studies of stump and soft tissue mechanics to elucidate the relationship between bone movement, soft tissue deformation, energy loss and energy transferred to the prosthesis. It also indicates a need for studies where bone anchored prostheses are compared with conventional prostheses. Such studies can give hints on the magnitude of energy loss in the soft tissue.

Comparisons of tibial motion and socket types would result in further understanding of this problem. The results of such studies might lead to recommendations of what kind of prosthetic socket should be used when problems occur, (for instance adherent scars, oedema, hyperaesthesia, different diseases e.g. vascular or dermal diseases etc.) or to present problems.

Studies of tibial movement in prosthetic socket might explain some problems of pain in patients with adherent scars, and indicate how to optimize the form of the socket to give maximal comfort to the patient.

From the theoretical considerations above, it can be concluded that movements of the bony end in amputation stumps may have important clinical implications, and may also be a measure of good or bad prosthetic fitting.

REFERENCES

Bao-Shan J, Villanueva R, Dodd GD (1977). A new radiological technique in evaluation of prosthetic fitting. *Radiology* **122,** 534–535.

BAUMGARTNER R, LANGLOTZ M (1980). Amputee stump radiology. *Prosthet Orthot Int* **4**, 97–100.

CUNNINGHAM DM (1958). Components of floor reactions during walking. Berkeley: University of California. Prosthetic Devices Research Project 14.

ERIKSON U, LEMPBERG R (1969). Roentgenological study of movements of the amputation stump within the prosthesis socket in below knee amputees fitted with a PTB prosthesis. *Acta Orthop Scand* **40**, 520–529.

FUNG YC (1987). Mechanics of soft tissues. In: handbook of bioengineering./edited by Skalak R, Chien S. – New York: McGraw-Hill. 1.1–1.12.

GREVSTEN S (1978). Ideas on the suspension of the below-knee prosthesis. *Prosthet Orthot Int* **2**, 3–7.

GREVSTEN S, ERIKSON U (1975). A roentgenological study of the stump-socket contact and skeletal displacement in the PTB-Suction Prosthesis. *Upsala J Med Sci* **80**, 49–57.

GREVSTEN S, STÅHLBERG E (1975). Electromyographic studies of muscular activity in the amputation stump during walking with PTB- and PTB-suction prosthesis. *Upsala J Med Sci* **80**, 103–112.

HASLAM TP, WILSON MT, URLAUB MJ, BORRELL RM (1983). The use of X-rays and xerograms in prosthetics. *Orthot Prosthet* **37(1)**, 32–35.

LAMOREUX L (1985). Measurement and analysis of ground reaction forces. In: (Proceedings of the 1985 Uppsala Gait Analysis Meeting), Gait analysis in theory and practice./edited by Lanshammar H. – Uppsala University, Sweden: Uppsala University, Department of Technology, p47–69.

LILJA M, JOHANSSON T (1992). Adherent cicatrix after below knee amputation (Accepted for publication), *J Prosthet Orthot.*

MATTSON HS (1980). Fantombesvär och andra smärttillstånd. (Phantom pain and other pain conditions) In: Amputationskirurgi och proteser. (Amputation surgery and prostheses.) 2nd edition. edited by Hierton T. — Sweden, Falkoping: Gummessons tryckeri AB p 88-94. (In Swedish.)

NEWTON RL, MORGAN D, SCHREIBER MH (1988). Radiological evaluation of prosthetic fit in below knee amputees. *Skeletal Radiol* **17**, 276–280.

PERSSON BM, LIEDBERG E (1983). A clinical standard of stump measurement and classification in lower limb amputees. *Prosthet Orthot Int* **7**, 17–24.

TROUP IM (1988). Pre-operative and post-operative care: stump environment. In: Amputation surgery and lower limb prosthetics./edited by Murdoch G, Donovan R. — Oxford: Blackwell Scientific Publ. p 21–28.

WHYLIE B (1991). The psycho-social impact of amputation. In: Clinical aspects of lower extremity prosthetics./edited by Weber D. — Winnipeg: Elgan Enterprise. p 3–9.

YARAMENKO D, ANDRUHOVA RV (1986). Below-knee amputations in patients with vascular disease and prosthetic fitting problems. *Prosthet Orthot Int* **10**, 125–128.

ÖBERG T, LILJA M, JOHANSSON T, KARSZNIA A (1992). Klinisk utvärdering av proteshylsor. (Clinical evaluation of prosthetic sockets) Jonkoping, Sweden: University College of Health Sciences, Department of Biomechanics and Orthopaedic Technology. (In Swedish.)

Prosthetics and Orthotics International, 1993, 17, 27–37

Gait patterns of elderly men with trans-tibial amputations

E. D. LEMAIRE*, F. R. FISHER* and D. G. E. ROBERTSON**

*The Rehabilitation Centre, Ottawa, Ontario, Canada

**University of Ottawa, School of Human Kinetics, Ottawa, Ontario, Canada

Abstract

Gait patterns for the non-amputated leg of eight elderly men with trans-tibial amputations were assessed using kinematic and kinetic measures. Kinematically, the subject's walking speed was faster than expected but less than normative non-amputee data. The stride length was also less than non-amputee norms. Net joint moment and power analyses showed various discrepancies between the amputee subjects and non-amputees. The amputees required a concentric ankle dorsiflexor moment just after heel-strike to help move the lower leg into mid-stance position. The concentric plantarflexor moment at push-off was much larger than comparative data. A large eccentric flexor moment was also found at the hip during late mid-stance. Most of these discrepancies could be explained by the lack of an ankle moment generator on the amputated side of the body.

Introduction

With the "baby boom" generation entering middle age, health care systems will soon be expected to service a large senior citizen population. This may lead to an increase in the number of elderly amputees since non-traumatic loss of limb is most prevalent in the aged (Hunter and Waddell, 1976). Although a satisfactory level of clinical experience with seniors exists in the prosthetics field. Scientific research involving the gait of this group is lacking.

Most of the current gait research has been performed on young men with trans-tibial amputations. Of these studies, the majority involve kinematic stride evaluation on young to middle aged subjects (Breakey, 1976; Doane and Holt, 1983; Enoka *et al.*, 1982; Ganguli *et al.*, 1974; Gonzalez *et al.*, 1974; Hannah, *et al.*, 1984; Robinson *et al.*, 1977). These studies describe a population which exhibits an asymmetric gait pattern and walks slower than non-amputees.

Studies involving the kinetics of trans-tibial amputee gait have provided important information regarding amputee locomotion. Seliktar and Mizrahi (1986) used force plate analysis to develop a clinical technique providing quantitative measures for prosthesis alignment. It was suggested that the vertical impulse ratio (ratio between the vertical impulse for the prosthetic leg and the vertical impulse for the sound leg), antero-posterior impulse ratio, and perturbations on the antero-posterior force curve would be adequate for the assessment of prosthetic alignment; however, the effects of alignment changes on gait dynamics were found to be transient until the patient had reached a new steady state. The time required for the amputee's gait to stabilize was considered detrimental to the use of force plate analysis in a clinical setting.

Lewallen *et al.* (1986) used net joint moments to examine the load exerted on the joints of children with trans-tibial amputations. These children encountered larger ground reaction forces on their intact leg than non-amputees; however, joint moments at the knee and the hip were less than or equal to results for normal children (even though the ankle produced a greater dorsiflexor moment). The lower moment values for the knee and hip were attributed to a shorter stride length, a slower

All correspondence to be addressed to Edward Lemaire, Prosthetics and Orthotics Department, The Rehabilitation Centre, 505 Smyth Road, Ottawa, Ontario, Canada K1H 8M2.

walking speed, and an increase in double support and stance.

Another study which focused on the role of the non-amputated limb in trans-tibial amputee locomotion was performed by Hurley *et al.* (1990). The subjects from this study were all under 45 years of age and over half of the subjects were not long-term prosthesis users. Through examination of angle-angle diagrams for the ankle, knee, and hip joints, it was concluded that amputee gait patterns were more asymmetrical than non-amputees (since the joint angles for the sound leg were different from the joint angles for the prosthetic leg). Examination of horizontal joint reactions forces at the ankle and the knee showed that non-amputees had higher peak positive values than amputees. The lower amputee joint reaction forces were attributed to a lower push-off force from the amputated side and a slower walking speed. These results were consistent with the results from Lewallen *et al.* (1986).

Recently, trans-tibial amputee gait studies have focused on the contribution of energy storing feet to walking and running kinetics (Winter and Sienko, 1988; Czerniecki *et al.*, 1991; Torburn *et al.*, 1990; Barth *et al.*, 1992). Winter and Sienko examined gait patterns for 5 trans-tibial amputees who used a SACH (Single Axis Cushioned Heel) foot. Two subjects were re-tested with a uniaxial foot and one subject was tested with a Greissinger foot. All subjects demonstrated a greater than normal hip extensor moment from early stance to mid-stance to compensate for the below average push-off from the prosthetic leg. This hip extensor moment accounted for a quadricpes co-contraction over the same period, thereby compensating for the resulting knee flexor moment. All other moment and power patterns at the knee and hip were comparable with normal data. There was no difference between the ankle plantarflexor moment curves for the various prosthetic feet; however, the magnitude of these curves was approximately ⅔ of normal. The energy recovery for the uniaxial foot and Greissinger foot was found to be 20% and 30% respectively. The moment patterns at the knee and hip for the Greissinger fitting produced results which were closer to normal than those of the SACH and uniaxial foot.

Czerniecki *et al.* (1991) found results similar to Winter and Sienko during their investigation of running characteristics for five male trans-tibial amputees. The subjects ran along a 20m runway while using either a Flex-Foot, a SACH foot, or a Seattle foot. The SACH foot results were consistent with the findings of Winter and Sienko, although the amount of energy recovered was slightly higher (31%). The Flex-Foot trials produced a superior result in terms of approximating normal gait pattterns and energy recovery (84%) while the Seattle foot had a moderate effect on energy recovery (52%). The energy recovery capabilities of these prosthetic feet were found to be better than those of the traditional units; however, they were not comparable in respect of energy generation to normal plantarflexor activity (241% of the energy absorbed).

Torburn *et al.* (1990) used stride characteristics, joint kinematics, joint kinetics, electromyography (EMG), and physiological assessment to compare the Flex-Foot, STEN, Seattle, and Carbon Copy II (CCII) feet with the SACH foot. Kinematic test results showed that the CCII foot produced or permitted a significantly higher cadence and shorter gait cycle duration than the SACH and Flex-Foot ($p=0.02$). The relatively low statistical power, however, indicated that more subjects would be necessary to account for the majority of effects. Joint mechanical analysis demonstrated a larger ankle dorsiflexor moment at push-off from the Flex-Foot trials but no difference was found between any other prosthetic feet. The lack of a difference between the Seattle foot and SACH foot in terms of the dorsiflexor moment at push-off does not correspond with the results of Czerniecki *et al.* (1991). No differences were found for the EMG or physiological energy cost tests.

A study similar to Torburn *et al.* (1990) was performed by Barth *et al.* (1992). Gait kinematics, ground reaction forces, and physiological energy cost were used to assess the function of SACH, SAFE II, Seattle Lightfoot, Quantum, Carbon Copy II, and Flex-Walk feet. It should be noted that a treadmill was used for the energy cost protocol in this study. Since a treadmill can actively pull the support leg backward during gait, the gait pattern of these amputees could have been altered and the energy storing capabilities of these feet may not have been realised. These factors could account for the lack of difference

Table 1. Subject and prosthesis characteristics.

Subject	Age	Height (m)	Mass (kg)	Socket	Foot	Time Since Amputation (years)
1	69	1.80	88.6	PTB	SACH	47
2	67	1.79	80.5	PTB	SACH	46
3	71	1.73	86.4	PTB	SACH	46
4	68	1.71	80.0	PTS	SACH	43
5	66	1.81	82.3	PTS	Single axis	46
6	67	1.87	95.5	PTS	Multiaxis	46
7	70	1.75	77.3	PTS	Flex-Foot	46
8	72	1.72	78.9	PTS	Seattle	46

found between feet in the energy cost trials. The kinematic results showed a significant difference in linear velocity, cadence, stride length, and single limb stance time between the young, traumatic amputees and the older, vascular amputees ($p<0.1$). The Carbon Copy II and Quantum feet were shown to have significantly higher peak ground reaction force values at weight acceptance than in the other test cases ($p<0.1$). There was no significant difference between prosthetic feet for push-off forces.

Although these studies provided valuable information on trans-tibial amputee gait, the gait patterns of the elderly amputee have not received adequate attention. The majority of the present studies have also focused on the amputated side of the body. Due to the importance of the non-amputated limb for propulsion during locomotion (Seliktar and Mizrahi, 1986) examination of the kinematic and kinetic gait parameters for a group of elderly amputees is warranted.

Methods

Eight men with trans-tibial amputations who were over 65 years of age, had lost their leg due to trauma, and had worn a prosthesis for at least 25 years were recruited through the War Amputations of Canada. Before testing, all subjects were assessed by a prosthetist to ensure optimal fit and function of the prosthesis. The prosthetist also ensured that none of the subjects had stump problems (pain, swelling, pressure sores, etc.). The majority of the subjects had a socket with supracondylar suspension and used a SACH foot (Table 1). Anthropometric measurements were taken from the non-amputated side and body segment parameters were estimated using the relationships defined by Dempster (1955) and reported by Winter (1979).

The gait testing session involved placing joint markers on the shoulder, hip, knee, heel, ball of the foot, and toe of the subject's non-amputated side (Plagenhoef, 1971). Following a series of warm-up trials, the subjects walked at a natural cadence along a walkway while cinematographic and force plate data were collected at 50 Hz (Locam camera and Kistier force plate connected to a Data General mini computer). Six trials were recorded for each subject. The resulting data were corrected for perspective, filtered at 6 Hz, and used to calculate joint kinematics, net joint moments, and joint powers via the BIOMECH analysis package (Robertson and Winter, 1980; Lemaire and Robertson, 1989). The kinematic data were used to calculate stride length and walking speed.

The individual results were normalised to 100% of the stride time and to body mass before ensemble averages were calculated for each subject and for all subjects (grand ensemble). The ensemble averaged curves were

Table 2. Stride length, stride length/height ratio, and stride velocity averaged over individual trials.

Subject	Stride Length (m)	Stride Length / Height	Velocity (m/s)
1	1.21	0.67	0.95
2	1.41	0.79	0.97
3	1.37	0.79	1.27
4	1.34	0.75	1.13
5	1.57	0.86	1.46
6	1.48	0.79	1.16
7	1.47	0.84	1.33
8	1.43	0.84	1.32
ALL	**1.41**	**0.79**	**1.20**

used for all analyses and comparisons between the test results and normal data.

Results

Stride analysis

Table 2 displays the average stride length, stride length represented as a proportion of height, and walking velocity. The velocity values had a standard deviation of 0.18 and a range of 0.51 m/s.

The subjects who used energy storing feet and PTS suspension had higher walking velocities than all other subjects except for the subject with a single-axis foot and PTS suspension. It should be noted, though, that the faster walker was the youngest and the tallest subject in the study.

Joint moments and powers

Although a variety of prosthetic feet and suspension techniques were utilised by the subjects in this study, the gait patterns for the non-amputated limb were very similar. This similarity is reflected in the relatively low coefficient of variation (CV) values found for the grand ensemble average curves (Figs 1, 2 and 3).

A very consistent temporal relationship existed among trials and among subjects. The maximum range found between corresponding event codes (intersubject) was 0.08 s. This similarity permits inclusion of the ensemble averaged codes on Figures 1, 2, and 3.

The results from the comparison between existing data on non-amputees and the subjects from this study are listed in Tables 3, 4, and 5. The events referred to in the tables are marked on Figures 1, 2, and 3.

No substantial differences in the shape of the

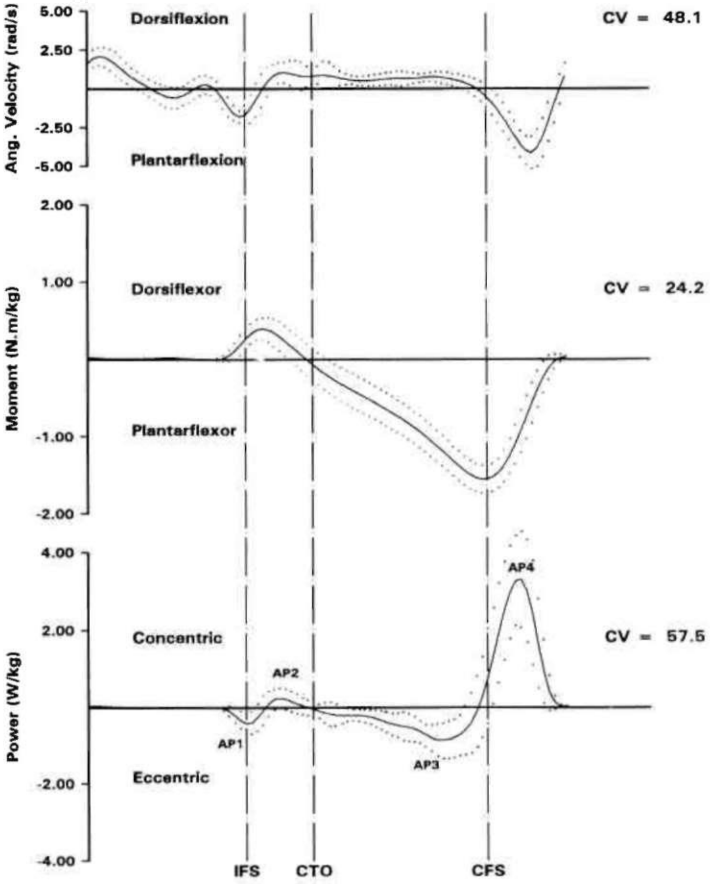


Fig. 1. Ensemble averages, standard deviations, and coefficients of variation for ankle angular velocity, moment, and power for all test subjects. The gait cycle started from non-amputated leg toe-off (IFS=ipsilateral foot-strike, CTO=contralateral toe-off, CFS=contralateral foot-strike).

moment and power curves were found during the between-subject evaluation. The magnitudes of the critical points along these curves were different; however, no relationship was found based on the type of suspension (PTB versus PTS) or the type of foot (SACH, multiaxis, energy storing). Generally, the slower walkers had lower moment and power values.

Discussion

Stride characteristics

It is generally assumed that the elderly walk more slowly and have a shorter stride length than the younger generation (Elble *et al.*, 1991; Murray *et al.*, 1969; Winter *et al.*, 1990), however, the amputees from this study had an average walking velocity and average stride length comparable to or above similar results from previous studies on younger amputees (Table 6). The results from Barth *et al.* (1992) involved three subjects who had an average age of 64.4 years and lost their leg as a result of vascular disease (average of five years since amputation). The extremely low results from this study may indicate a difference between elderly amputees who lost their leg due to trauma and people who had an amputation due to vascular disease. Another possible explanation is that a difference exists between long-term and short-term elderly prosthesis users.

The lack of a distinct age difference for walking speed or stride length may occur because the prime limitation for the amputee is the device and not physical capabilities which diminish with age. As technology provides more efficient means to store and/or generate energy on the prosthetic side, age related

Table 3. Gait characteristics at the ankle for all subjects (non-amputated limb). Comparisons are made with corresponding data for young non-amputees (†Winter, 1988) and elderly non-amputees (‡Winter, 1990).

Ankle				
	Event	Test Subjects	Young Non-amputees†	Elderly Non-amputees‡
Swing Phase	Concentric dorsiflexion (generation by dorsiflexors)	Small dorsiflexor moment to lift toe at initiation of swing phase (eliminate drop-foot).	Essentially no swing phase moment or power.	Essentially no swing phase moment or power.
	20% – 80% of swing phase	Essentially no ankle moment or power.	Essentially no swing phase moment or power.	Essentially no swing phase moment or power.
	Eccentric dorsiflexion (absorption by dorsiflexors)	Ankle resists plantarflexion as foot prepares for heel-strike.	Essentially no swing phase moment or power.	Essentially no swing phase moment or power.
Support Phase	Eccentric dorsiflexor (absorption by dorsiflexors)	Ankle resists plantarflexion to limit footslap at heel-strike.	Very small eccentric dorsiflexor moment.	Very small eccentric dorsiflexor moment.
	Concentric dorsiflexion (generation by dorsiflexors)	Ankle actively dorsiflexes until foot-flat to assist in moving the lower leg to mid-stance position and for stability.	Eccentric plantarflexor moment to control leg as it rotates over flat foot.	Very small concentric dorsiflexor moment.
	Eccentric plantarflexion (absorption by plantarflexors)	Ankle resists excessive dorsiflexion during mid-stance to control leg as it rotates over flat foot.	Same curves as normals but amputees have a larger peak power value.	Same curves as normals but amputees have a larger peak power value.
	Concentric plantarflexion (generation by plantarflexors)	Large plantarflexor moment and power during push-off.	Same curves as normals but amputees have a much larger peak power value.	Same curves as normals but amputees have a larger peak power value.
General				
– Amputee curves show much lower CV's than normals. This is contrary to expected results for amputees. – Essentially no ankle moment or power during the majority of swing phase. – Peak moments are similar to normals. – Peak powers larger in magnitude than normal during mid-stance and during push-off.				

differences in stride characteristics may become apparent. The tendency for the SACH foot users to have a slower walking speed than the other subjects may support this view, although the subjects who switched to energy storing prosthetic feet may have done so since they naturally walked faster or were in better physical condition. More research concerning energy storing prostheses and elderly amputee gait is required to determine the effectiveness of these devices on the senior population.

Gait

To determine the relevance of the results from amputee test trials, the test data were compared with the results from Winter (1988) and Winter *et al.* (1990). The earlier document by Winter provided gait results at three cadences, thereby facilitating comparison of relative values without accounting for walking speed. The 1990 document was used as a source for comparison since it contained moment and power results for elderly amputees (intra-subject data for a sample which averaged 68 years).

It is generally considered that the variability between amputee gait trials is greater than the variability between normal (Seliktar and Mizrahi, 1986). The CVs from this study contradicted this concept since the amputee trial variabilities were substantially lower than for the normal gait results. The low CV values may occur because the trans-tibial amputees were a more homogeneous group than the non-amputees and, since these elderly amputees were long-term prosthesis users, this group was

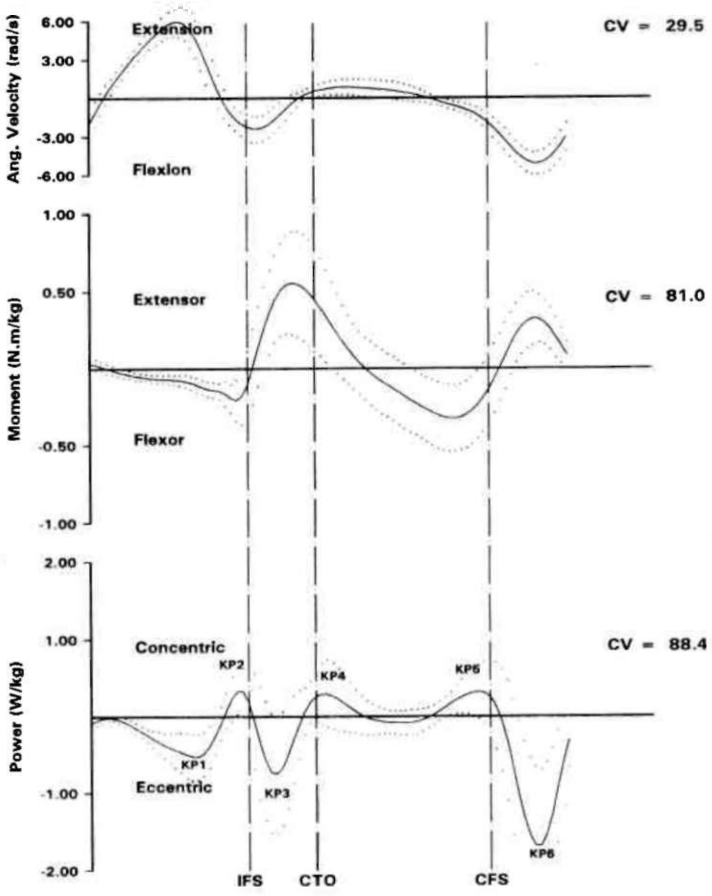


Fig. 2. Ensemble averages, standard deviations, and coefficients of variation for knee angular velocity, moment, and power for all test subjects. The gait cycle started from non-amputated leg toe-off (IFS=ipsilateral foot-strike, CTO=contralateral toe-off, CFS=contralateral foot-strike).

more homogeneous than the amputees who participated in previous studies.

Ankle

The shapes of the moment and power curves for the amputee subjects were essentially the same as those of results obtained from non-amputees (Winter, 1988), however, discrepancies were noted at the start of the stance phase. Normal subjects used the ankle dorsiflexors to absorb power at the beginning of stance while the amputee subjects had both power absorption (AP1) and power generation (AP2) over the same period. The concentric power burst may be required by the amputees to assist in "pulling" the lower leg segment through to mid-stance since the energy return capabilities of prosthetic legs will not produce the required moment at push-off. The lower push-off forces and moments from the prosthetic side have to be compensated by power generation from the dorsiflexors of the intact leg. The non-amputees may not require this concentric activity; therefore, they use the ankle dorsiflexors to control lower leg motion. The lack of two power bursts at the beginning of stance in the non-amputee data (Winter, 1988) may also be due to averaging of a group of non-homogeneous subjects, thereby masking small individual differences. In two studies using similar data collection and processing methods these two power events were present (Winter and Sienko, 1988; Winter *et al.*, 1990). These curves show an eccentric-concentric power pattern similar to the amputee subjects, although the magnitudes of these power bursts are much lower for the non-amputees. An eccentric dorsiflexor moment is initiated just before heel-strike for all amputee subjects but at heel-strike for elderly non-amputees. This may indicate a preparatory phase for the amputee group before weight is being transferred from the amputated side of the body to the non-amputated side.

The shapes of the moment and power curves during mid-stance and push-off were consistent

Table 4. Gait characteristics at the knee for all subjects (non-amputated limb). Comparisons are made with corresponding data for young non-amputees (†Winter, 1988) and elderly non-amputees (‡Winter, 1990).

Knee				
	Event	Description	Young Non-amputees†	Elderly Non-amputees‡
Swing Phase	Eccentric Flexor (absorption by flexors)	Knee flexors reduce the amount of knee extension during swing (prevent knee hyperextension).	Same curve shape, although the peak moment is higher for amputees.	Same curve shape, although the peak moment is higher for amputees.
	Concentric Flexor (generation by flexors)	Knee actively flexes to prepare for heel-strike (i.e., receiving load by *breaking the knee*).	Same as amputee subjects.	Same as amputee subjects.
Support Phase	Eccentric Extensor (absorption by extensors)	Knee controls flexion during mid-stance.	Same as amputee subjects.	Same as amputee subjects.
	Concentric Extensor (generation by extensors)	Knee actively extended to support bodyweight and raise centre of gravity at toe-off (amputated leg).	Same as amputee subjects.	Same as amputee subjects.
	Concentric Flexor (generation by flexors)	Knee actively flexes to prepare for push-off.	Same as amputee subjects.	Same as amputee subjects.
	Eccentric Extensor (absorption by extensors)	Knee extensors control knee flexion during push-off.	Similar curve but the amputee trials have a higher peak moment and power.	Similar curve but the amputee trials have a higher peak moment and power.

General

- Shape of moment and power curves comparable with Winter.
- Power curve has similar shape but is slightly offset temporally to the left (i.e. Winters KP3 curve occurs after the same curve for the amputees).
- Power and moment curves have higher magnitudes than normal during push-off.

for all available data but the moment and power values were higher for the amputee group. The larger power values for the AP3 power burst are used to control the amount of dorsiflexion before push-off. Controlled dorsiflexion is necessary to slow the forward progression of the tibia after mid-stance. This function is necessary to prepare for weight transfer to the prosthetic side.

The push-off power (AP4) was larger for the amputee group since the non-amputated leg must compensate for the lack of an energy generating segment on the amputated side. The amputee moment and power results were much larger than the elderly non-amputee results.

Knee

The moment and power curves for both groups were similar in shape but the peak values during swing and at push-off were higher for the amputees. The large eccentric flexor moment during swing may occur in response to a faster leg movement. Since the support time for the amputated side is less than the support time for the intact side the non-ampuated leg must complete the swing phase in less time in order to maintain a degree of symmetry. The resulting increase in speed would require a larger eccentric flexor moment to limit knee extension at the end of the swing phase.

The larger knee moment and power at push-off was in response to a larger concentric plantarflexor moment at the ankle and concentric flexor moment at the hip. Since peak powers at the hip and ankle were larger than for normals, a larger eccentric extensor moment was required to control the rate of knee flexion during push-off.

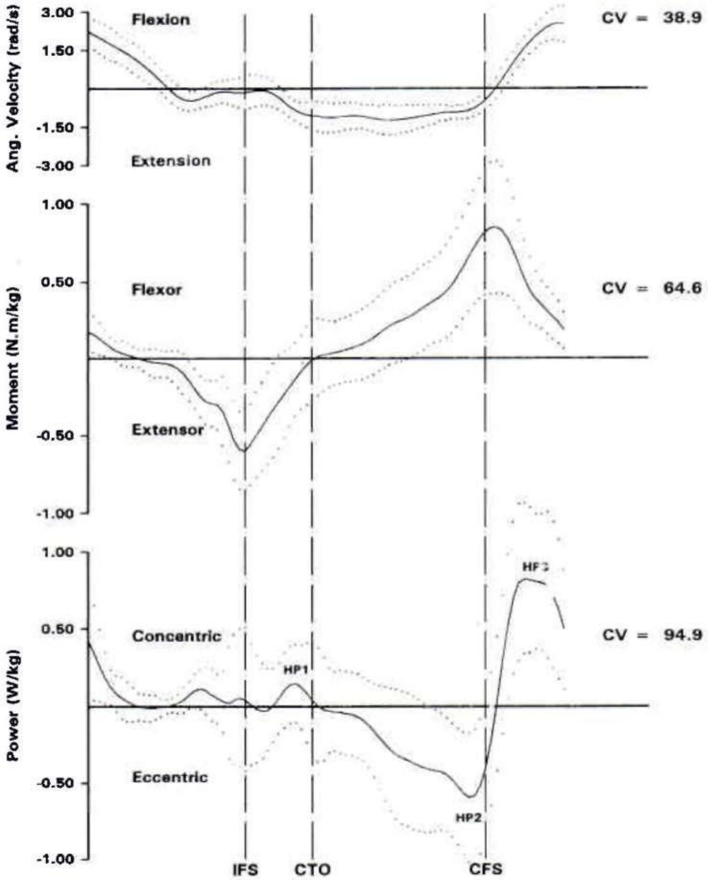


Fig. 3. Ensemble averages, standard deviations, and coefficients of variation for hip angular velocity, moment, and power for all test subjects. The gait cycle started from non-amputated leg toe-off (IFS=ipsilateral foot-strike, CTO=contralateral toe-off, CFS=contralateral foot-strike).

Hip

The shapes of the amputee hip moment and power curves were very similar to the reference curves for non-amputees; however, larger moment and power values were found during the support phase. A greater concentric flexor moment at push-off was necessary to initiate a faster swing phase (since the non-amputated leg has a shorter swing phase period than the amputated leg). The larger eccentric flexor moment just before push-off may be the result of the same mechanism as the eccentric extensor moment at the ankle (preparing for transfer of weight to the prosthetic limb). The hip flexors were required to control extension of the hip before active flexion was initiated at push-off.

Intersubject evaluation

The lack of intersubject difference between non-amputated leg gait patterns for PTB or PTS sockets was expected since a good fit with either design should provide a satisfactory interface for walking at a natural pace. The type of prosthetic foot did not appear to have an effect on the non-amputated leg; however, this study had an insufficient number of subjects to reach such a conclusion. Based on observations

Table 5. Gait characteristics at the hip for all subjects (non-amputated limb). Comparisons are made with corresponding data for young non-amputees (†Winter, 1988) and elderly non-amputees (‡Winter, 1990).

Hip				
	Event	Description	Young Non-amputees†	Elderly Non-amputees‡
Swing Phase	Concentric Flexor (generation by flexors)	Hip flexion at, and just after, toe-off to lift leg (allow foot to clear floor).	Same as amputee subjects.	Same as amputee subjects.
	Concentric Extensor (generation by extensors)	Hip actively extended during last half of swing.	Similar curve but the amputee trials have a slightly higher peak moment.	Same as amputee subjects.
Support Phase	Concentric Extensor (generation by extensors)	Hip extends at weight acceptance (pull leg into support position).	Same as amputee subjects.	Much longer concentric extensor moment than amputees (66% of stance phase).
	Eccentric Flexor (absorption by flexors)	Limit hip extension as thigh rotates backward after amputated leg push-off (prevent collapse at hip).	Similar curve from Winter but larger values for both moment and power.	Shorter duration than amputees.
	Concentric Flexor (generation by flexors)	Initiation of swing phase (moves leg upward and forward).	Similar curve but the amputees trials have a larger peak moment and power.	Same as amputee subjects.
General				
– The moment and power curve for both the amputee groups and the normal groups have the same shape. – Larger peak values were found for the amputees at the initiation of swing and during mid-support.				

Table 6. Comparison of average walking velocities and stride lengths.

Study	Subject	Subject Age (years)	Average Velocity (m/s)	Average Stride Length (m)
Doane and Holt (1983)	Amputee	55–67	1.22	
Gonalez *et al.* (1974)	Amputee	43–77	1.07	
Robinson *et al.* (1977)	Amputee	21–73	1.07	1.32
Torburn *et al.* (1990)	Amputee	39–57	1.17	1.40
Barth *et al.* (1992)	Amputee	36–67	0.75	1.10
Waters *et al.* (1988)	Non-amputee	60–80	1.19	1.27
This Study	**Amputee**	**66–72**	**1.20**	**1.41**

from the prosthetist, some elderly amputees may not deform the keel contained in energy storing feet to the same degree as younger amputees. The subject who used a Flex-Foot did not appear to make use of the energy return capabilities of this device during walking but used this foot because of its light weight. Additional research is required to determine if energy storing feet have an effect on the moments and powers for the non-amputated leg of elderly amputees.

Conclusion

Gait patterns from the non-amputated leg of experienced, elderly, men with trans-tibial amputations were shown to be comparable with data from non-amputees, although anomalies were found just after heel-strike and at push-off. All differences could be explained by the lack of an ankle moment generator on the amputated leg. The between-subject variability was much lower than expected, indicating a degree of homogeneity for long-term prosthesis users. Further research involving the elderly population is essential to quantify the benefits to them of modern prosthetic components and techniques.

Acknowledgements

The authors would like to acknowledge the clinical contribution of Sam Smith C.P. (c) and the administrative assistance of Guy Martel C.P.O. (c), M.H.Sc. towards completion of this project. The technical assistance of Mr. Wayne Albert is also acknowledged. This project was funded by the War Amputations of Canada.

REFERENCES

Barth DG, Schumacher L, Sienko-Thomas S (1992). Gait analysis and energy cost of below-knee amputees wearing six different prosthetic feet. *J Prosthet Orthot* **4**, 63–75.

Breakey J (1976). Gait of unilateral below-knee amputees. *Orthot Prosthet* **30(3)**, 17–24.

Czerniecki JM, Gitter A, Munro C (1991). Joint moment and muscle power output characteristics of below-knee amputees during running: the influence of energy storing prosthetic feet. *J Biomech* **24**, 63–75.

Dempster WT (1955). Space Requirements of the Seated Operator. Wright Patterson Air Force Base (WADC-TR.)

Doane NE, Holt LE (1983). A comparison of the SACH and single axis foot in the gait of unilateral below-knee amputees. *Prosthet Orthot Int* **7**, 33–36.

Elble RJ, Sienko-Thomas S, Higgins C, Colliver J (1991). Stride-dependent changes in gait of older people. *J Neurol* **238**, 1–5.

Enoka RM, Miller DI, Burgess EM (1982). Below-knee amputee running gait. *Am J Phys Med* **61**, 66–84.

Ganguli S, Mukherji P, Bose KS (1974). Gait evaluation of unilateral below-knee amputees fitted with patellar-tendon-bearing-prostheses. *J Indian Med Assoc* **63**, 256–259.

Gonzalez EG, Corcoran PJ, Reyes RL (1974). Energy expenditure in below-knee amputees: correlation with stump length. *Arch Phys Med Rehabil* **55**, 111–119.

Hannah RE, Morrison JB, Chapman AE (1984). Prosthetic alignment: effect on gait of persons with below-knee amputations. *Arch Phys Med Rehabil* **65**, 159–162.

Hunter G, Waddell J (1976). Management of the patient requiring leg amputation for peripheral vascular disease. *Can Med Assoc J* **115**, 634–638.

Hurley GRB, McKenney R, Robinson M, Zadravec M, Pierrynowski MR (1990). The role of the contralateral limb in below-knee amputee gait. *Prosthet Orthot Int* **14**, 33–42.

Lemaire ED, Robertson DGE (1989). Power in sprinting. *Track Field J* **35**, 13–17.

Lewallen R, Dyck G, Quanbury A, Ross K, Letts M (1986). Gait kinematics in below-knee child amputees: a force plate analysis. *J Pediatric Orthop* **6**, 291–298.

Murray MP, Kory RC, Clarkson BH (1969). Walking patterns in healthy old men. *J Gerontol* **24**, 169–178.

Plagenhoef S (1971). Patterns of Human Movement. – Englewood Cliffs: Prentice Hall. p11-14.

Robinson JL, Smidt GL, Arora JS (1977). Accelerographic, temporal, and distance gait factors in below-knee amputees. *Phys Ther* **57**, 898–904.

Robertson DGE, Winter DA (1980). Mechanical energy generation, absorption and transfer amongst segments during walking. *J Biomech* **13**, 845–854.

Seliktar R, Mizrahi J (1986). Some gait characteristics of below-knee amputees and their reflection on ground reaction forces. *Eng Med* **15**, 27–34.

Torburn L, Perry J, Ayyappa E, Shenfield SL (1990). Below-knee amputee gait with dynamic elastic response prosthetic feet: a pilot study. *J Rehabil Res Dev* **27**, 369–384.

Waters RL, Lunsford B, Perry J, Byrd R (1988). Energy-speed relationship of walking: standard tables. *J Orthop Res* **6,** 215–222.

Winter DA (1979). Biomechanics and human movement. – Toronto: John Wiley and Sons. p150–152.

Winter DA (1988). The biomechanics and motor control of human gait. – Waterloo; University of Waterloo Press.

Winter DA, Sienko SE (1988). Biomechanics of below-knee amputee gait. *J Biomech* **21,** 361–367.

Winter DA, Patla AE, Frank JS, Walt SE (1990). Biomechanical walking pattern changes in the fit and healthy elderly. *Phys Ther* **70,** 340–347.

Prosthetics and Orthotics International, 1993, 17, 38–48

Clinical measurement of normal and shear stresses on a trans-tibial stump: characteristics of wave-form shapes during walking

J. E. SANDERS,*/** C. H. DALY,*** and E. M. BURGESS*

*Prosthetics Research Study, Seattle, Washington, USA
**Center for Bioengineering, University of Washington, Seattle, Washington, USA
***Mechanical Engineering Department, University of Washington, Seattle, Washington, USA

Abstract
Stresses on the surface of a stump within a prosthetic socket during walking can potentially traumatise stump tissues. To gain insight into stresses and design parameters that affect them, normal and shear interface stresses were measured on three unilateral trans-tibial amputee subjects during walking trials. During stance phase repeated characteristics in wave-form shapes from different subjects were apparent. They included "loading delays", "high frequency events (HFE's)", "first peaks", "valleys", "second peaks", and "push-off". Characteristics did not necessarily occur at the same time from one step to the next but their timings matched well with events in shank force and moment data which were collected simultaneously. For "plantarflexion" and "dorsiflexion" alignment changes, the above wave-form characteristics were still pesent but their timings within the stance phase changed. The physical meaning and relevance of the characteristics to stump tissue mechanics are discussed.

Introduction

Two types of localised stresses are generated between a stump and a prosthetic socket during ambulation: (i) *normal stresses* are perpendicular to the interface, and (ii) *shear stresses* are in the plane of the interface. Normal and shear stresses are important because they can traumatise stump skin tissues. Excessive static normal stress has been shown to cause blood flow occlusion (Daly *et al.*, 1976) which can result in necrosis and ulceration (Levy, 1962). Trauma from dynamic shear is apparent as separation at the dermal-epidermal junction, followed by fluid deposition and blister formation (Stoughton, 1957; Hunter *et al.*, 1974).

It is the authors' hypothesis that it is not exclusively the magnitude of the stress that is important in causing tissue breakdown. Instead, it is proposed that it is a combination of parameters. For example magnitude, frequency, and loading in other directions simultaneously may possibly all contribute to breakdown. The basis for this hypothesis comes from tissue mechanics literature and clinical experience. Naylor (1955) found that it took less work (defined as $\int$ Force · Number of cycles) to induce friction blisters on the anterior tibial surface of normal subjects if a high number of cycles at low stress were applied compared to a low number of cycles applied at high stress. Lanir and Fung (1974) demonstrated that under an applied uniaxial strain, stress was higher if displacement in the perpendicular direction was restricted. From clinical experience on trans-tibial amputees, Radcliffe and Foort (1961) noted that: excessive pistoning of the stump in the socket resulted in an abrasion; excessive friction irritated existing bursae; adventitious bursae formed over bony prominences and around tendons; and under excessive load bursae become distended, tender, and infected.

The purpose of this paper is to report characteristics of interface stress wave-form shapes from unilateral trans-tibial amputee subjects ambulating with prosthetic limbs. Both normal and shear stresses are described. Effects of those characteristics on stump tissue mechanics are discussed. Results for altered alignment settings and differences between static and

All correspondence to be addressed to J. E. Sanders Ph.D., Center for Bioengineering, WD-12, University of Washington, Seattle, Washington 98195, USA.

Table 1. Physical descriptions of amputee subject stumps.

PARAMETER	SUBJECT #1	SUBJECT #2	SUBJECT #3
Dimensions: Length (from patellar tendon to distal end):	14.6cm	15.2cm	11.4cm
Antero-posterior diameter at patellar tendon:	7.5cm	7.6cm	8.6cm
Medio-lateral diameter at femoral condyles:	9.2cm	9.5cm	10.2cm
Unusual characteristics:	very bony stump; little soft tissue, especially distally; very dry skin antero-distally	prominent medial distal osteophyte	excessive superficial tissue; prominent peroneal nerve; 1.3cm deep clefts in suture line scars at antero-distal end
Scar locations:	single scar on lateral surface	antero-distal and lateral surfaces	three parallel axially-oriented scars on antero-lateral and antero-medial surfaces; also a single scar on the distal end
Adherent scar tissue sites:	none	antero-distal	distal
Common clinical problems:	ingrown hair antero-medially; folliculitus removed surgically 16 months prior to first test date; no reoccurrence; dry skin antero-distally	antero-distal tissue breakdown medio-distally near osteocyte; notes increased frequency of sores with pistoning	folliculitus at postero-medial trimline; lipoma was removed surgically 9 months prior to first test date; no reoccurrence

dynamic magnitudes are also reported and interpreted.

Methods

Subjects: Test subjects were three active male unilateral trans-tibial amputees between the ages of 23 and 46 who regularly used patellar-tendon-bearing sockets with sleeve suspensions. They were non-diabetic, had no diagnosed neurological problems, and had no history of peripheral vascular disease. In the period 6 months before the clinical studies and during the clinical studies, no major stump skin breakdown problems occurred. For all subjects the amputated leg was the right leg. Descriptions of subject stumps are given in Table 1.

Instrumented prosthesis: All prosthetic design, fabrication, and fitting was performed by certified prosthetists.

A "total contact" hard socket and Pelite® liner were designed for each subject using Seattle ShapeMaker© (Prosthetics Research Study, Seattle, Washington) software and a vacuum-forming socket fabrication system (Davies and Russell, 1979). Sockets and liners were designed to be slightly smaller than those normally worn by a subject since in the interface stress studies no socks or nylon sheaths were worn between the stump and socket. Several design iterations were usually required before a fit was determined appropriate.

Normal and shear interface stresses were measured during walking trials. Custom-designed transducers (Sanders and Daly, 1991) that measured stresses in three orthogonal directions were positioned in mounts bonded to the external socket surface. Transducers protruded through holes in the socket and Pelite liner so that their sensing surfaces were flush with the inside liner surface. Because the transducer surface was made of Pelite, no foreign material was introduced to the stump. Forces and moments in the prosthetic shank were measured simultaneously using instrumentation described elsewhere (Sanders, 1991).

Sites for interface stress monitoring were selected. Clinical locations of interest on the anterior, posterior, and lateral surfaces were used. The tibial crest, patellar tendon, and fibular head regions were avoided because their high curvatures would have caused improper function of the instrumentation. Medial site locations were not used because of interference by the contralateral limb. Brim locations were avoided since the mounts would have interfered with the function of latex sleeve suspension. Example transducer sites for a subject are shown in Figure 1a. The seven regions were the same for all subjects except Subject #3 where the antero-medial distal site was located further proximal because of scar tissue clefts in the skin. In Subject #1 the lateral site was not tested. Only four of the seven sites were monitored in a data collection session because of limitations of the data acquisition system. Sites that were not monitored were filled with dummy transducer plugs to ensure that the pressure difference across the socket wall was maintained. An instrumented socket is shown in Figure 2.

The socket was bonded to a wood block using standard methods and materials (polyester resin and cellulose filler). To complete the prosthesis a Berkeley adjustable leg with an instrumented shank (Sanders, 1991), a Seattle™ Lite Foot (Model and Instrument Development (M+IND), Seattle, Washington) and a latex sleeve suspension were attached. Subject #2 used an elastic waist belt suspension in addition to the latex sleeve. The total mass of each instrumented prosthesis was approximately 3.2kg. The mass of the subjects' normal prostheses, which were thermoplastic limbs with Seattle™ System components, was approximately 1.5kg.

Signal conditioning was performed by equipment housed within a backpack box carried by the subject. A heavily-shielded cable extended from the backpack box to the prosthesis while a second thin cable extended from the backpack box to a computer data storage facility. The cables did not restrict a subject's normal range of motion but they did add mass to the prosthesis. The mass of the backpack box and cable was approximately 3.2kg.

Using standard techniques, static and dynamic

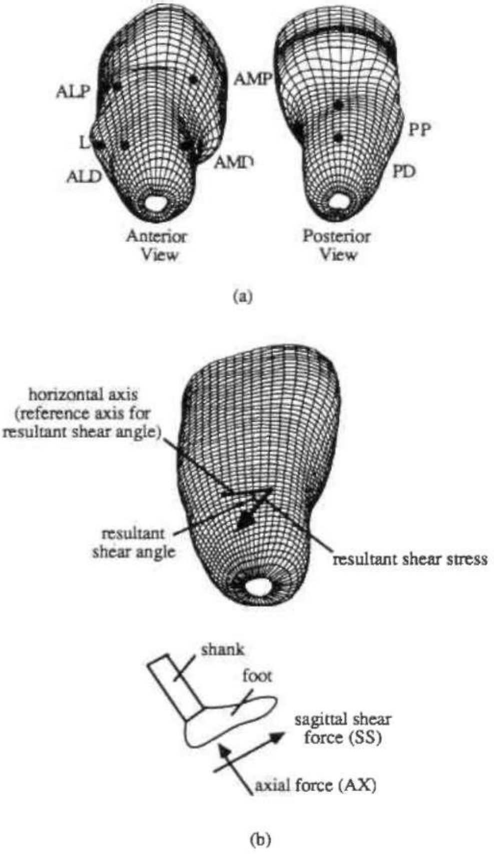


Fig. 1. (a) Transducer sites for Subject #2 are shown. ALP=antero-lateral proximal, AMP=antero-medial proximal, ALD=antero-lateral distal, AMD=antero-medial distal, L=lateral, PP=postero-proximal, PD=postero-distal. (b) "Resultant shear stress" is in the plane of the interface. Resultant shear angles are referenced to a horizontal axis and they increase in a counterclockwise direction. An approximately +45 degree resultant shear angle is shown. In the prosthetic shank, axial force is along the pylon axis and sagittal shear force is in the shank cross-section in a sagittal plane.

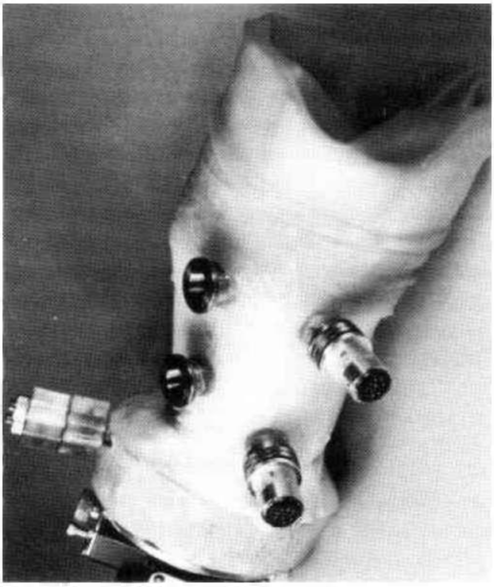


Fig. 2. Transducers occupy antero-medial sites. A strain-relief clamp for the cables (not shown) is affixed to the wood block at the lower-left.

prosthetic alignment were performed by certified prosthetists. The selected socket/shank alignment setting was defined as the "zero" or optimal alignment. "Zero" alignment was the same for all sessions conducted on a subject.

Testing protocols: Both walking and standing trials were conducted.

The pathway for walking was a 1.2m × 18m hallway that had a concrete floor covered with vinyl tiles. Data collection began ½ to 1 step after walking was initiated and lasted for 8 seconds. Usually 6 to 7 steps were collected in a trial though only the central 4 to 5 complete steps were used in analysis.

Spectral analysis of interface stress data collected at 500Hz sampling rate showed signal bandwidth to be less than 40Hz. In subsequent walking trials a 125Hz sampling rate with appropriate anti-aliasing filtering was used.

Trials were conducted with sagittal plane angular alignment at one of three settings: "zero", "plantarflexion", or "dorsiflexion". "Zero" was optimal alignment as determined by a team of prosthetists. "Plantarflexion" was an angular change of approximately 9 degrees with respect to an axis through the Berkeley jig and perpendicular to the sagittal plane. "Dorsiflexion" was an angular change of approximately 6 degrees. A group of at least four consecutive trials was conducted at each setting, thus the alignment was changed twice over the course of a data collection session. Sessions were approximately 30 minutes in length. The order of alignment setting was randomly selected.

Walking rate was controlled to a value between 94 and 99 steps/min using a metronome. The value selected for each subject was approximately his normal walking speed. Mid-trial stride length was measured during three of the trials for each subject using reference markers on the floor. Walking velocities for all subjects were approximately 1.3m/s.

Standing trials were conducted at the beginning and at the end of each session. During standing trials, a subject was asked to achieve one of four weight-bearing levels: (i) full weight-bearing, (ii) equal weight-bearing, (iii) low weight-bearing, or (iv) no weight-bearing. Prosthetic shank force data displayed on the computer monitor was used for feedback to the subject. Data was collected approximately 5 seconds after the subject had achieved a stable position. Results were not dependent on the order of weight-bearing tests.

The number of sessions conducted for Subjects #1, #2, and #3 were two, three, and four respectively.

Results

The following conventions are used in data presentation. Results are for "zero" alignment unless otherwise stated. By definition "normal stress" is perpendicular to the interface. "Resultant shear stress" is orthogonal to normal stress and is the resultant stress component in the plane of the interface (Fig. 1b). Compressive forces on the transducers are positive in sign. Resultant shear angles during the central portion of stance phase are in block parenthesis on the right side of interface stress plots. Stresses and angles are referenced to a transducer coordinate system as shown in Figure 1b.

Walking

All subjects loaded their prosthetic leg during approximately 62% of each step cycle. The average prosthetic stance phase duration for all subjects was 0.79s and the average step duration was 1.26s.

At a site, shapes of interface stress wave-forms changed from one step to another, probably as a result of compensation. However, repeated characteristics were evident in data from different steps. They are described in this section.

Interface stress curves during stance phase were divided into seven sections for analysis as shown in Figure 3. All sections were not necessarily present in all wave-forms and their timings were not necessarily as shown in Figure 3. This figure is intended only as a section-identification reference.

Loading delays: Stresses at transducer sites did not begin to increase immediately at heel contact. For Subjects #1 and #2 there was a delay of between 5% and 12% of stance phase

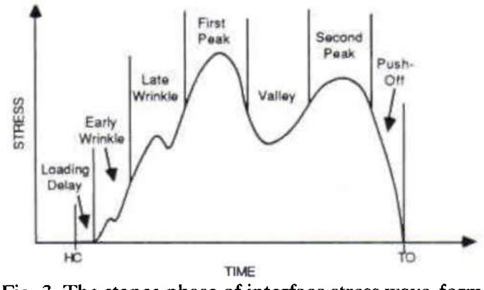


Fig. 3. The stance phase of interface stress wave-forms was divided into seven sections for analysis. HC=heel-contact, TO=toe off.

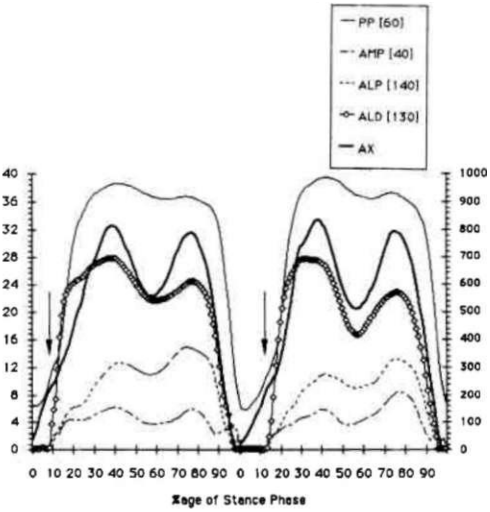


Fig. 4. Resultant shear stresses and shank axial force (AX) for stance phase of two consecutive steps for Subject #1 are shown. The swing phase has been removed from the record. Arrows indicate the ends of "loading delays" for antero-proximal sites. Numbers in block parentheses in the figure legend are resultant shear angles during the central phase of stance. The left scale is normal stress in kilopascals (kPa) and the right scale is shank shear force in Newtons (N). 1kPA=0.14lb/in^2, 1N=0.22lb. PP=postero-proximal, AMP=antero-medial proximal, ALP=antero-lateral proximal, ALD=antero-lateral distal.

(approximately 0.04s to 0.10s) while for Subject #3 the delay was approximately 4% (0.03s). As shown in Figure 4, for Subject #1, axial force in the prosthetic shank could reach as high as 25% of its maximal value when anterior site interface stresses began to increase. Usually "loading delays" were longer at anterior sites compared to posterior sites.

High freqency events: In most of the steps there were "high frequency events" (HFE's) in interface stress curves soon after heel contact. "Early HFE's" were high frequency spikes at about 7% to 12% into stance phase. "Late HFE's" were of lower frequency content and occurred later, usually partway up the rising part of interface stress curves.

Early HFE's: "Early HFE's" were apparent in both normal and shear stress wave-forms but they were not always present at all sites. If they occurred in a given step they happened at approximately the same time at different sites and their timings matched up well with HFE's in the shank wave-forms, particularly axial and sagittal shear forces (Fig. 5).

Late HFE's: "Late HFE's" were not as clearly defined as "early HFE's" in that they did not all occur simultaneously in a given step (Fig. 6). Nor were the times at which they occurred at a site consistent from one step to another. They could occur early and separate from the rest of the curve or they could occur late near the first peak and be virtually buried under that peak (Fig. 6b). However, they consistently occurred before the shank resultant force changed from being directed anteriorly to being directed posteriorly (Point "A" in Fig. 10a).

The ratio of resultant shear stress to normal stress was investigated to determine if there was a threshold value for the occurrence of "late HFE's". No consistent value was found at any site. Resultant shear angles showed no maxima, minima, or inflection points during "late HFE's".

First peaks: Peaks in interface stress wave-forms usually occurred near the maxima of the absolute values of axial force, shear force, and sagittal bending moment about the pylon centre in the prosthetic shank but they were not necessarily within a window surrounding those peaks (Fig. 7). Also, if distorted by the presence of "late HFE's", they could occur earlier. An example is the second step in Figure 6a at the postero-proximal site.

Timings of maxima at different sites in the same step were not simultaneous nor were the timings of maxima at the same site in different directions simultaneous. For example in Figure 7 antero-medial distal normal stress reaches a maximum

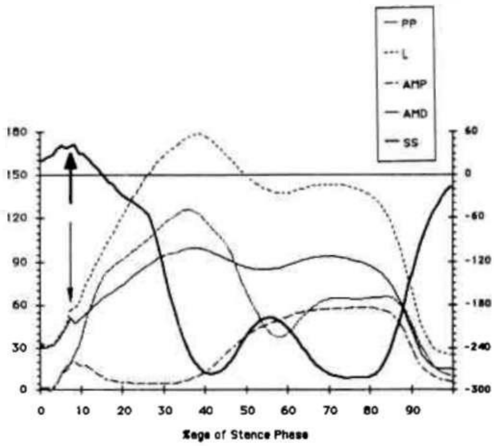


Fig. 5. Timings of "early HFE's" in normal stresses are indicated with a thin arrow for Subject #3. Timing of an "early wrinkle" in shank sagittal shear force (SS) is indicated with a thick arrow. The left scale is normal stress in kilopascals (kPa) and the right scale is shank shear force in Newtons (N). 1kPa=0.14lb/in^2, 1N=0.22lb.

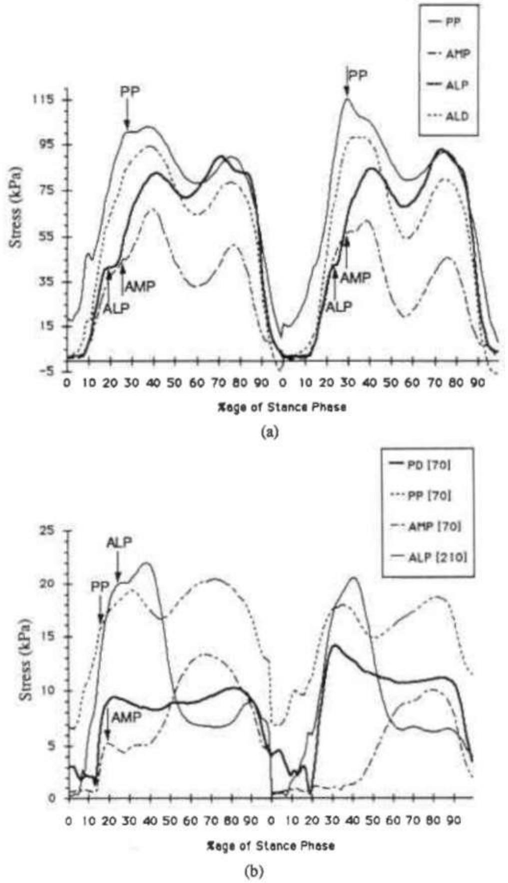


Fig. 6. (a) Normal stresses for stance phase of two consecutive steps for Subject # 1 are shown. Late HFE's are indicated by arrows. PP=postero-proximal, AMP=antero-medial proximal, ALP=antero-lateral proximal, ALD=antero-lateral distal. (b) Resultant shear stresses for stance phase of two consecutive steps for Subject #2. PD=postero-distal, PP=postero-proximal, AMP=antero-medial proximal, ALP=antero-lateral proximal.

slightly earlier than either postero-proximal normal stress or lateral normal stress. Antero-medial distal normal stress peaks earlier than antero-medial distal resultant shear stress.

Valley and second peaks: Valley and second peak regions were of lower frequency content compared with earlier in stance. An absolute value shear force/axial force ratio of approximately 0.2 to 0.4 corresponding to a resultant shear angle range of $-11°$ to $-21°$ was maintained throughout the valley and second peak regions (Fig. 10). The axial force on the bottom of the foot stayed at approximately the same sagittal plane position (Fig. 8).

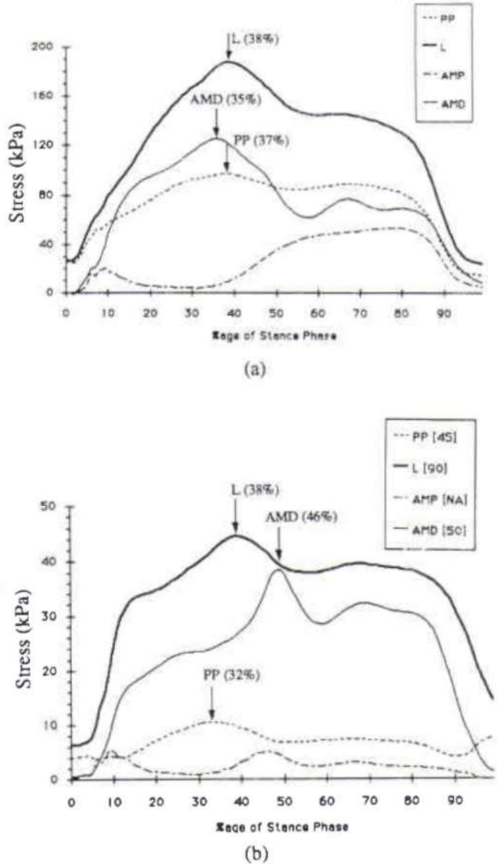
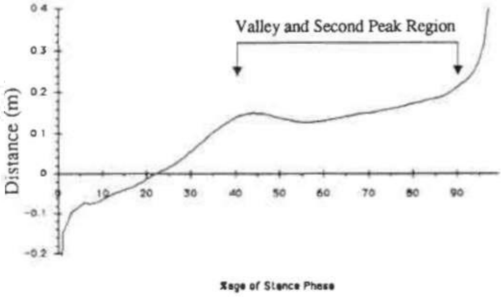


Fig. 7. (a) Normal stresses and (b) resultant shear stresses during stance phase for Subject #3. Data are from the same step. Shank axial force and sagittal shear force first peaks occur at 38% and 41% respectively. PP=postero-proximal, L=lateral, AMP=antero-medial proximal, AMD=antero-medial distal.

Push-off: Interface loading immediately before toe-off was not simply a reversal of that immediately after heel contact. For all subjects there were usually no "HFE's" or "loading delays" on the unloading phases of interface stress wave-forms. Ordering of loading did not necessarily match ordering of unloading. Unloading resultant shear vectors were not opposite in magnitude and direction from loading resultant shear vectors.

Different alignments

Changes in angular alignment did not significantly affect stance durations, step durations, or stance/step ratios. Thus the subjects still maintained their cadence well with the metronome despite the modifications.



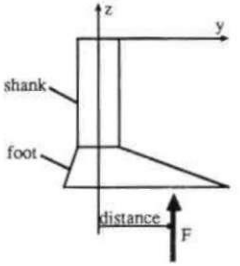


Fig. 8. Distance of the force "F" from the pylon axis is shown for Subject #3. "F" is parallel with the pylon axis.

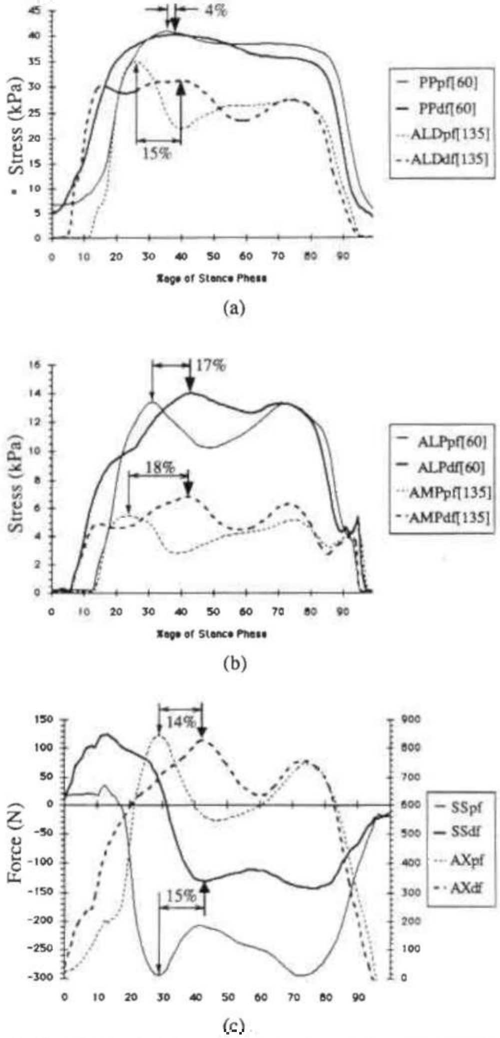


Fig. 9. (a) Resultant shear stresses for postero-proximal (PP) and antero-lateral distal (ALD) sites for "plantarflexion" (pf) and "dorsiflexion" (df) alignments. (b) Resultant shear stresses for antero-lateral proximal (ALP) and antero-medial proximal (AMP) sites. (c) Shank axial force (AX) and sagittal shear force (SS). All data are from the same step of Subject #1.

Interface stress "loading delays" and "early HFE's" occurred at approximately the same time as for "zero" alignment. "Late HFE's" still occurred before the shank resultant angle in the sagittal plane became negative.

The times at which peak stresses occurred did change. For Subject #1 and to a lesser degree for Subject #3, peaks in the first half of stance occurred earlier for the "plantarflexion" setting than for the "dorsiflexion" setting (Fig. 9). For Subject #2 the opposite trend was found. Peaks occurred later for "plantarflexion" than for "dorsiflexion". Resultant force directions in the sagittal plane remained approximately the same for different alignment settings (Fig. 10).

Mid-stance interface stress wave-forms, in general, shifted to the left for "plantarflexion" changes and shifted to the right for "dorsiflexion" changes. These shifts were usually not more than a few percent of stance phase.

Standing vs. walking

In walking trials, in general, antero-lateral distal and antero-medial distal sites were loaded during stance and unloaded during swing while postero-proximal, postero-distal, lateral, antero-lateral proximal, and antero-medial proximal sites were loaded during both stance and swing (Fig. 11). Standing trial data showed related patterns. Sites loaded during standing with equal or full weight-bearing were usually loaded during the stance phase of gait. Sites loaded during standing with minimal weight-bearing matched well with swing phase loaded sites. As shown in Figure 11 there were some exceptions to these trends. For example, Subject #3 unloaded the antero-lateral proximal site during stance but loaded it during swing. Subject #1 unloaded all anterior sites during swing.

Under equal weight-bearing on both feet or full

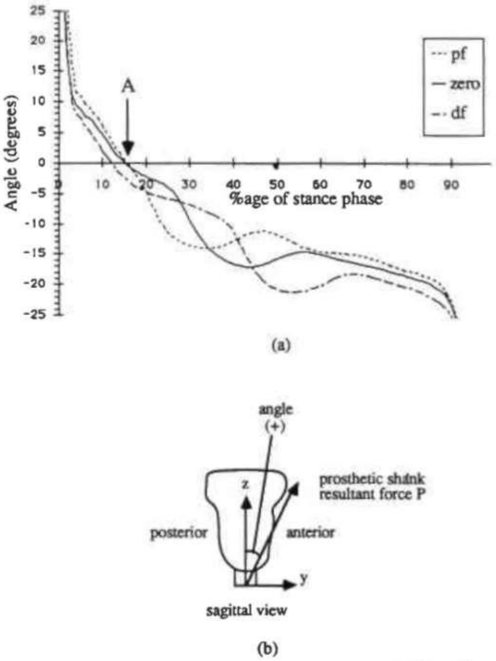


Fig. 10. Resultant force measured with the instrumented shank is shown for stance phase from steps at three alignment settings for Subject #3, pf="plantarflexion", df="dorsiflexion". At point "A" the resultant force for "zero" alignment changes from being directed anteriorly to being directed posteriorly.

weight-bearing on one leg, usually antero-distal sites were loaded more than antero-proximal sites. Ratios of resultant shear stress to normal stress usually stayed approximately the same or decreased from equal weight-bearing on both feet to full weight-bearing on one leg at all sites except at some antero-proximal locations.

Stance phase peak normal stresses averaged 2.4($\pm$0.6) times stresses achieved during standing with equal weight-bearing on both feet and 3.3($\pm$2.5) times stresses achieved during one-legged standing with full weight-bearing. Stance phase resultant stresses averaged 1.4($\pm$0.6) times stresses achieved during standing with equal weight-bearing on both feet and 2.1($\pm$2.0) times stresses achieved during one-legged standing with full weight-bearing.

Discussion

To the authors' knowledge this is the first research investigation of interface shear stress measurements on trans-tibial amputee subjects. It is also the first time both normal and shear stresses have been measured simultaneously.

In interpreting these data three characteristics of the instrumented prostheses which made them different from subjects' normal prostheses should be recognised. (i) Instrumented limbs were

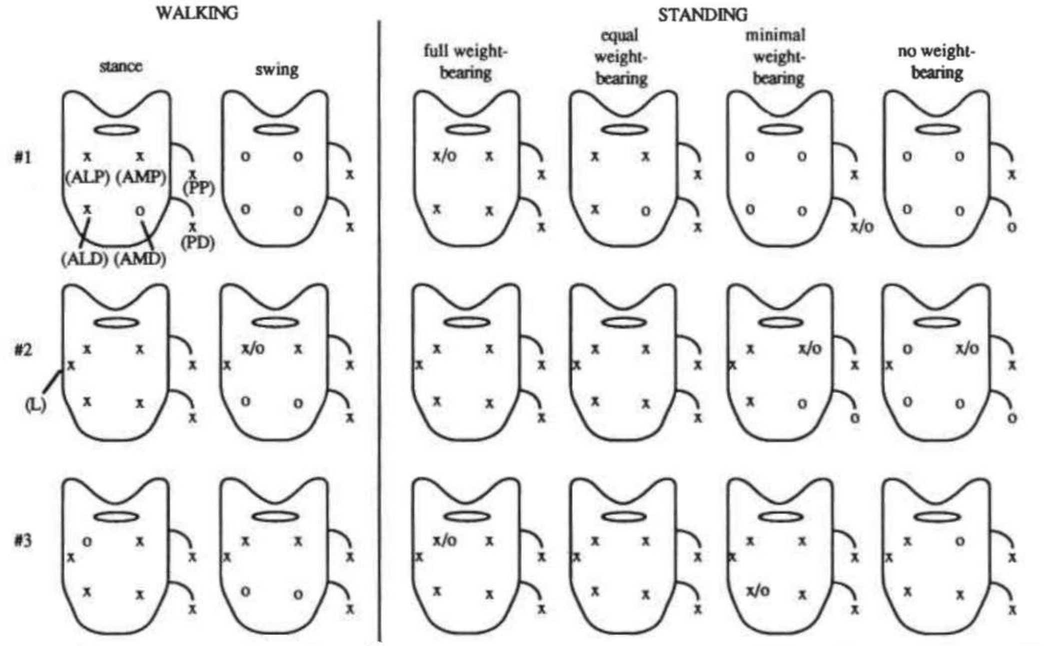


Fig. 11. Results for Subjects #1, #2, and #3 are shown. A site was considered loaded during stance or swing phase if it experienced greater than 5kPa stress during at least 50% of the phase. x=loaded, o=unloaded, x/o=loaded in some but not all of the sessions.

heavier than subjects' normal prostheses. (ii) No sock was worn. (iii) A 3.2kg backpack and cable apparatus was worn.

Loading delays: "Loading delays" were probably a result of one or both of the following sources: (i) "pistoning", i.e. slip between the stump and the socket. (ii) the anteriorly-directed resultant shank force (force "P" in Fig. 10b).

Clinical relevance of "loading delays" is that at the end of the delays the position of the stump in the socket is probably set for the rest of stance. If this position is different from that set under clinical fitting conditions, there may possibly be undesirable contour mismatches between the stump and prosthetic socket surfaces. Stress concentrations would be induced which could traumatise stump tissues.

Time lengths of "loading delays" may possibly also be important. If delay phases in longitudinal shear stresses end at unequal times at different sites, then intermediate skin tissue would be exposed to in-plane tension in a longitudinal direction. High in-plane tension can cause skin blanching and obstruction of blood flow (Kenedi *et al.*, 1965). Tension can also change the mechanical response of skin in the perpendicular direction, i.e. the transverse shear direction. Biaxial mechanical testing of rabbit skin has shown that high perpendicular stretch ratios cause the stress-strain curve to shift towards the origin (Lanir and Fung, 1974). This means that the stress level for blanching would happen at a lower uniaxial strain and at a lower internal energy change than if perpendicular strains were zero. Presumably the potential for tissue trauma would also be increased. As shown in this research, in stance phase immediately subsequent to "loading delays", transverse shear was usually greater than longitudinal shear on the anterior limb surface, thus it is possible that biaxial loading was induced.

Early HFE's: "Early HFE's" could have been due to several sources: (i) the foot being forced to foot-flat, (ii) deformation of the foot, (iii) activity of the other leg, (iv) muscle activation, or (v) slip at the interface. Because interface stress "early HFE's" matched well with "early HFE's" in the prosthetic shank, (i) or a combination of (i), (ii), (iii), and (iv) is probable.

If due to the foot being forced to foot-flat, deformation of the foot, or activity in the other leg, then "early HFE's" may possibly affect an amputee's sense of stability but they probably have minimal effect on tissue mechanics. Unless the skin is highly sensitive, tissue damage due to high-frequency low-intensity loading is unlikely (Naylor, 1955). In mechanical testing, skin response has been shown to be virtually insensitive to strain rate in the range of interest here (Lanir and Fung, 1974).

However, if "early HFE's" were due to muscle activation, then they may be providing important information early in stance. Muscle activation would be expected to change stump muscle geometry and stiffness. Thus "early HFE's" could serve as a signal for this change which could be important in subsequent interface stress generation. Knowledge of changes in muscle geometry and material properties could also be important when attempting to model mechanically the stump (Steege *et al.*, 1987; Krouskop *et al.*, 1987; Quesada and Skinner, 1991; Sanders, 1991; Torres-Moreno *et al.*, 1992) for the purpose of predicting interface stresses for use in socket design.

Late HFE's: It is possible that the principal source of "late HFE's" was the change in direction of shank sagittal shear force from being directed anteriorly to being directed posteriorly (Fig. 10). The bending moment on the tibia reversed direction and, because the bone was imbedded in soft nonlinear tissue, "HFE's" appeared in interface stress curves.

Clinical relevance of "late HFE's" is that, at these load levels, a quick interface stress reduction followed by a quick stress increase may possibly induce tissue damage. Friction blister studies (Naylor, 1955) have shown that blister occurrence is related to the frequency of the applied load cycles. Thus interface stress waveforms with "late HFE's" may possibly be more detrimental than those without "late HFE's". Also, because "late HFE's" did not all occur simultaneously in a step, it is possible that stress concentrations were induced in soft tissues.

First peaks: First peaks in interface stress waveforms did not usually occur simultaneously with each other nor did they necessarily occur simultaneously with shank force and moment maxima. This may, in part, be due to the nonlinear viscoelastic nature of stump tissues.

The lack of simultaneously-occurring peak interface stresses is significant because the tissue mechanics of the stump is affected. Peaks not occurring simultaneously in resultant shear stresses at different sites means that tensile stresses may possibly occur between loaded sites. At regions where skin is adherent to underlying bone, concentrated shear stresses would occur adjacent to or within adherent scar tissue. This could be particularly detrimental on the anterior surface where skin over bone is thin.

Valleys and second peaks: Mid- and late-stance interface stress curves lacked the characteristic features of early stance phase.

Push-off: There may possibly be important tissue mechanics manifestations of differences between heel-contact and push-off. Because wave-forms were not symmetrical, skin experienced resultant shear stress in principally one direction. Fluids were pushed unidirectionally, which changes their distribution beneath the skin surface. For example, at the antero-distal end, fluid was possibly pushed proximally and away from the midline of the tibia but not in the reverse directions. This may possibly explain why dry skin is sometimes seen at antero-distal ends of stumps. Subject #1 had particularly dry skin in the antero-distal area (Table 1).

Different alignments

Changes in interface stress wave-form shapes for "plantarflexion" and "dorsiflexion" alignment modifications reflect compensation. Subjects changed their gait to achieve stability and comfort. Wave-form shape changes were not consistent across transducer sites or across subjects, however, thus each subject compensated his gait style differently. However there were two parameters that consistently did not change for all subjects. They were: (i) ranges of shank resultant force directions in the sagittal plane (Fig. 10) and (ii) interface stress peak magnitudes (Sanders, 1991).

All subjects in this research were well-conditioned amputees. Possibly to maintain their usual sense of stability and comfort, amputee subjects kept ranges of shank resultant force directions and interface stress peak magnitudes consistent. New amputees however might not have that conditioning. Thus consistent angles and interface stress peaks would possibly not be achieved when alignment modifications are made during first-time fittings.

Results presented here are consistent with work by Pearson *et al.* (1973). Interface pressures at the patellar tendon, the lateral tibial condyle, and the medial tibial condyle stayed approximately the same for 10 degrees of flexion or extension alignment modification. Translational antero-posterior alignment changes were shown to cause changes in interface pressure wave-form shapes.

Standing vs. walking

Results showing loss of contact for swing phase and for low or minimal weight-bearing conditions are expected. It is likely that "pistoning" occurred at the interface. Exceptions such as the antero-lateral proximal site on Subject #3 and the antero-medial distal site on Subject #1 show that despite the intended socket design, sockets were not "total contact".

Ratios of "equal weight-bearing (symmetrical standing) stress: maximal stance phase stress" showed lower standard deviations than ratios of "full weight-bearing (one-leg standing) stress: maximal stance phase stress". Thus equal weight-bearing was a more accurate reflection of dynamic stress distribution during peak loading than was full weight-bearing. However, it should be noted that stress distribution differences between equal weight-bearing and dynamic loading were still rather substantial thus it would not be appropriate to conclude that there was a single ratio relating them. Standard deviations/averages for ratios of "equal weight-bearing stress: maximal stance phase stress" were 25% and 43% for normal and resultant shear stress respectively. It is likely that the high sagittal plane bending moment and the more posteriorly-directed shank resultant force vector achieved during gait compared to those achieved during standing were responsible for the differences.

Further research

Further research will be in two directions. First, interface stress measurement instrumentation will be modified so that the weight of the instrumented prosthesis is reduced, the backpack box is eliminated, a stump sock or sheath can be worn, and more transducers can be monitored simultaneously. The instrumentation will be used to investigate other parameters including: alignment changes in other directions, effects of socket shape modifications, and different prosthetic feet. A principal goal is to understand interface stress sensitivity to changes in prosthetic design. Another goal is to understand amputee adaptation to prosthetic modifications. In the long-term, knowledge of sensitivity and adaptation will be integrated into computer-aided prosthetic design.

A second direction of research is to investigate skin response to normal and shear loads. The intent is to better understand how skin adapts to mechanical stress to become durable and load-tolerant. Knowledge gained will be applied to design of rehabilitation treatment strategies that encourage tissue adaptation and minimise risk of tissue breakdown.

Acknowledgements

Support from the Whitaker Foundation, the Veterans Administration Rehabilitation Research and Development Section, and the Graduate School, Department of Rehabilitation Medicine, and Center for Bioengineering at the University of Washington is gratefully acknowledged. Assistance from Prosthetics Research Study prosthetist David Boone and University of Washington machinist Eric Strakbein is appreciated.

REFERENCES

Daly CH, Chimoskey JE, Holloway GA, Kennedy D (1976). Effect of pressure loading on blood flow rate in human skin. In: Bedsore biomechanics./edited by Kenedi RM, Cowden JM. — London: Mcmillan Press Ltd. p 69–78.

Davies RM, Russell D (1979). Vacuum formed thermoplastic sockets for prostheses. In: Disability./ edited by Kenedi RM, Paul JP, Hughes J. — London: Macmillan Press Ltd. p 385–390.

Hunter JAA, McVittie E, Comaish JS (1974). Light and electron microscopic studies of physical injury to the skin: II. friction. *Br J Dermatol* **90**, 491–499.

Kenedi RM, Gibson T, Daly CH (1965). Bioengineering studies of human skin II. In: Biomechanics and related bio-engineering topics./ edited by Kenedi RM. — Oxford: Pergamon Press. p 147–158.

Krouskop TA, Muilenberg AL, Dougherty DR, Winningham DJ (1987). Computer-aided design of a prosthetic socket for an above-knee amputee. *J Rehabil Res Dev* **24(2)**, 31–38.

Lanir Y, Fung YC (1974). Two-dimensional mechanical properties of rabbit skin — II. Experimental results. *J Biomech* **7**, 171–182.

Levy SW, Allende MF, Barnes GH (1962). Skin problems of the leg amputee. *Arch Dermatol* **85**, 65–81.

Naylor PFD (1955). Experimental friction blisters. *Br J Dermatol* **67**, 327–344.

Pearson JR, Holmgren G, March L, Oberg K (1973). Pressures in clinical regions of the below-knee patellar-tendon bearing prosthesis. *Bull Prosthet Res* **10(19)**, 52-76.

Quesada P, Skinner HB (1991). Analysis of a below-knee patellar tendon-bearing prosthesis: a finite element study. *J Rehabil Res Dev* **28(3)**, 1–12.

Radcliffe CW, Foort J (1961). The patellar-tendon-bearing below-knee prosthesis. — Berkeley, CA: University of California, Biomechanics Laboratory.

Sanders JE (1991) Ambulation with a prosthetic limb: mechanical stresses in amputated limb tissues. Ph.D. Dissertation. — Seattle: University of Washington.

Steege JW, Schnur DS, VanVorhis RL, Rovick JS (1987). Finite element analysis as a method of pressure prediction at the below-knee socket interface. In: RESNA '87 Meeting the Challenge; Proceedings of the 10th Annual Conference on Rehabilitation Technology, San Jose, California, June 19–23, 1987. — Washington, DC: RESNA. 814–816.

Stoughton RB (1957). Mechanisms of blister formation. *AMA Arch Dermatol* **76**, 584–590.

Torres-Moreno R, Solomonidis SE, Jones D (1992). Three-dimensional finite element analysis of the above-knee residual limb. In: Proceedings of the 7th World Congress, International Society for Prosthetics and Orthotics, Chicago. June 28–July 3, 1992. — Alexandria, VA: ISPO. p 274.

Prosthetics and Orthotics International, 1993, 17, 49–55

The ICEROSS concept: a discussion of a philosophy

Ö. Kristinsson

Össur hf, Reykjavík, Iceland

Abstract

Prefabricated ICEROSS (Icelandic Roll On Silicone Socket) sockets have been in use in Iceland since early 1986. Use of custom-made silicone sockets began several years earlier, and a paper devoted to the subject was presented at the 1984 AOPA Assembly by the author of this article.

The ICEROSS system is primarily used for suspension. At the same time the author believes it considerably improves the weight-bearing capability of the prosthesis and the interface between prosthesis and user. After being turned inside out and rolled over the stump, the silicone sleeve forces skin in a distal direction, stabilising soft tissue and minimising pistoning. Both prosthetist and user may experience some problems initially, although most can be overcome by careful socket design and skin care.

Introduction

There are two conceptually different aspects to the quality of prosthetic fittings. Firstly, there is the quality of available components and materials, i.e. tools, database and the craftsmanship necessary to put them together as a functional unit. Secondly, there are the latest methods of fitting a socket and the involved prosthetist's skill.

Apart from the limitations inherent in artificial joints, the single most crucial part of the prosthesis is the socket, which often establishes the threshold of the user's performance.

Fitting trans-tibial amputees is not often regarded as a major problem. In comparison to fitting trans-femoral amputees, it may not be. Nevertheless, because we have very few parameters to judge by, other than previous, empirically-gained knowledge and subjective feedback by our patients, we may often be misled into thinking we are doing a much better job than is the case. That is, we tend to accept our results as the best obtainable. Problems which appear later may be attributed to uncontrollable factors such as oedema, atrophy, redistribution of soft tissue, change in body weight, etc.

Of course, this is in no way unethical or unnatural. No one can be blamed for fittings, whether good or bad, that do not meet standards he is unaware of or are as yet undeveloped. Just as carriage makers had no idea of pneumatic tyres when constructing primitive wooden wheels covered with wrought iron, modern prosthetists have to rely on the state of the art of the present. This is not to say we are in the same predicament as were the carriage makers, we certainly are not. For a start we have an abundance of new materials; we have a multidisciplinary database to tap from, and hopefully, our intentions are undisputed. Most likely, what we lack are the tools that fit every prosthetist's hands and mind and yield comparable results, regardless of location and the artisan. CAD CAM may be the answer, but more research should be directed towards understanding the complex interaction between socket and limb or more properly socket and skeleton.

In any case the author cannot accept any current state CAD as a toolbox, simply because his view is that all systems seem to have inherited the PTB concept, or rely on the information obtained on the topography of a cast or a "hanging limb". It is not considered that CAD has anything to offer yet besides documentation and ease of fabrication. This will hopefully change in the near future as the systems evolve.

All correspondence to be addressed to Össur Kristinsson, Össur hf., Hverfisgata 105, Box 5288, 105 Reykjavík, Iceland.

Suspension

For a long time, suspension has been recognised as an important factor determining the general effectiveness and comfort of the trans-tibial prosthesis. Unfortunately research and development aimed at solving the problems of suspension has not been very successful. In addition to the thigh corset, the PTB strap and similar strapping systems, there are several variations of supracondylar pads and neoprene sleeves, rubber sleeves etc. All these different systems prevent the prosthesis from falling off the limb, and to some degree limit pistoning. This is especially true of the sleeves. None of them, however, can be considered to be very effective.

The trans-tibial suction socket provides a different approach to suspension from the above mentioned systems. Several attempts have been made to solve the problems connected to such devices, with the most promising involving flexible or semisoft sockets. The use of such sockets has not been widespread, perhaps because (with the exception of very voluminous limbs), if the air seal is not to be lost, the sockets have to be made very tight and tend to strangle the limb. Also, they are difficult to fit, and are sensitive to rhythmic or permanent changes of the limb. Such sockets tend to create more problems than they solve.

Generally said, suspension has been so poor that users of trans-tibial prostheses have seen their performance suffer as a result. All the pistoning, due to the prosthesis and limb accelerating relative to one another at every step, and the impact initiating the weight-bearing phase, exerts great stress on the skin, joints and skeleton. To cushion impact, trans-tibial sockets are lined with some kind of soft material to ease shock. This is helpful, of course, but if the fit is not optimal, and only limited areas of the limb are subjected to load, the socket will be uncomfortable, regardless of the quality of the cushioning.

Socket shape and volume

This author's firm belief is that a trans-tibial socket, designed to transfer loads primarily to limited areas of the limb such as the patellar tendon and the medial flare and condyles of the tibia, for instance, is in most cases both ineffective and uncomfortable. The most effective socket, in the author's view, is one that relies on the hydrostatic principle for load transfer. This principle is at work in the majority of "best" trans-femoral fittings. To obtain such a fit, one must realise that the goal of a hydrostatic fit can only be realised to a certain degree. The stump is of a complex mechanical nature, but for means of simplification will be referred to as an elastic solid with low stiffness surrounding a piston, the tibia with its condyles. If this mass can be fitted into a containing vessel corresponding exactly to its volume, the fitting can be expected to behave like a hydrostatic system when loaded. Then, in the absence of motion, there is no shear stress; the internal state of pressure at any point is determined by applied pressure alone. Hence, the pressure at a point is the same in all directions and the pressure required to support the weight would be determined by the cross-section of the socket opening. A hydrostatic system is stable only as long as it is tight. As soon as it begins to leak, it begins to lose its mechanical stiffness. Even if it cannot be referred to as an effectively closed hydrostatic system, the limb-socket interface (interaction) may probably pass as being an elastic coupling with hydrostatic characteristics during weight-bearing. Given that the posterior wall is high enough and that the consistency of the soft tissue prevents it from leaking out of the vessel and past the condyles, this system should be able to transfer the forces generated during weight bearing without any need for limited area loading or even conformity to the underlying skeleton. There are questions about mechanical stability and the internal dynamics of the limb, enclosed and acting in such an environment, that have to be answered. Research in this field is much needed.

Of course, the force transference between socket and limb must be a compromise between limited area loading and hydrostatic loading. Furthermore, the stump of an active amputee most likely is not of the same volume towards evening as it is in the morning, or in June and December of the same year. Hence, the socket fits differently at different times, and hydrostatic stability may be lost at regular intervals, even with a reasonably well fitting socket. What needs to be done, for a good initial fit, is to ensure that the soft tissue present at fitting, is stabilised at ideal volume by an ideally matching container during loading; and, while being suspended, the prosthesis is securely anchored to the limb with minimal longitudinal displacement. No method that solves both these problems was known when development of the ICEROSS was begun.

ICEROSS

The ICEROSS (Icelandic Roll On Silicone Socket) consists of a cylinder closed at one end, with an attachment coupling for distal fastening integrated into the closed end (Fig. 1). ICEROSS is made of a very soft, stretchable silicone material, with extraordinary elongation and tear resistance capability. Being a silicone, however, the material has certain limitations. Careless handling can result in tearing of the wall and destruction of the socket. Consequently, prosthetists must be aware of this and instruct the user in proper care and handling.

From the closed distal end, where wall thickness averages 4-5mm, the socket wall thin out to 2mm at roughly 100mm above the bottom. When the socket is turned inside out, the inner wall (now facing out) of the relatively thick distal portion stretches, and the outer wall similarly compresses. When donning, the user presses the exposed bottom of the ICEROSS against the end of the stump and then rolls the socket all the way over the knee (Fig. 2).

Tension previously created in the stretched inner wall during the inside-out procedure is then released; the surface contracts, displacing the skin downwards. The phenomenon can be observed by the user, who can both feel and see the skin being displaced down into the socket during the rolling-on process. When the ICEROSS has been rolled properly over the stump, the soft tissue is stabilised. The combination of radial pressure and

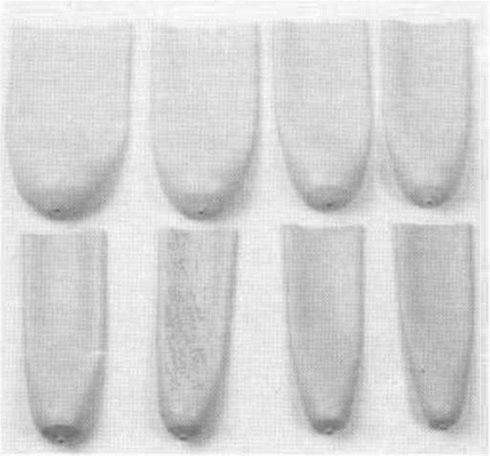


Fig. 1. The ICEROSS socket.

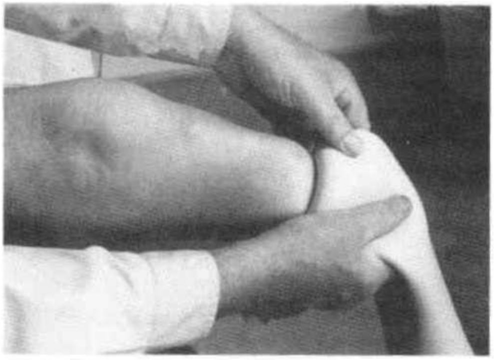


Fig. 2. Donning the ICEROSS socket.

the displacing of the skin in the distal direction presents the prosthetist with an entity quite different than the freely hanging limb. The possibility of longitudinal displacement of the skin envelope has been greatly reduced. Gripping the silicone covered limb with one's hands and attempting to move the soft tissue in distal and proximal direction, in order to imitate pistoning, makes clear the tremendous improvement in body/prosthesis interface this simple device represents.

The silicone is covered with a thin nylon sheet, and then goes into a rigid, laminated or thermoplastic socket. If the limb becomes smaller in volume, a thicker sock can be pulled over. Should it become larger, a new socket is usually required.

Being so thin and easily stretched, the ICEROSS can follow movements of the skin over the knee, and even collapse into concavities without sacrificing its ability to adhere to the skin. The silicone could almost be referred to as a second skin. The interface between skin and socket is free of friction, which has been transferred to the interface between the ICEROSS and the socket. As a result, considerably less strain is exerted on the skin.

An important feature of the ICEROSS is a reinforcement, which is integrated into the thicker distal portion. This part of the socket will stretch radially, but only slightly axially (longitudinally). Restricting axial stretching distally to the knee contributes greatly to the effective suspension. The arrangement makes it possible to counteract the shear forces during suspension. This is possible at least as far up from the bottom as the reinforcement extends, as the skin is locked against a relatively unstretchable

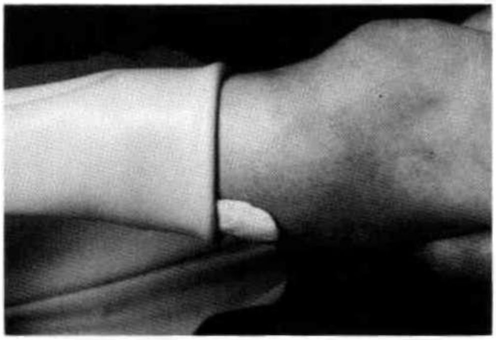


Fig. 3. Positioning of the fibular pad.

Fig. 5. Applying plaster of Paris bandages.

wall with the stump under radial compression. Furthermore, it prevents the socket from stretching away from the end of the stump, which creates problems associated with negative pressure.

The cushioning characteristics of silicones are quite different from those of foams, which are porous or cellular. Cells can be open or closed, and are filled with gas. In the case of open cells, the gas escapes when the material is compressed, collapsing the pores and decreasing the volume. If cells are closed, the encapsulated gas compresses. During this process, the compressing force works against a component of elasticity — the spring component. Depending on the size of the spring component these materials rebound more or less on impact, with more or less "kickback", more for the rubbery foams, less for polyethylene foam for example. Silicone, on the other hand, is solid, and accepts very little, if any, compression. Instead, it redistributes, flowing from high to lower pressure areas. The silicone used is somewhat energy absorbing. Almost no spring component is present, making it an excellent shock absorber.

During initial theorising about hydrostatic fitting or total surface bearing for trans-tibial amputees, we realised that an ideal fit should eliminate the need for cushioning. We were also, however, aware of the fact that even if an ideal matching were achieved between limb and body during fitting, the situation would probably be short-lived. The mature stump is conical. Small fluctuations in its volume result in a mismatch as the limb enters the socket at various depths, and the angle of attack for the sloping surface changes. Hence, if the limb does not sit exactly in the receptable as intended, trouble is bound to occur over bony prominences. To rectify the situation, we build up reliefs for the fibular head and tibia. The only cushioning used is the silicone wall (2mm thick proximally). A 3mm Pelite liner was used initially, but soon abandoned in light of its high friction and because we realised it was unnecessary.

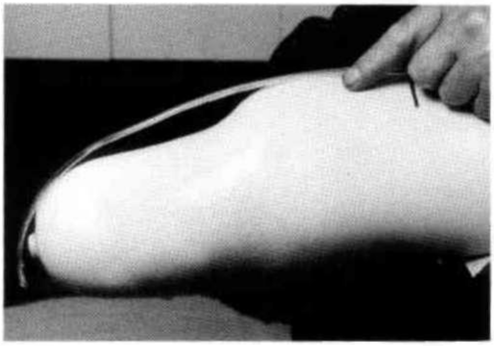


Fig. 4. Position of the stump for casting.

Casting and rectification

When casting for a total surface bearing socket, we recommend the silicone socket be fitted and the cast subsequently taken thereover. Build-up over the tibia and the fibular head can be placed directly on the skin, under the silicone, by using silicone pads of appropriate thickness. In such cases, the fibular pad should not be positioned before the rolled on silicone approaches the relief area (Fig. 3).

Casting is then done with the limb in full extension and relaxed. The silicone is richly lubricated with hand lotion (Fig. 4).

Normal plaster of Paris bandages are used, 15cm wide, and most of the plaster is drained off. No local pressure is applied anywhere, the cast is only smoothed with firm hands and the anterior towards the posterior (Fig. 5). Pressure is not applied under the patella or on the patellar

Fig. 6. Removal of the cast.

tendon. Excessive casting material collects over the posterior aspect.

The mediolateral (ML) dimension is measured with a pair of calipers. The cast is cut over the patella (inclined cut with a scalpel) and pulled off (Fig. 6).

Those afraid of cutting the silicone should place a plastic tube under the cast. Using the measurement taken with the calipers, the ML measurement of the cast is adjusted. Do not cut out the posterior opening, just pour plaster into the unmodified cast. A build-up for the posterior opening is added to the model and the only rectification is for volume (Fig. 7).

Check sockets

A check socket procedure is considered essential. These are made of a transparent thermoplastic, and should be stiff enough to prevent losing their shape during the test procedure. Sockets which are too tight initially, a problem sometimes difficult to spot, are detrimental for our purpose. If anything, we prefer to begin with a socket that is too loose. It is easy to go back to the model, remove material, reheat the socket locally and shrink it with the aid of an elastic bandage.

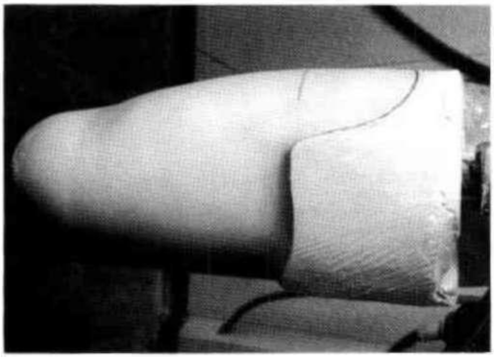


Fig. 7. Build-up for the posterior opening.

A socket made to fit by stepping into, but lacking any allowance for forced elongation, may have to be made too short and too wide. Pistoning can be the result. In many cases it is preferable to use a pulling string system rather than a shuttle lock of some kind. On the other hand, eliminating all relative movement between limb and prosthesis may not be desirable. Even a snug socket, fitted by first displacing the skin into the ICEROSS, and followed by stretching the silicone covered limb distally so that it will bottom in the receptacle, shows some degree of pistoning when intermittently loaded in both directions. When the string has been locked against the check socket and a load applied in the distal direction, a gap can be observed between the socket and the silicone covered limb. Usually, a nylon sheath is pulled over the silicone to reduce friction, because, by allowing for the limited pistoning, less strain is put on the skin at the interface with the silicone. In addition, a more dramatic mismatch becomes obvious when the knee is flexed, and disallowing movement of the limb and socket relative to each other is a sure way to invite trouble.

Trans-femoral prosthesis

The combination of an ICEROSS and a vaguely narrow mediolateral shaped trans-femoral socket is often very successful for elderly patients (Fig. 8).

The socket can be made of thermoplastic with a medial frame, having small proximal extensions, just enough to secure the socket against rotation relative to the frame (Fig. 9).

An even more flexible socket is made of polyurethane elastomer with an integrated frame. Elderly patients are very pleased with the softness of the polyurethane sockets and it is amazing to see how easily they roll on their ICEROSS and step into their prosthesis. It only takes a minute or two.

Contraindications

Use of silicone sockets for the elderly (and the disabled in general) is only recommended when adequate care and handling can be guaranteed. Consequently, sockets should not be prescribed for persons unable to put them on and exercise proper hygiene, unless caretaking personnel or relatives are available for assistance.

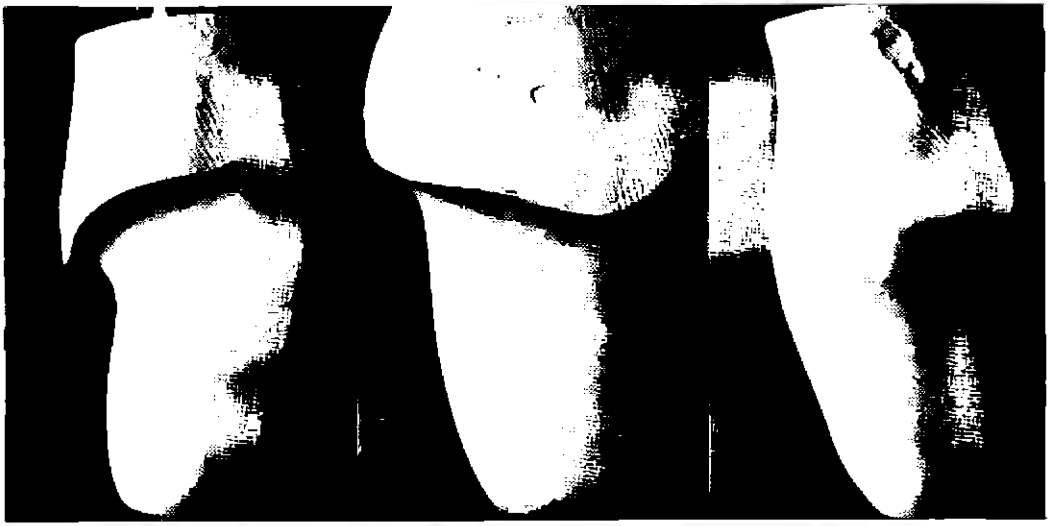


Fig. 8. The cast for the trans-femoral socket.

Open wounds and skin necrosis: Silicone sockets are not employed before the post-operative wound has healed. They should, however, be prescribed as soon as possible as a shrinker, replacing elastic bandaging. A study on post-operative application is underway, but at present no benefits have been verified.

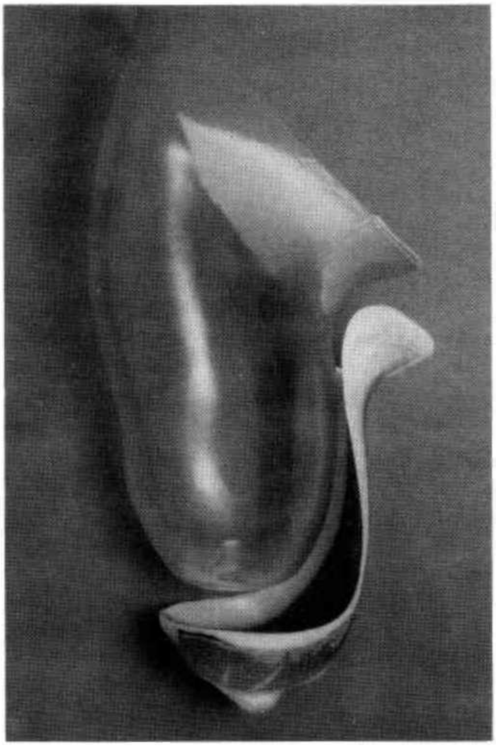


Fig. 9. The trans-femoral socket.

People occasionally come in with damaged skin caused by excessive local pressure. Healing is effected by relieving the area under pressure, and usually does not require rest and airing.

Complications

Elderly patients may develop blisters, especially on the patella, in the early stages of training. Apart from the fragility of the skin, one possible reason may be the selection of a socket size which is too small. This problem can be remedied by cutting out an opening to relieve the patella, and using a soft dressing as protection. After healing, trials show blistering ceases if careful readaptation is exercised, accompanied by regular resting and airing of the limb.

Conclusion

The ICEROSS system is primarily used for suspension. At the same time it apparently considerably improves the weight-bearing capability of the prosthesis and the interface between prosthesis and user.

Total surface bearing or hydrostatic sockets are recommended. Properly fitted such a socket does not require an additional soft liner.

BIBLIOGRAPHY

Dietzen C, Harshberg J, Pidikiti R (1991). Suction sock suspension for above-knee prostheses. *J Prosthet Orthot* **3(2)**, 90–93.

FILLAUER CE, PRITHAM C, FILLAUER K (1989). Evolution and development of the silicone suction socket (3S) for below-knee prostheses. *J Prosthet Orthot* **1(2)**, 92–103.

GALDIK J (1955). The below-knee suction socket. *Orthot Prosthet App. J* **9**, 43–46.

GREVSTEN S (1974). Studies of the suspension of below-knee (BK) prostheses. *Acta Orthop Scand* **45**, 971–972.

GREVSTEN S (1977). Patellar tendon bearing suction prosthesis. Clinical experiences. *Upsala J Med Sci* **82**, 209–220.

GREVSTEN S, ERIKSON U (1975). A roentgenological study of the stump-socket contact and skeletal displacement in the PTB-suction prosthesis. *Upsala J Med Sci* **80**, 49–57.

GREVSTEN S, MARSH L (1971). Suction-type prosthesis for below-knee amputees: a preliminary report. *Artificial Limbs* **15**, 78–80.

ISHERWOOD P (1977). The controlled pressure distribution casting technique. A hydraulic technique for PTB sockets. *Prosthetic Instruction Manual, 1977.*

PRITHAM C (1979). Suspension of the below-knee prosthesis: an overview. *Orthot Prosthet* **33 (2)**, 1–19.

RADCLIFFE CW, FOORT J (1961). The patellar-tendon bearing below-knee prosthesis. – Berkeley, CA: University of California. p 82.

REDHEAD RG (1979). Total surface bearing self-suspending above-knee sockets. *Prosthet Orthot Int* **3**, 126–136.

SABOLICH J, GUTH T (1986). Below-knee prosthesis with total flexible socket (T.F.S.): preliminary report. *Clin Prosthet Orthot* **10**, 93–99.

Prosthetics and Orthotics International, 1993, 17, 56–64

A clinical experience with a hierarchically controlled myoelectric hand prosthesis with vibro-tactile feedback

P. J. KYBERD*, N. MUSTAPHA†, F. CARNEGIE**, P. H. CHAPPELL††

**Oxford Orthopaedic Engineering Centre, NOC, Headington, Oxford, UK.*
†Garston Manor Medical Rehabilitation Centre, London, UK.
***Arm Training Unit, Queen Mary's Hospital, London, UK.*
††Department of Electrical Engineering, University of Southampton, UK.

Abstract
Improved performance of externally powered myoelectric hands is possible when the direct control of the digit flexion and grip force are given over to an electronic controller which frees the operator to concentrate on other demands.
Design: A commercial myoelectric hand was modified to take the new touch and slip sensors and novel control method.
Subject: An adult male with a traumatic mid-forearm amputation.
Outcome measure: The range and ease of use of the prosthetics system.
Result: The hand was easily and usefully operated in the home and work environment.
Conclusion: Hierarchical control of a hand is possible using sensory feedback to a sophisticated electronic controller. Such a control method reduces the demands on the user's concentration and enhances the hand's range.

Introduction

The survival rate of an individual following amputation, prior to the development of successful anaesthesia, was poor. History records a few hardy individuals who survived (Pliny; Herodotus; Childress, 1985). The replacement limbs were often simple. The more sophisticated hands were often based on the techniques developed by armourers in building articulated gloves. Once the survival rate improved, the opportunities for commerical exploitation also grew and companies formed, the oldest in the UK being over a hundred years old.

All practical prostheses were body powered and this continued to be the major form of actuation until the Seventies when electric sources became practical: such devices are fitted to a small proportion of the population.

In the research arena, other forms of power sources have been used as far back as 1916 (Childress, 1985), for example carbon dioxide gas under pressure. However, none of these have achieved clinical significance, though small numbers of people continue to use gas powered arms. This is due to a variety of reasons, from the limited capabilities of the power supply, to the availability of power sources (Millstein *et al.*, 1986; Simpson, 1972).

The increase in the levels of complexity and the integration of electronic circuits and some improvements in the technology of electrical storage, have encouraged experimental designs of hand and controller that provide better performances or longer periods between re-charging than current designs (Gow and Douglas, 1990; Chappell and Kyberd, 1991).

Control of prosthetic hands

In the clinical setting there are still only two widely used means of control of prostheses. The first is in body powered terminal devices which usually are in the form of a split hook where control is by body movement. The second form

All correspondence to be addressed to P. J. Kyberd, Oxford Orthopaedic Engineering Centre, NOC, Headington, Oxford, UK.

is the electrically driven cosmetic hand controlled by myoelectric signals derived from muscles (usually antagonistic pairs) on the wearer's stump. Both types of prostheses have a single driven action and a passive wrist.

Other hand geometries and prosthesis forms have been suggested (LeBlanc *et al.*, 1987; Childress, 1972). Few examples have achieved commercial exploitation, the exceptions including the NU-VA Synergetic prehensor, the Steeper powered hook and the CAPP-II-TD (Patton, 1988; Bennett Wilson, 1989).

The level of acceptance and use of a particular hand depends on a large number of factors. However ease of use is an important aspect. The body powered hook is successful in part due to its geometry, allowing a wide range of objects to be held. In addition its control is based on simple body movements. This allows a high degree of control and so a high level of confidence to be developed by the operator. The system has a number of drawbacks including poor cosmesis. In cases where the stump is poorly padded, following traumatic amputation, the force required to open the hook must be transferred through the stump end causing pain.

The action of an electrically operated hand is commonly voluntary opening and voluntary closing. The commands are obtained from the

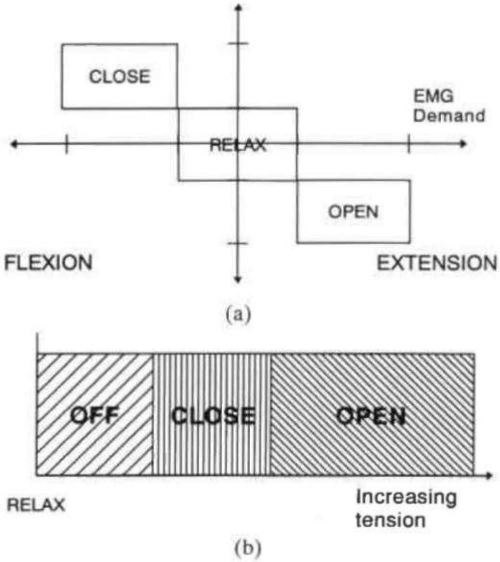


Fig. 1. Examples of conventional control schemes for myoelectrically controlled hands. (a) Two-site, two-state. (b) Single site, three-state (Scott, 1988).

electrical signals generated when the muscles contract (known as myoelectric signals or EMG signals). The strength of the signal is dependent on the muscle tension. Extensor tension which exceeds a set threshold opens the hand, while sufficient flexor tension closes it. Relaxation disables the motor (Fig. 1a). This representation in Figure 1 of muscle activity is used throughout (Scott, 1988). The signals due to the flexor and extensor muscles are plotted on the horizontal axis. Extensor tension increases to the right, flexor tension to the left. For antagonistic pairs of muscles there is a single range from maximum flexion to maximum extension, through a central region where both muscles are relaxed. The drive mechanism leaves the hand locked open at that point. If the hand is left fully extended the result looks unnatural, so during training users are taught to leave the hand closed if it is to remain idle. Alternatively, a single muscle can be used to command opening and closing of the hand (Fig. 1b) but the operator must pass through one direction command to reach the other.

To hold an object the hand is opened wide enough to admit the object and is then closed around it. The wearer either judges when to stop closing around the object by eye, or, when the sight-line is obscured, allows the hand to stop when the controlling circuitry stops the hand. This latter option is easier on mental effort but provides very coarse control of the grip force; a delicate grip is difficult to achieve using this method. To enhance their control the users may utilise other information that is available from the prosthesis but this accidental path is not designed into the mechanism.

The geometry of the hand limits its functional range and the coarse grip force control ensures some operations cannot be performed successfully by such a hand. The hands are thus most often used in cosmetic and support roles.

An additional limitation of the hand's use is that myoelectric channels are inherently noisy. In setting up a working prosthesis the users must set the threshold of action for the EMG pickups so the hand is useful to them. A large amplification sets the threshold low so the hand is easily operated. This may lead to inadvertent opening of the hand when an object is being gripped. To ensure a more secure grip flexor tension must be periodically reapplied thus losing any finesse in the management of the grip tension. Alternatively the actuating threshold can be set higher making the grip more secure but the hand more tiring to use.

Another form of arm control has been proposed (Gow *et al.,* 1983; Simpson and Kenworthy, 1973). This is known as Extended Physiological Proprioception (EPP), developed by D. C. Simpson at the Bioengineering Unit of Princess Margaret Rose Hospital in Edinburgh, Scotland, for use on the Simpson arm. The key to this form of control is the appropriate nature of the input and return signals. A force command is returned to the input lever as a change in position in proportion to the position of the end of a multi-degree of freedom arm. This mimics the body's own system where the effect of a muscular force is detected as a positional change in the arm. Thus the EPP driven arm extends the users awareness along the length of the arm. The result is accurate and easily controlled (Gow *et al.,* 1983). It is essentially a method of accurately and easily positioning a terminal device in space. However, recent studies suggest that it can be modified to transfer *touch* information, (Meek *et al.,* 1989). This suggestion ignores the central aspect of EPP, the appropriate nature of the feedback, It is the *position* of the lever that reflects the *position* of the arm.

A widely functional hand requires a greater number of degrees of freedom than the conventional number available. When an arm prosthesis has multiple degrees of freedom it requires separate myoelectric channels for the control of each independent axis of motion. This technique is not easy to learn or control for hand or arms.

A practical multi-degree of freedom hand is potentially capable of a wide range of actions providing its control is of a sufficient standard yet it must also be easy to use. Thus it is necessary to devolve the detailed control of the hand to a computer and allow the supervisory actions to be taken by the operator, much as the human Central Nervous System (CNS) separates the positional reflexes from the gross intentions of the person. The Southampton Adaptive Manipulation Scheme (SAMS) performs this task.

SAMS

SAMS was initially conceived to be used in a multi-degree of freedom hand (Chappell and Kyberd, 1991; Baits *et al.,* 1969; Kyberd, 1990; Senski, 1980) but the system's virtues can be applied to simpler hands as well. The hand described here is of that form.

Although there are forms of proportional voluntary control available, these are crude and require constant attention by the user. Good control is achievable by exceptional operators. SAMS achieves enhanced performance by feeding information about the grip force and any movement of the held object back to the controller using a vibro-tactile sensor (Chappell *et al.,* 1987). Using this method the mode of operation becomes simpler.

The control input is via two single channel EMG amplifiers. Figure 2 shows the two signals from the flexor and extensor muscles plotted on the horizontal axis. The central region is where both muscles are relaxed. Extensor tension increases to the right, flexor tension to the left. The vertical axis shows the control states, (OPEN, HOLD, POSITION, SQUEEZE and RELEASE) the boxes show the degree of hysteresis available to each command. Figure 3 shows the regions of proportional response. The scale represents the degree of flexion of the finger, which is in direct proportion to the extensor tension. Grip force in response to the SQUEEZE demand is in proportion to the flexor tension, and is shown as increasing in the negative direction.

Extensor tension opens the hand by an amount in proportion to the tension. Relaxation of the muscle allows the hand to close once more. This is voluntary opening, involuntary closing, in a similar form to the body powered split hook. Extensor tension below a set threshold is taken by the controller to be a relaxed state. If the sensor makes contact with an object while the hand is closing,

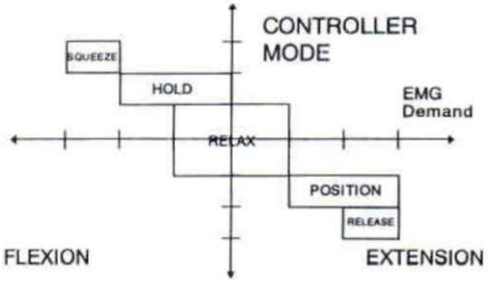


Fig. 2. Command states showing hysteresis of EMG bands.

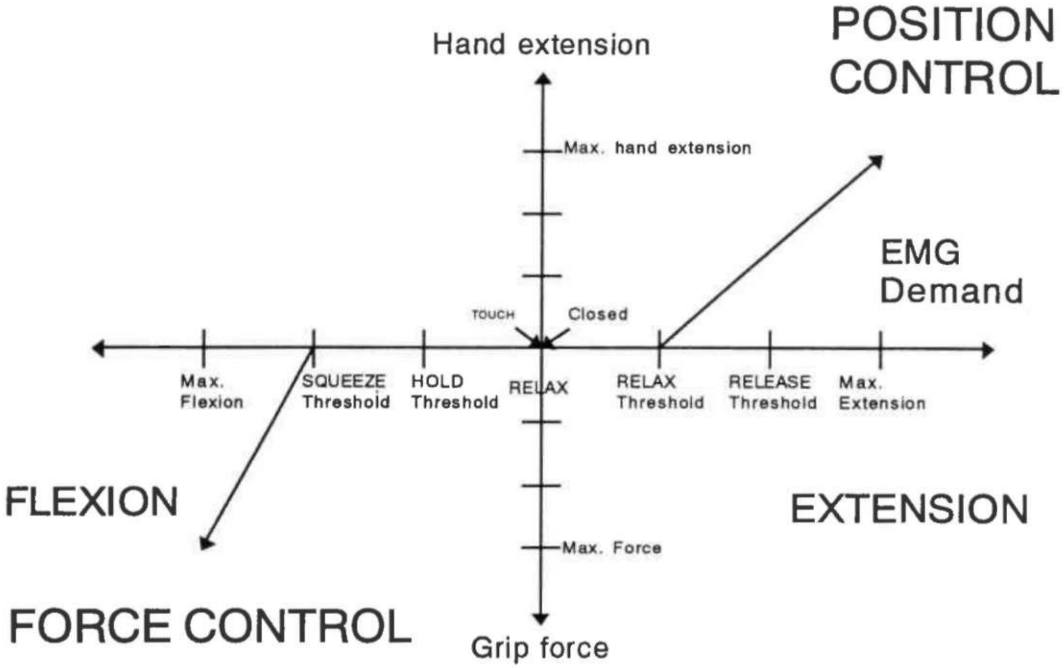


Fig. 3. Control scheme of single degree of freedom Southampton hand.

the movement is stopped so only the lightest touch is applied. When the touch sensor is activated sufficient flexor tension enables the HOLD mode. Once in HOLD the automatic force control is activated. Input from the vibro-tactile sensor detects a slipping object and the controller increases the tension accordingly. Muscle tension less than the SQUEEZE and RELEASE thresholds does not affect the controller.

Whilst is the HOLD mode, should the wearer wish to increase tension and override the slip control, a SQUEEZE mode is available. The motor force increases in proportion to flexor tension.

To prevent inadvertent squeezing when the HOLD mode is initially invoked, the EMG must return to within the relax range before an excursion beyond the SQUEEZE threshold is recognised as a squeeze demand. At any time when in HOLD or SQUEEZE extensor tension beyond the release threshold will cause the hand to release the object by opening the hand slowly to the position of the first contact, or (if the hand still touches the object) beyond to the point where contact is broken. Relaxing the user demand to the RELAX range returns the control to the position mode.

It can be seen that the result of such action is to free the operator from the detailed control tasks. The hand only applies the minimum force to maintain a stable grip. In addition the hand automatically closes when empty and not in use and the knowledge of whether the hand is empty or holding an object means the different EMG thresholds can be set widely differently making prehension automatic and object release very deliberate.

Although a bipolar command channel was employed, the controlling programme can simply be modified to allow a single site channel operation.

Similarly, the EMG input could easily be changed to a number of different forms. By converting the input to an electrical signal, the contoller can then be reprogrammed to use the data accordingly. Thus, for example, a purely mechanical input via a pull cord or acoustic myography (AMG) could be used, (Barry *et al.*, 1986).

It is the autonomous nature of the control of the hand which implies that the use of the system would be broadly subconscious. Thus the only effective test of the system is its long term use in the field. To assess the potential of the concept a portable controller was devised and fitted to an individual to allow a limited field test to be carried out (Chappell *et al.*, 1987; Kyberd, 1990).

Materials and methods

A modified commercial single degree of freedom myoelectric hand was adapted to carry sensors controlled by an electronic controller. The controller was built upon a single printed circuit board within a case (220mm × 150mm × 25mm). The EMG amplifier was mounted in a separate metal enclosure to shield it from external interference.

The hand was an MM3 produced by Viennatone, driven by two conventional 12V batteries, the electronics by a further two PP6 batteries. The hand was mounted on a standard self-suspending socket with a passive wrist. The wiring led to a pouch worn over the shoulder of the subject. The basic elements of the system are shown in Figure 4.

Although this arrangement is far from optimal the aim of the experiment was to assess the potential of the system prior to a more extended trial. The electronics and software had undergone extensive testing within the laboratory. It was therefore a test of the control philosophy and not the hardware. Similarly, as the design was made independently of the prosthesis it could be applied to any electric hand. The choice of the Viennatone device was made on the basis of availability. No assessment was made on the comparative speed or lightness of devices as the factors are dependent on the vehicle rather than the control strategy.

A display of lights was used for training purposes but was removed when the hand was used continuously. The individual lights showed the status of the controller in HOLD or POSITION states.

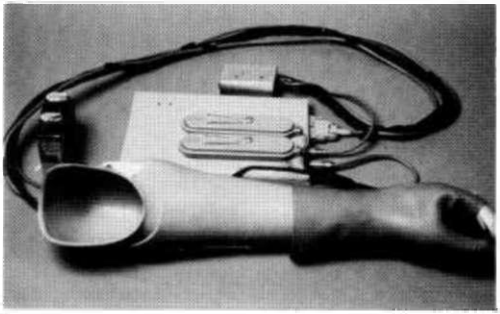


Fig. 4. The single degree of freedom Southampton hand system used on field test.

In the initial experiments the EMG amplifier/processor was mounted near to the electrodes, but it was necessary to move it to a more distant site for the field trial, with a subsequent decrease in its reliability.

Patient trials and results

Initial tests were carried out by one of the authors (PJK) using a right hand prosthesis mounted upon a splint. The splint was made of the thermoplastic Plastazote. The splint extended from the elbow to the knuckles, which immobilised the natural hand, and so encouraged the use of the prosthesis. It also allowed the muscles to contract principally in an isometric fashion similar to the manner of control by an amputee.

The hand was mounted with its centre line 10cm from the centre of the wearer's arm, and was positioned parallel to the natural hand. Though this made the hand off-centre to the arm, which changed the natural hand/eye relationship, most problems were overcome with practice. The one exception was that heavy objects caused the splint to rotate on the arm and disturb the EMG pickup.

The tasks performed ranged from manipulation of abstract objects, to the performance of everyday actions.

During the tests the natural right hand was not used and attempts were made to use the prosthesis in the dominant role. Certain tasks, such as rewiring a plug required the prosthesis to be used as a vice with the more dexterous tasks being performed by the left hand. The tests were also performed in parallel with the experiment in the field, providing insights into the systems's capabilities.

The system was then used in the field by a single adult male who had lost his left hand at the wrist in an industrial accident. He normally used a Steeper myoelectric hand for daily activities. His myoelectric control was good, thus he would derive less advantage with the new controller compared with that enjoyed by a less adept user. It can therefore be argued that any advantage experienced by the subject represents a significant improvement in control.

The hand was fitted at the limb fitting centre at Queen Mary's Hospital, Roehampton, London. The left hand was fitted to a standard arm socket copied from his normal prosthesis. The outer moulding was made to accommodate the wires to the controller and the EMG amplifier could be held in a pouch slung over the shoulder (Fig. 5). Five two-hour sessions were conducted with the subject accustoming himself to the different manner of control of the device. In the interval between the later sessions he took the system home and used it in the domestic and work environments. A video recording was made of the test.

The training consisted of performing the standard manipulative tasks conducted at the Arm Training Unit.

The tests included picking up a range of standard objects from a board and placing them in a box. The shapes were cones, cylinders and blocks of varying sizes as well as a cup and ball. The time taken for the board to be cleared was recorded. The varying sizes and shapes of the test objects ensured the need for the subject to reorient the hand and to adjust the grip and approach for each object. Less abstract tasks were also used, such as rewiring a plug, cutting and manipulating paper.

Results

The subject rapidly became adept at using the system.

The initial control was inaccurate as he attempted to use it as if it were his normal prosthesis (voluntary closure as well as opening). This tendency was reduced as his confidence grew and the advantages of automatic grip tension became appreciated. Pick and place tasks were performed with skill and precision.

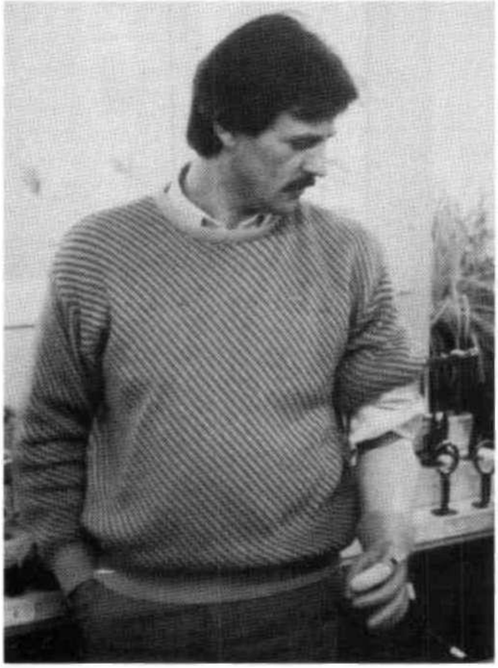


Fig. 5. The single degree of freedom Southampton hand on field test.

The speeds of operation of his conventional hand and the SAMS hand were very similar, so that the time taken for the pick and place operation was broadly the same. However the measure of the utility of the hands was the level of concentration required for the tasks.

System operation

Problems were encountered in four areas:

Socket: The subject suffered discomfort when wearing the socket as it was too tight and this limited the time the hand could be worn over one session. Each time the arm was put on it required talcum powder or a water based cream to ease the insertion. These problems were unrelated to the electronic system itself.

Wiring: The principle causes of failure of the system were due to the wiring to the EMG amplifier. This alone was responsible for 80% of all failures while on the subject. The signal from the muscles is small and presents a high impedance to the wires conveying the signal. Pre-amplification close to the site would alleviate the problem. Very few failures were encountered when the splint was used. For this the EMG amplifier and signal processor were mounted close to the muscles and wires were short. This arrangement would have been inconvenient with the field tests. The construction of a new compact amplifier/processor is possible, but a more logical path would be to use a commercial analogue amplifier and perform the additional processing in software (Kyberd, 1985).

Electronics: The electronic controller itself suffered no failures. The sensor ultimately fatigued. The sensor had been used over a period of 18 months of testing. Once the worn component was replaced the sensor operated satisfactorily.

Batteries: Two rechargeable 12V batteries were used. A single unit normally powers the unmodified hand, thus they were adopted as a matter of course. The batteries themselves had poor reliability and charge retention qualities. Three cells burst during operation or recharging. The batteries used on the other commercial system the subject normally wears, were much more reliable but were only 6V. The 12V cells supplied power for 1 hour's continuous use but the modifications of the system to 6V from 12V was not feasible.

Discussion

Minor alterations were made to the system as a result of these tests. One such change was suggested specifically by the subject. It was an alteration to the invocation of the HOLD command. The user must pause in the relaxed state before proceeding to HOLD to ensure that an accidental HOLD command is avoided. Also HOLD is only invoked by flexor tension once an object is in contact with the hand. The point of contact occurs when the hand is partly extended so the user may still have some extensor tension, thus there is a delay in waiting for the HOLD state to be asserted after contact and the EMG must be relaxed and then applied. This was felt to be inconvenient.

The provision of a pre-HOLD state, when the hand is touching the object prevents the hand from gripping tighter inadvertently despite object movement. This was provided to allow objects to be manoeuvred before being held. This especially caters for bilateral users who do not have fine control with either hand and so must manoeuvre the object while in the hand. In the present subject's case, he felt the if he wished to change the grasp of the object or its attitude in the hand, he would relax and re-grasp it using his natural hand to assist the task. This option is not present for the bilateral user. For the subject this feature was removed. In any future trials such customising could be made available depending on user taste and requirements. This is the advantage of a computer controlled hand, as the expensive structural modifications to the system can be kept to a minimum. In addition the adaptable nature of the control philosophy means it can be applied to any form of externally powered prosthesis, mechanical or anthropomorphic. It is important that such choices are available to the user population at large.

One aspect of the user control of standard prostheses not widely reported concerns the use of motor vibration feedback to the wearer. Discussions with the subject showed he utilised this a great deal. Since he possessed a long stump, his arm was very close to the motor and so could easily feel it vibrating. Anecdotal evidence suggests this is a commonly used feedback path with many patients, whether the loss is congenital or traumatic. This includes those with much shorter stumps. This use of other signals depends on the current limit circuitry on the motor drive. Once the motor stalls the drive power is cut and the user feels the change in vibrations. A second application of closure demand indicates resilience of the object and the stability of grip, based on the length of time the motor runs before it stops once more. Greater compliance or a slipping grip will allow the motor to turn for longer before it cuts out. Thus the user can 'feel' this without recourse to visual cues. This unexpected use of feed-in demonstrates the adaptability of the human subject. However it is a less sensitive form of grip tension control than that realised using SAMS control.

A single touch/force sensor was used and was mounted on the tip of the index finger. The prosthesis is designed to hold objects in a three jaw chuck grip between the thumb and the first two fingers. The thumb is therefore a better location for the primary sensor. In addition all objects must be held in contact with this sensor, so great care had to be exercised when gripping certain objects to ensure this was the case. When additional touch/no touch sensors were mounted on the palmar surface of the hand these objects became easier to grip and manipulate (Kyberd, 1990).

Conclusions

The limited field test of the single degree of freedom variant of the Southampton hand can be considered to be a success, though limited in nature. Five conclusions can be drawn;

1. It is possible to modify conventional prostheses to take the Southampton control scheme.
2. The control scheme is easy to learn and use. Its utilisation enhances the function of a prosthesis.
3. The use of adaptive thresholds on a myoelectric prosthesis eases the use of the device.

4. Microprocessor control of a prosthesis allows for easy adaptation to different user requirements.
5. The additional cost in power consumption is negligible and far outweighed by the benefits.

Compared to the 'subjects' prescribed hand, for the test hand the standard pick and place operations were very similar in length of time to execute. To provide qualitative data a broader range of tasks would have to be devised to explore the differences, not merely to exploit the modified device while hiding its flaws.

It is possible to construct a much more compact device that would reduce the bulk, weight and problems of unreliability of the current system. This would use more recently developed electronics such as microcontroller devices. The use of a Very Large Scale Integration technology would also allow far better recording of the use of the hand whilst in the field, although this facility and the application of the information gained must be used with caution to avoid arousing the fears of the user population. This was revealed at a United Kingdom meeting of the International Society of Prosthetics and Orthotics where representatives of the British Limbless Ex-Servicemans Association expressed concern over the use of such data to justify the withdrawal of appliances when a similar data logger for a leg prosthesis was demonstrated, (Pearson *et al.*, 1990).

It is important to appreciate the comparison of the prototype system described herein with a standard single degree freedom device is based on the control aspects and not on the mechanical function. The mechanical attributes were similar, as were many of their drawbacks. It is these drawbacks that a four degree of freedom hand addresses (Kyberd, 1990).

Many of the problems encountered with the hardware were related to the highly experimental nature of the equipment used. What is required is a longer trial on a device developed specifically for the application. This task is now in progress under the TIDE (Technology for the socio-economic Integration of the Disabled and Elderly) initiative of the European Community. The experiment is designed to place six units based on the SAMS technology in two centres (UK and Italy) for an extended trial.

Acknowledgements

The authors would like to thank J. E. Hanger and Co. for support of the project, John Cooper and Julian Hawkings of the company for their help. In addition we acknowledge and thank Robin Cooper and Barry Fazakerly of Hugh Steeper Ltd. for their help in the trial and finally to George Lawson for his help and patience.

REFERENCES

Baits JC, Gatto E (1969). The feasibility of an adaptive control scheme for artificial prehension. In: Proceedings of a symposium at the Institute of Mechanical Engineers Czechoslovakia 31st October — 1st November 1968. — Czechoslovakia: I Mech Eng, 1969.

Barry DT, Leonard JA, Gitter AJ (1986). Acoustic myography as a control signal for an externally powered prosthesis. *Arch Phys Med Rehabil* **67**, 267–269.

Bennet Wilson A (1989). Limb Prosthetics./6th edition New York: Demos.

Chappell PH, Nightingale JM, Kyberd PJ (1987). Control of a single degree of freedom artificial hand. *J Biomed Eng* **9**, 273–277.

Chappell PH, Kyberd PJ (1991). The prehensile control of a hand prosthesis by a microcontroller. *J Biomed Eng* **13**, 363–369.

Childress D (1985). Historical aspects of powered limbs prostheses. *Clin Prosthet Orthot* **9(1)**, 2–13.

Childress DS (1973). An approach to powered grasp. In: Advances in External Control of Human Extremities; proceedings of the Fourth International Symposium, Dubrovnik, August 28 — September 2, 1972. — Belgrade, Yugoslav: Committee for Electronics and Automation.

Gow DJ, Dick TD, Draper ERC (1982). The phsiologically appropriate control of an electrically powered hand prosthesis. In: Abstracts; International Society of Prosthetics and Orthotics 4th world congress, London, 5-9 September 1983. – Copenhagen: ISPO.

Gow DJ, Douglas WB. The REACH electrical hand prosthesis for transcarpal amputees. a progress report. In: Abstract; UK annual scientific meeting, 4–6 April 1990, Edinburgh. International Society for Prosthetics and Orthotics — ISPO.

Herodotus, Calliope IX./Translated by H. Cary. — London: George Routledge and Sons Ltd, 1987.

Kyberd PJ (1990). Algorithmic control of a multifunction hand prosthesis. PhD. Thesis. — Southampton: University of Southampton.

Kyberd PJ (1985). Processing of electromyograms for the control of a multifunctional hand prosthesis. MSc Thesis. — Southampton: University of Southampton.

LeBlanc M, Parker D, Nelson C (1987). Do we need a prosthetic prehensor which is neither a hook nor a hand? In: RESNA '87 Meeting the challenge; proceedings of the 10th annual conference on Rehabilitation Technology, San Jose, California, June 19–23 1987. – Washington, DC: RESNA.

Meek SG, Jacobsen SC, Gould PP (1987). Extended physiological traction: design and evaluation of a proportional force feedback system. *J Rehabil Res Dev* **26(3)**, 53–62.

Millstein HG, Heger H, Hunter GA (1986). Prosthetic use in adult upper-limb amputees: a comparison of body powered and electrically powered prostheses. *Prosthet Orthot Int* **10**, 27–34.

Patton JG (1988). Upper-limb prosthetic components for children and teenagers. In: The comprehensive management of the upper-limb amputee./edited by D. J. Atkins, R. H. Meier — New York: Springer-Verlag.

Pearson T, Convery P, Simpson D (1990). Development of miniaturised equipment to monitor the use of prescribed prosthetic and orthotic equipment. In: Abstract: UK Annual Scientific Meeting, 4–6 April 1990, Edinburgh. International Society for Prosthetics and Orthotics — ISPO.

Pliny the younger: Natural histories. **2(7)** xxvii./ Translated by H. Rackham. — London: W Heinemann 1942. 104–106.

Scott RM (1988). Biomedical engineering in upper-limb prosthetics. In: The comprehensive management of the upper-limb amputee./edited by D. J. Atkins, R. H. Meier — New York: Springer-Verlag.

Senski T A consumer's guide to "bionic arms". *Br Med J* **12 July**, 126–127.

Simpson DC (1973). The control and supply of multimovement externally powered upper limb prosthesis. In: Advances in External Control of Human Extremities; proceedings of the Fourth International Symposium, Dubrovnik, August 28 — September 2, 1972. — Belgrade Yugoslav: Committee for Electronics and Automation.

Simpson DC, Kenworthy G (1973). The design of a complete arm prosthesis. *Biomed Eng* **8**, 56–59.

Book Review

How to Select and use Manual Wheelchairs.
A. Bennet Wilson Jr.
Rehabilitation Press, 1992.
ISBN 1-880902-04-4.
67pp. USD 12.50

This is a useful small book intended, in the author's words, "to introduce new users to manual wheelchairs". It commences with good summaries of manual wheelchair types and their features and a brief history of how these have developed. Useful sections summarise the types of cushions which are available and the prescription process to select the most appropriate wheelchairs. The following section on Seating however is very brief probably because it is too complex a subject to incorporate in a book of this nature. From the user's viewpoint, probably the most interesting sections relate to operating the wheelchair from a simple theoretical viewpoint, followed by practical hints on usage, maintenance and care of the wheelchair.

The book is written from an American viewpoint as reflected in the types of wheelchairs described and the appendix of American manufacturers. An example of this characteristic is that there is virtually no mention of attendant propelled manual wheelchairs. These account for some 50% of the wheelchairs in some countries such as the United Kingdom.

The most useful and valuable message which comes out of this book relates to the wide range of needs a user may have on a wheelchair and the corresponding wide range of wheelchairs which are available to fulfil these needs. The selection of an apparently simple device is in fact quite complex.

Given its limitations, this small book could be very valuable for the user who is sufficiently enthusiastic and interested to participate actively in the process of selecting a wheelchair and subsequently using it to its best potential. Such users would derive considerable benefit from this information but not all users would necessarily have this motivation to derive the benefits from the book. Despite its leaning towards the user, the book would also be extremely beneficial as an introduction to professionals embarking on work involving wheelchairs.

In summary, this book is a most welcome addition to the literature on wheelchairs and is of relevance and interest to both users and professionals. It highlights the need for a similar book on powered wheelchairs.

Geoff Bardsley
Rehabilitation Engineer
Dundee Limb Fitting Centre
Dundee, Scotland

Calendar of Events

10–12 May, 1993
Annual Conference of the American Spinal Injury Association, San Diego, USA.
Information: ASIA, 2020 Peachtree Road NW, Atlanta, GA 30309, USA.

26–28 May, 1993
2nd European Conference on the Advancement of Rehabilitation Technology, Stockholm, Sweden.
Information: Ms. Catarina Brun, ECART 2, Swedish Handicap Institute, Box 510, S-162 15 Vallingby, Sweden.

1–4 June, 1993
9th European Congress of Physical Medicine and Rehabilitation, Ghent, Belgium.
Information: Medicongress, Waalpoel 28, B-9960 Assende, Belgium.

10–12 June, 1993
7th Congress of the European Society for Shoulder and Elbow Surgery, Aarhus, Denmark.
Information: Orthopaedic Hospital, Randersvej 1, DK-8200 Aarhus N. Denmark.

12–17 June, 1993
16th Annual Conference of RESNA Rehabilitation Technology, Las Vegas, USA.
Information: RESNA, Association for the Advancement of Rehabilitation Technologies, Suite 700, 1101 Connecticut Ave. NW, Washington, DC 20036, USA.

13–16 June, 1993
9th Nordic Meeting on Medical and Biological Engineering, Lund, Sweden.
Information: 9th Nordic Meeting on Medical and Biological Engineering, c/o Medicinsk Teknik-kansli, Lasarette i Lund, S-221 85 Lund, Sweden.

13–17 June, 1993
Annual Conference of the American Physical Therapy Association, Cincinnati, USA.
Information: APTA, 111 N. Fairfax St., Alexandria, VA 22314, USA.

15–19 June, 1993
Annual Meeting of the International Society for the Study of the Lumbar Spine, Marseille, France.
Information: Dr. J. Weinstein, Room A309, Sunnybrook Medical Centre, 2075 Bayview Ave., Toronto, Ontario M4N 3M5, Canada.

1–4 July, 1993
2nd International Symposium on 3-D Analysis of Human Movement, Poitiers, France.
Information: Dr. P. Allard, Centre De Recherche, Hopital Sainte-Justine, 3175 Cote Ste-Catherine, Montreal H3T 1C5, Canada.

4–8 July, 1993
14th Congress of the International Society of Biomechanics, Paris, France.
Information: Convergences ISB '93, 120 Gambetta Ave., 75020 Paris, France.

8–9 July, 1993
Symposium on the Biomechanics of Joints and Joint Replacement, Leeds, England.
Information: Dr. J. Fisher, Dept. of Mechanical Engineering, University of Leeds, Leeds LS2 9JT England.

2–4 August, 1993
Eurable, 1st European Conference of People with a Disability, Maastricht, The Netherlands.
Information: Secretariat, Eurable, PO Box 3028, 2480 AA Woubrugge, The Netherlands.

5–7 August, 1993
81st Annual Meeting of the American Podiatric Medical Association, New Orleans, USA.
Information: APMA Annual Meeting Dept, 9312 Old Georgetown Rd, Bethseda, Maryland 20814–1621, USA.

16–20 August, 1993
Myo-Electric Control Symposium, New Brunswick, Canada.
Information: MEC '93, Institute of Biomedical Engineering, University of New Brunswick, PO Box 4400, Fredericton, NB, Canada E3B 5A3.

22–26 August, 1993
The Ljubljana FES Conference Ljubljana, Slovenia.
Information: Cankarjev Dom, Cultural and Congress Centre, Presernova 10,61000 Ljubljana, Slovenia.

24–26 August, 1993
SICOT Pre-Congress and 15th Annual Meeting of the Thai Orthopaedic Association, Bangkok, Thailand.
Information: Chusakdi Suwansirikul M.D., Thai Orthopaedic Association, 94/2 Supavdee Tower, 100 Soi Mitanant, Nakornchaisri Rd., Dusit, Bangkok 10300, Thailand.

28 August–3 September, 1993
19th World Congress of SICOT, Seoul, Korea.
Information: SICOT 93 Seoul Secretariat, c/o Korea Exhibition Centre, KWTC PO Box 4, Seoul 135-650, Korea.

13–15 September, 1993
Autumn Scientific Meeting of the British Orthopaedic Association, Torquay, England.
Information: BOA, 35-43 Lincoln's Inn Fields, London WC2A 3PN, England.

13–15 September, 1993
Annual Scientific Meeting of the Biological Engineering Society, Bath, England.
Information: Mrs. B. Freeman, BES, RCS, 35 Lincoln's Inn Fields, London, England.

15–17 September, 1993
Orthopaedica Belgica Annual Congress (in collaboration with the Belgian Ameroscopy Association on the Belguim Society for Foot Medicine and Surgery).
Information: Medicongress, Waalpool 28, B–9960 Assende, Belgium.

22–25 September, 1993
12th International Congress of Interbor, Lisbon, Portugal.
Information: Mundicongressos, Edif, Alcantara-Tejo, R. Maria Luisa Holstein, 15,1300 Lisbon, Portugal.

29 September–1 October, 1993
2nd International Conference on Computers in Biomedicine, Bath, England.
Information: Audrey Lampard, Wessex Institute of Technology, Ashurst Lodge, Ashurst, Southampton SO4 2AA, England.

12–16 October, 1993
Annual National Assembly of the American Orthotic and Prosthetic Association, Reno, USA.
Information: AOPA, 717 Pendleton St., Alexandria, VA 22314, USA.

17–22 October, 1993
10th International Conference of Neurological Surgery, Acapulco, Mexico.
Information: Dr. F. Rueda-Franco, Secretariat Office, PO Box 101-108, Col. Insurgentes Cuicuiclo Deleg. Coyoacan Mexico, DF-04530, Mexico.

23–29 October, 1993
1st North American Regional Conference of Rehabilitation International, Atlanta, USA.
Information: 1st North American Regional Conference of Rehabilitation International, c/o U.S. Council on International Rehabilitation, International Square, 1825 I Street N.W. Suite 400, Washington, D.C. 20006, USA.

31 October–5 November, 1993
Joint Meeting of the American Congress of Rehabilitation Medicine and American Academy of Physical Medicine and Rehabilitation, Miami, USA.
Information: AAPMR, 122 South Michigan Ave., Suite 1300, Chicago, IL 60603, USA.

8–11 November, 1993
International Workshop on Cerebral Palsy and Other Severe Disabilities, Frankfurt, Germany.
Information: Joanna Large, Workshop Secretariat, Community Publishing, Pamwell House, 160 Pennywell Rd., Bristol B55 0TX, England.

7-12 December, 1993
17th Annual Convention of the American Academy of Neurological and Orthopaedic Surgery, Las Vegas, USA.
Information: Dr. Michael R. Rask, 2320 Rancho Drive, Suite 108, Las Vegas, Nevada 89102-4592, USA.

1994

9–11 February, 1994
ISPO (UK) Annual Scientific Meeting, Blackpool, England.
Information: Mr. D. Simpson, ISPO Blackpool '94, NCTEPO, University of Strathclyde, 131 St. James' Rd. Glasgow G4 0LS, Scotland.

5–6 March, 1994
10th Annual Conference of the Association of Prosthetists and Orthotists, Liverpool, England.
Information: Mr. W. Dykes, APO Conference Co-ordinator, NCTEPO, University of Strathclyde, 131 St. James' Rd. Glasgow G4 0LS, Scotland.

9–16 April, 1994
7th World Congress of the International Rehabilitation Medicine Association, Washington, USA.
Information: Ms. D. Jones, 1333 Moursund A-221, Houston, Texas 77030, USA.

13–14 April, 1994
Combined Meeting of the British, Dutch and Scandinavian Orthopaedic Associations, London, England.
Information: BOA, 35-43 Lincoln's Inn Fields, London WC2A 3PN, England.

17–22 April, 1994
11th Congress of the World Federation of Occupational Therapists, London, England.
Information: British Association of Occupational Therapists, 6–8 Marshalsea Rd., London SE1 1HL, England.

5–8 July, 1994
Dundee '94—International Conference on Clinical Gait Analysis, Dundee, Scotland.
Information: Dundee '94 Secretariat, Dundee Limb Fitting Centre, 133 Queen St., Broughty Ferry, Dundee DD5 1AG, Scotland.

20–26 August, 1994
17th International Conference on Medical and Biomedical Engineering, Rio de Janeiro, Brazil.
Information: Dr. C. G. Orton, International Organization for Medical Physics, Gershenson Radiation Oncology Center, Harper-Grace Hospitals, 3990 John R., Detroit, MI 48201, USA.

4–9 September, 1994
6th European Regional Conference of Rehabilitation International, Budapest, Hungary.
Information: Rehabilitation Secretariat, ISM Ltd., The Old Vicarage, Haley Hill, Halifax HX3 6DR, England.

1995

2–7 April, 1995
Eighth World Congress of the International Society for Prosthetics and Orthotics, Melbourne, Australia.
Information: Congress Secretariat, Eighth World Congress ISPO, PO Box 29, Parkville 3052, Victoria, Australia.

Graph-Lite Pyramid Connector System

TPS-F4

TPS-M4

TPS-M4S

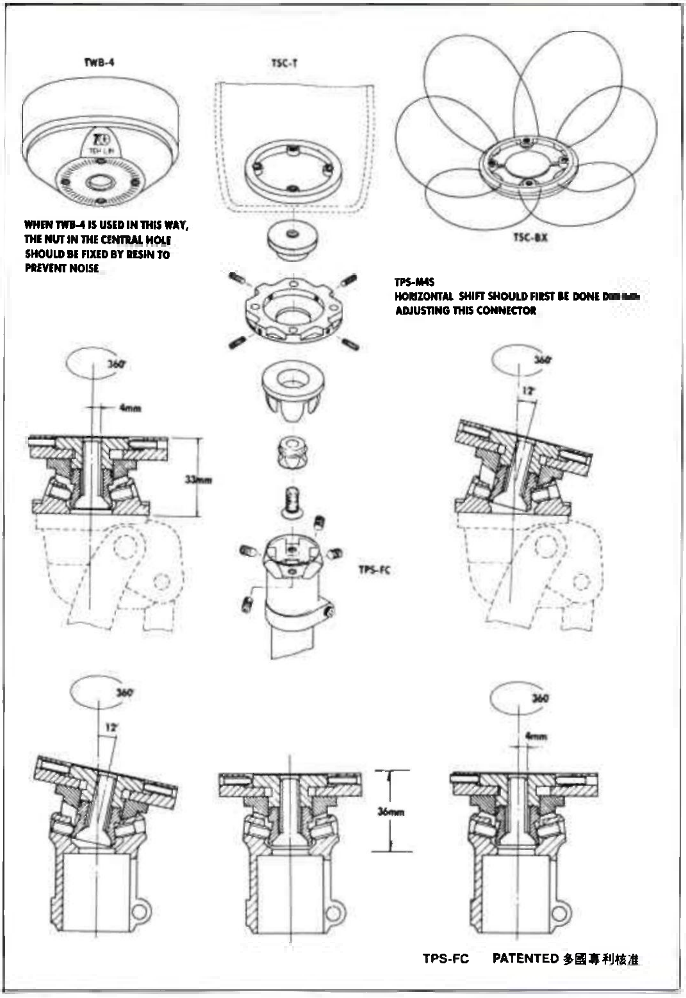
TPS-FC

TPS-M

TGA-PO

TPS-FA

TGA-PM



TEH LIN PROSTHETIC & ORTHOPAEDIC INC.

Representatives 236-238, Sec.3, Hoping W.Rd., Taipei, Taiwan, FAX: (02) 308-7890 TEL: 302-3071

USA/CANADA
DAW INDUSTRIES, INC

GERMANY/NETHERLANDS
WILH. JUL. TEUFEL GMBH

SWITZERLAND
FREY ORTHOPADIE-BEDARF

AUSTRIA
ELEKTROMEDIZIN VERTRIEBS-GMBH

BELGIUM/LUXEMBURG
ORTEC b.v.b.a. TRCHNISCHE ORTHOPEDIE

ITALY
SLO s.r.l.

FRANCE
DAW S.A.

UNITED KINGDOM
TECHNOLOGY WORLD INC (UK)

GREECE
ORTHOPEDIC INSTITUTION "REGINA"
M. TZANGARAKIS & CO.

TURKIYE
HC MEDICAL PRODUCTS, INC

HUNGARY
MS SANITATSHAUS HANDEL AKTIENGESELLSCHAFT

MEXICO
MARTI ORTOPEDIA, S.A. DE C.V.

ARGENTINA/PARAGUAY/URUGUAY
ORTOPEDIA ALLIMANA S.A.

BRAZIL
CENTRO DE PRESERVACAO E ADAPTACAO DE MEMBROS DE

AUSTRALIA
GEORGE W. MASSON & SONS PTY LTD.

SOUTH AFRICA
NATIONAL MEDICAL & ORTHOPAEDIC SUPPLIERS CC

MALAYSIA
TEH LIN PROSTHETIC & ORTHOPAEDIC CO. SDN. BHD.

THAILAND
TEH LIN PROSTHETIC & ORTHOTIC CO., LTD

INDONESIA
INDO TEH LIN PROSTEHTIC & ORTHOPAEDIC INC

KOREA
SHIN SUNG IND. CO.

JAPAN
TAZAWA MFG CO

HONG KONG
QUINCY PROSTHETIC & ORTHOTIC APPLIANCES CO

PHILIPPINES
TRUEFROM REHABILITATION AIDS CENTER

SRI LANKA
FERN ENTERPRISES (PVT) LTD

INDIA
INTERNATIONAL SURGICAL WORKS

SAUDI ARABIA
ALI RAYES TRADING EST.

ISRAEL
ESHED ADVANCED ORTHOPAEDICS LTD

PAKISTAN
SAMEER ENTERPRISES

IRAN
GLOBAL OVERSEAS CO. LTD