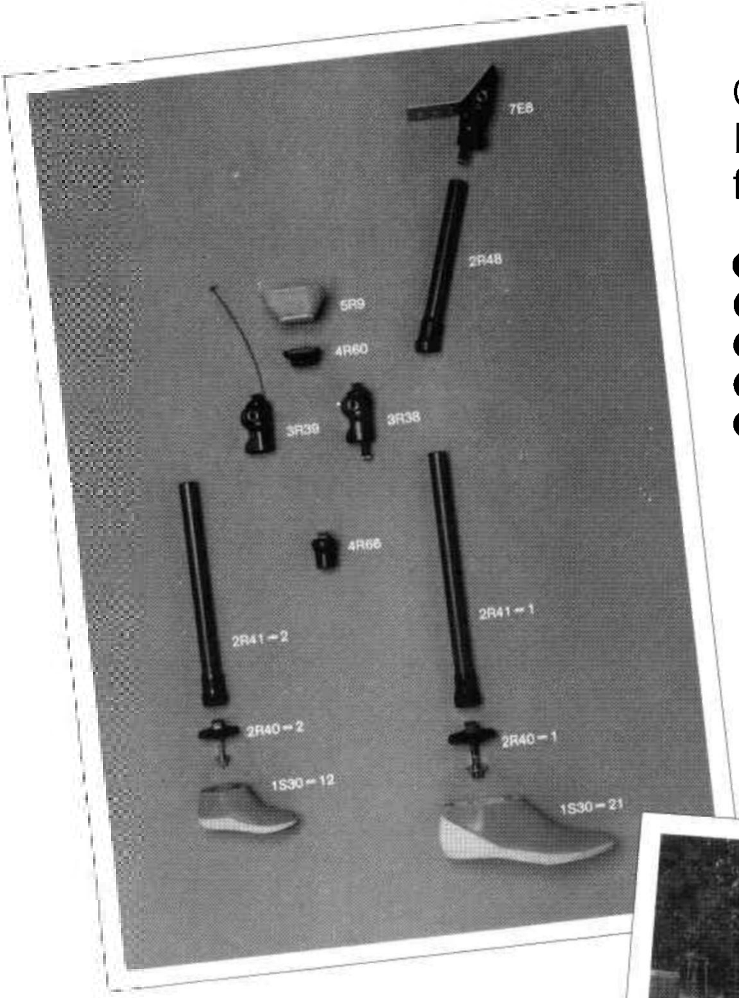
The Journal of the International Society for Prosthetics and Orthotics

Prosthetics and Orthotics International

April 1990, Vol. 14, No. 1

Orthopaedic Technique for Children



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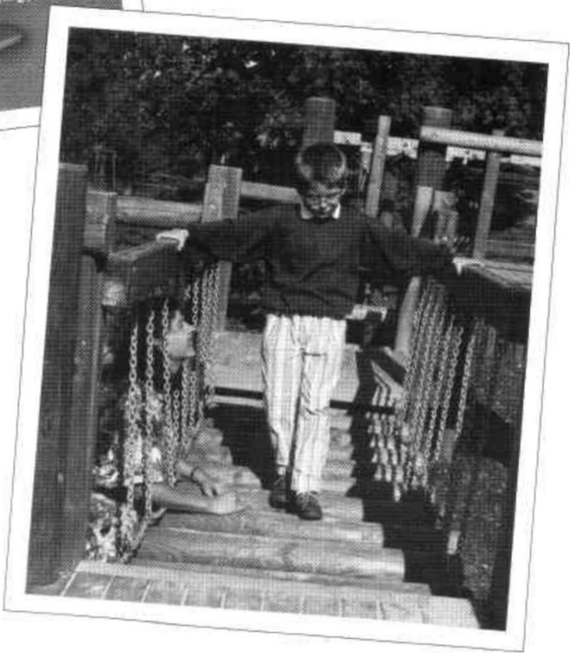

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Prosthetics and Orthotics International

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The Journal of the International Society for Prosthetics and Orthotics

April 1990, Vol. 14, No. 1

Contents

ISPO

Editorial

As has become our tradition the April 1990 issue of Prosthetics and Orthotics International contains the financial statement for the previous fiscal year.

ISPO has devoted much interest and effort to the 1989 World Congress in Kobe, Japan under the succesful and competent guidance of Secretary General, Seishi Sawamura. As this marvellous event took place in November 1989, it is technically not possible to include the financial results of the congress in the 1989 accounts.

For the third continuous year the primary result was negative, meaning that the wheels of the Society could not run without the generous contributions from our sponsors, SAHVA and the War Amputations of Canada. As in 1987 the deficit could not be covered by the interest from our capital, meaning that we had to reduce our reserves, although marginally. Because of the strains on the international monetary market we have also in 1989 experienced a minor decrease in the market value of our securities, but we continue our attempts to assure the Society the highest possible, but secure, investment income. During the later months of the fiscal year this goal has been achieved by placing a large amount of cash on a month-to-month bank account with a high interest rate. Over the past triennium the total assets have been unchanged in spite of the turmoil in the monetary market and a fairly large investment in international education programmes.

As already announced in last year's Editorial to the accounting it was found necessary to increase the fee for 1990 to 450 DKK (175 DKK for developing countries), but we certainly hope that it will be possible to keep the fee constant for the remainder of the triennium.

This year's deficit can in part be explained by our continuing engagement in international work together with other bodies, especially in Africa, with the purpose of securing the quality of the prosthetist-orthotist edcuation in the developing world. Furthermore the last year of a triennium always accrues increased expenses related to printing and mailing of the Membership Directory, and to increased Executive Board activity.

The financial situation of the Society is still sound and allows for further engagement in course and conference programmes and for international work, some of which however is undertaken partially under the ISPO banner without being displayed in the accounts, because other sources of funding have been found.

The administration expenses of the Society have increased by less than five per cent over the past year.

It should be remembered, that the basis of our sound economics and international engagements, is the office facilities provided free of charge by SAHVA, and the hard work done by volunteers as task officers and at the National Centre for Training and Education in Prosthetics and Orthotics in Glasgow. Our gratitude is expressed to these loyal collaborators, who gives us the required support to permit our involvement on equal terms with larger and state subsidized international bodies.

J. Steen Jensen
Honorary Treasurer

Prosthetics and Orthotics International, 1990, 14, 2–5

ISPO Statement of Accounts, 1989

Auditors' Report

We have audited the financial statements for the year ended December 31, 1989.

The audit has been performed in accordance with approved auditing standards and has included such procedures as we considered necessary. We have satisfied ourselves that the assets shown in the financial statements exist, have been fairly valued and are beneficially owned by the association and that all known material liabilities on the balance sheet data have been included.

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 association's affairs on December 31, 1989 and of the profit for the year then ended.

Copenhagen, March 23, 1990
Schøbel & Marholt
Søren Wonsild Glud
State Authorized Public Accountant

Accounting Policies
Securities
Bonds and shares are valued at lower cost on market.

Office equipment
Computer and office furniture have been valued at cost less depreciation, computed linearly over 5 years.

Income Statement for the Year 1989

		1989	1988
SUMMARY			
Society membership fees		899.545	758.759
Sponsorship (note 2)		135.444	107.064
Conferences, courses etc. (note 3)		(135.565)	(50.019)
World Congress 1989. Kobe Japan (note 4)		(117.523)	—
Professional register (note 5)		(8.681)	—
Prosthetics and Orthotics Int. (note 6)		(65.242)	(78.214)
Publications (note 7)		(106.018)	12.212
		601.960	749.802
Administration expenses (note 1)		(998.111)	(952.306)
Primary result		(396.151)	(202.504)
Interest (note 8)		217.308	223.471
Dividend (note 8)		74.628	53.863
Maturity yield (note 8)		2.792	153.022
Change in market value of securities		(85.304)	203.904
Result for the year	DKK	(186.727)	431.756

Balance sheet as of December 31, 1989

ASSETS			
Cash		1.519.649	189.177
Accounts due			
Accrued interest		48.855	48.467
Advertising receivable		27.725	16.910
Advance funding of World Congress 1980		87.437	87.437
		164.017	152.814
Securities (note 10)		1.999.849	3.307.660
Office equipment (note 9)		48.626	72.941
Total Assets	DKK	3.732.141	3.722.592
LIABILITIES AND CAPITAL			
		1989	**1988**
Liabilities			
Short-term liabilities			
Accrued expenses		85.110	37.334
Accrued printing expenses		111.679	–
Prepaid membership fees		40.000	4.700
Prepaid advertising income		5.285	5.359
Prepaid subscription income		93.665	92.070
		335.739	139.463
Provisions			
Provisions World Congress 1980		87.437	87.437
Capital			
Capital January 1, 1989		3.495.692	3.063.936
Result for the year		(186.727)	431.756
Capital December 31, 1989		3.308.965	3.495.692
Total Liabilities and Capital	DKK	3.732.141	3.722.592

Contingent liabilities (note 11)

Notes to the Financial Statements

		1989	1988
1. ADMINISTRATION EXPENSES			
Expenses — Executive Board and officers:			
Travel and hotel		(392.056)	(283.307)
Meeting expenses		(21.047)	(32.562)
Meeting with other organizations		(29.745)	(149.785)
		(442.848)	(465.654)
Secretariat, Copenhagen:			
Staff salaries		(277.728)	(277.254)
Labour tax		(16.648)	(7.181)
Data service		(32.447)	(25.242)
Business meeting expenses		(48.560)	(27.105)
Bank expenses		(8.066)	(5.734)
Postage		(77.302)	(24.154)
Telephone		(5.200)	(5.219)
Stationery printing		(22.642)	(24.979)
Office suppliers		(5.376)	(2.098)
Accountant		(28.400)	(26.100)
Lawyer		—	(10.648)
Literature		—	(11.683)
Sundries		(8.581)	(14.942)
Depreciation		(24.313)	(24.313)
		(555.263)	(486.652)
Administration expenses	DKK	(998.111)	(952.306)
2. SPONSORSHIP			
Contribution from the War Amputations of Canada		35.444	32.064
Contribution from SAHVA		100.000	75.000
	DKK	135.444	107.064
3. CONFERENCES, COURSES, etc.			
Strathclyde 1987		—	8.474
Heidelberg 1988		—	(6.066)
Miami AK-socket Workshop		—	(1.950)
Amputation Surgery, prep. meeting		(20.487)	—
CAD/CAM — Seattle		(915)	(46.000)
Education Committee		(24.868)	—
WOC		(3.916)	(4.477)
WHO — Geneva		(5.843)	—
WHO — Senegal		(21.251)	—
American Academy Prosth. Orth.		(14.646)	—
Zimbabwe ARI		(43.639)	—
	DKK	(135.565)	(50.019)
4. WORLD CONGRESS 1989, KOBE JAPAN			
Knud Jansen Lecturer		(25.000)	—
Secretariat expenses		(92.523)	—
	DKK	(117.523)	—
5. PROFESSIONAL REGISTER			
Professional register	DKK	(8.681)	—

6. PROSTHETICS AND ORTHOTICS INTERNATIONAL

Advertising		137.043	174.436
Subscriptions		169.227	108.287
		306.270	282.723
Printing & mailing		(331.216)	(319.843)
Production editor		(12.172)	(18.449)
Meeting expenses		(28.124)	(22.645)
		(371.512)	(360.937)
	DKK	(65.242)	(78.214)

7. PUBLICATIONS

Membership directory		(97.216)	—
Reports Strathclyde		(19.622)	—
Book sale		10.820	12.212
	DKK	(106.018)	12.212

8. INVESTMENT INCOME

Maturity yield		2.792	153.022
Interest, bonds (note 10)		191.475	224.152
Interest, bank		25.833	11.458
		217.308	235.610
Dividends (note 10)		74.628	53.863
	DKK	294.728	442.495

9. 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, 1989		(48.626)	(24.313)
Depreciation for the year		(24.313)	(24.313)
		(72.939)	(48.626)
	DKK	48.628	72.941

10. SECURITIES

		Nominal Value	Rate 31/12/89	Original cost	Value 31/12/89	Interest/ Dividends
Bonds						
9% KD, 22.S.2007		2.113.000	95,60	1.969.089	1.977.768	191.475
	DKK	2.113.000		1.969.089	1.977.768	191.475
Shares						
Dividend sold shares						73.128
Handelsbanken		10.000	264,98	26.498	28.900	1.500
Handelsbanken		1.700	289,00	4.262	4.913	—
	DKK	11.700		30.760	33.813	74.628
	DKK	2.124.700		1.999.849	2.011.581	266.103

11. CONTINGENT LIABILITY

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, 6–8

Executive Board Meetings

10th November 1989 and 18th November 1989
Kobe, Japan

Two Executive Board Meetings were held at the time of the World Congress in Kobe. The first was the final meeting of the out-going Executive Board and the second was the inaugural meeting of the in-coming Executive Board.

The following paragraphs summarize the major discussions and conclusions of these meetings. They are based on the approved minute of the first meeting and the unconfirmed minute of the second meeting.

Standing Committee and Task Officer Reports

The Chairman of the Finance Committee reported on the Society's finances for the past triennium, indicating that the Society was in a sound financial position. He indicated that income from membership, subscribers and sponsors is insufficient to cover the present activities of the Society and care must be taken that as far as possible, only the profits from the Society's investments should be used to cover these expenses. He recommended that the capital should remain untouched. The Honorary Treasurer presented the proposed Budget for 1990 which was approved by the Executive Board.

Membership of the Society had increased by over 300 during the past year, giving a present total of 2,272. Subscribers to "Prosthetics and Orthotics International" had increased slightly over the year and now totalled 368. A new National Member Society has been formed in Pakistan and Taiwan and Indonesia were showing interest in establishing one in their country. Interest has also been shown in establishing Regional Societies of ISPO in Central America and Africa and this is at present being pursued.

The Society has been involved in a number of activities related to education over the past triennium and prominent among these has been a Workshop organised by the Society in 1987 on Upgrading in Prosthetics and Orthotics for Technicians from Developing Countries Trained on Short Courses. In addition, the Society has been involved in a number of seminars on education and training in the developing world organised by other agencies. In 1987 the International Labour Office/African Rehabilitation Institute seminar in Cairo, Egypt; 1988, the United Nations seminar in Moshi, Tanzania; 1989, the African Rehabilitation Institute seminar in Bulawayo, Zimbabwe and 1989, the World Health Organization seminar in Dakar, Senegal.

The Chairman of the Publications Committee presented the activities over the past triennium during which the following reports have been published or are nearing completion: Training and Education in Prosthetics and Orthotics for Developing Countries, Jonkoping, Sweden; Upgrading in Prosthetics and Orthotics for Technicians from Developing Countries Trained on Short Courses, Glasgow, Scotland; International Workshop on Above-knee Fitting and Alignment, Miami, USA/ Workshop on Teaching Material for Above-knee Socket Variants, Chicago, USA; Prosthetics and Orthotics Education, Toronto, Canada; Amputation Surgery and Lower Limb Prosthetics, Dundee, Scotland. "Prosthetics and Orthotics International" continues to flourish and now includes a regular International Newsletter section. The Publications Committee are anticipating publication of a new ISPO brochure describing the activities of the Society in order to promote membership and understanding of the workings of the Society. This matter is now in hand. The new Directory of Officers and Members of the Society has now been published and distributed to all members.

The Executive Board agreed that an Amputation Consensus Workshop would be held at the beginning of October, 1990 in Glasgow. The purpose of this Workshop is to provide information for the development of instructional material to train surgeons in amputation techniques which provide optimum results. It is the intention that this Workshop should be followed up by a course(s) on upgrading of amputation surgery and prosthetics about six months later. The venue for this course(s) is at present being investigated.

It has not been possible to publish the Proceedings of the Symposium on the Limb Deficient Child held in Heidelberg in September, 1988. It is, however, the intention to publish a special edition of "Prosthetics and Orthotics International" on this topic.

Work on the Professional Register continues and it is hoped that the Register will be available in the near future. A second mailing will be carried out to ensure that as many members as possible are included.

International Organizations

The Executive Board continues to have reciprocal representation on the Board of Directors of INTERBOR and close collaboration continues between the two Societies especially in relation to each others Congresses and in particular ISPO's VI World Congress in Kobe and INTERBOR's XI World Congress in Rome. The ISPO/INTERBOR Joint European Education Committee continues with its efforts to influence the European Community in its regulations on reciprocal recognition of prosthetics and orthotics qualifications and to press for a Special Directive for the profession. John Hughes represents ISPO on the INTERBOR Board and the President, John Hughes, Sepp Heim and the Honorary Secretary represent the Society on the Joint Education Committee.

The Society continues in its efforts to establish official working relations with the World Health Organization (WHO).

The Society has agreed to organise a session on Prosthetics and Orthotics at Rehabilitation International's V European Regional Conference to be held in Dublin in May 1990.

It was agreed that Acke Jernberger would represent the Society on the Board of the Internationaler Verband der Orthopadie-Schutechniker (IVO). It was further agreed that Thamrongrat Keokarn would represent the Society on the Executive Board of the World Orthopaedic Concern (WOC) and that George Murdoch would represent the Society's interests at the mini Executive meetings of WOC held in London.

The Society expressed its concern with regard to the two orthopaedic technicians from the International Committee of the Red Cross (ICRC) who had been taken hostage in Beirut. It was agreed that the Society should contact ICRC to find out if there was any way in which the Society could be of assistance in this matter.

Congresses

The VI World Congress held in Kobe had been very successful with an excellent Scientific Programme and a most memorable Social Programme. A grand total of 1,231 persons attended the Congress including accompanying persons, students and one-day participants. The President, on behalf of the Society, thanked Seishi Sawamura and all his colleagues for organizing such a successful Congress.

The VII World Congress shall be held in Chicago, USA on June 28—July 3 1992. Detailed planning for this Congress is now underway. Dudley Childress is the Secretary General for the event.

Two invitations have been received to host the VIII World Congress in 1995: one from Australia and the other from the Netherlands. It is hoped to reach a decision as to the venue for this Congress in the near future.

Conferences and Meetings

The President reported on the Society's involvement in the European Conference on Rehabilitation Technology to be held in the Netherlands on 5-8 November, 1990. The President represents the Society on the organizing committee and John Hughes represents the Society on the Scientific Committee thus ensuring ISPO's representation on all sections of the Conference related to prosthetics and orthotics.

It was hoped to hold a meeting on the Shoe and Foot in Jonkoping, Sweden in June 1991. A proposal regarding the programme and budget will be put to the Executive Board in due course.

International Committee Meeting

The major outcome of the International Committee Meeting held prior to the Congress, was the establishment of a joint International Committee/Executive Board Working Group, under the Chairmanship of John Hughes, to examine the following:

a) ways in which there could be greater involvement of National Member Societies in the decision making process
b) the provision of better information to National Member Societies and the membership at large about international activities
c) the system of election of the Executive Board.

The representatives of the International Committee are: H. Arendzen (Netherlands), D. Condie (UK) and John Edelstein (USA). The Executive Board representatives are: the President, John Hughes and Sepp Heim.

Fellowships

Fellowships of the Society have been awarded to: Anders Starkhammer (Sweden) and Magnus Wall (Sweden).

Norman A. Jacobs
Honorary Secretary

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 to 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.

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

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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.

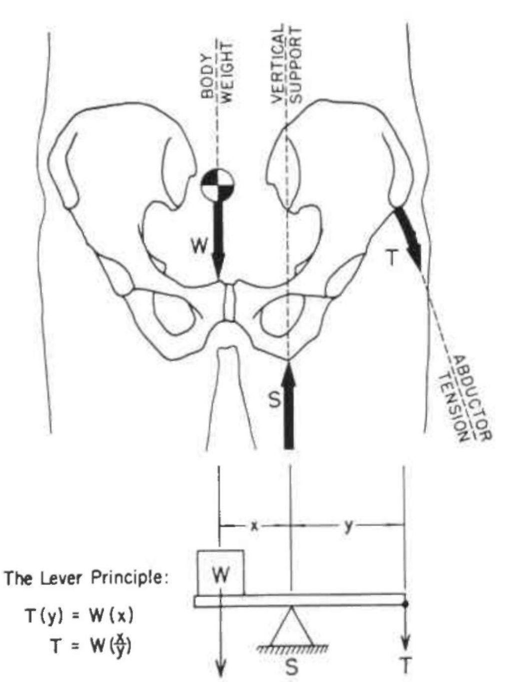

Fig.1. Moments acting about the support point in the frontal plane in lateral stabilization of the pelvis (Radcliffe, 1955).

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.
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).

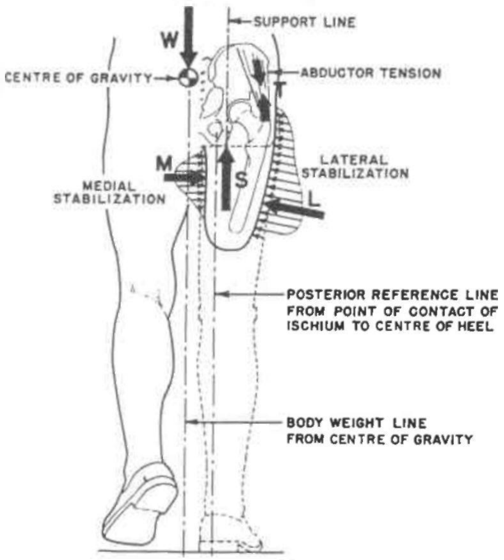

Fig. 2. Lateral stabilization of the pelvis (Radcliffe, 1970).

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

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.

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

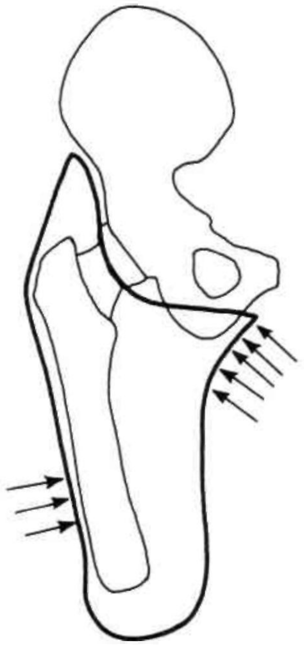

Fig. 4. Medial forces borne by bones of the pelvis and soft tissues, the principle of ischial containment.

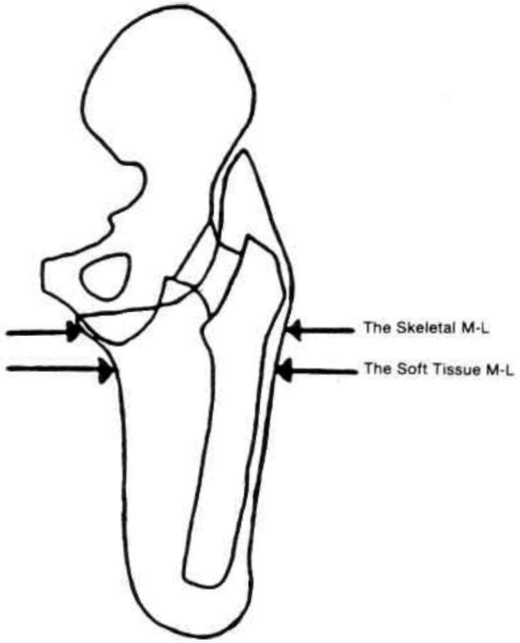

Fig. 5. Skeletal and soft tissue M-L dimensions of the Ischial Containment Socket (Hoyt et al, 1985).

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

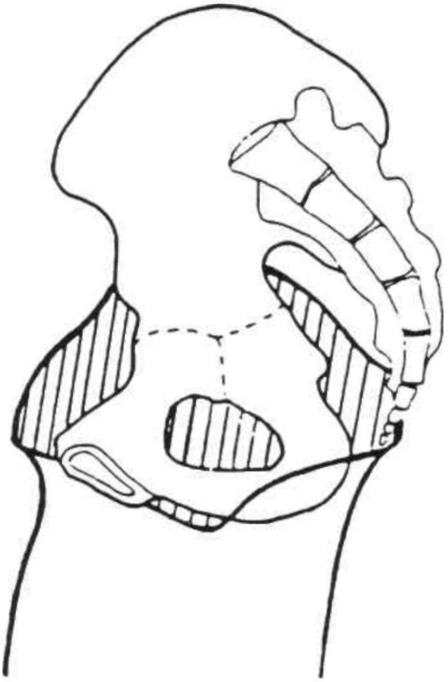

Fig. 6. Medial view of the Ischial Containment Socket in the sagittal plane showing relationship between the proximal edge of the medial wall and the bones of the pelvis (Hoyt et al, 1985).

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.

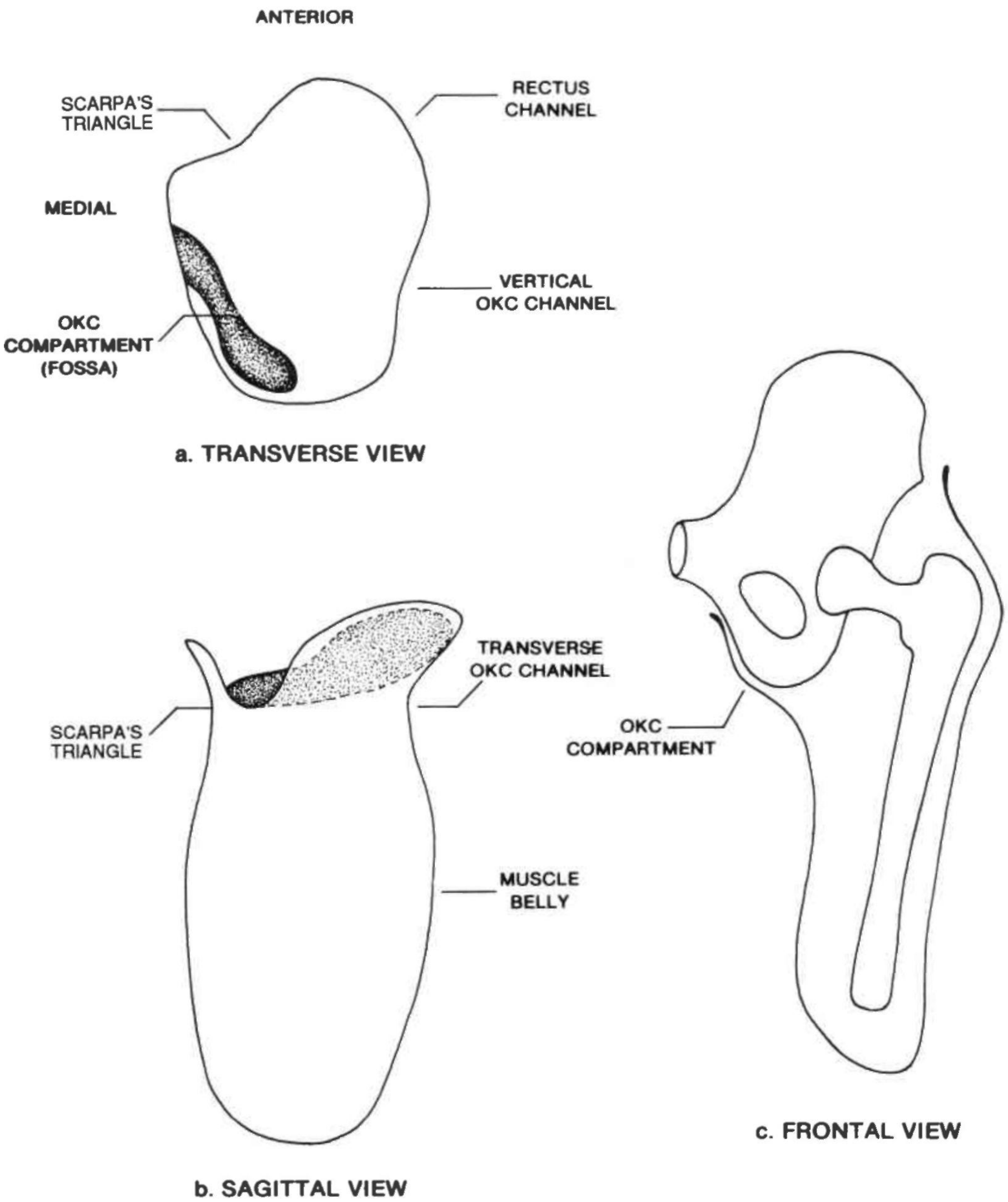

Fig. 7. The SCAT-CAM socket as described by Sabolich. Redrawn from an original by him to clarify some points. According to him some contours are exaggerated for emphasis, but not by much.

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

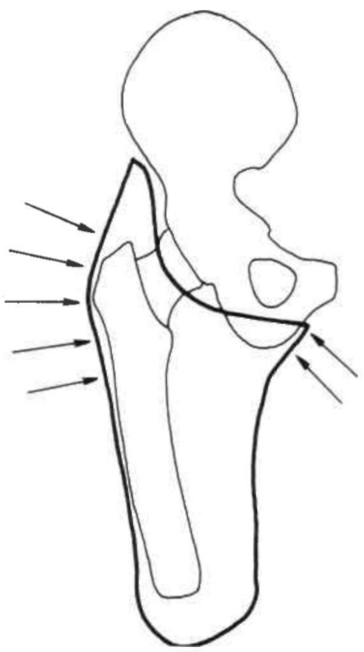

Fig. 8. Counterpressure generated by the lateral wall.

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

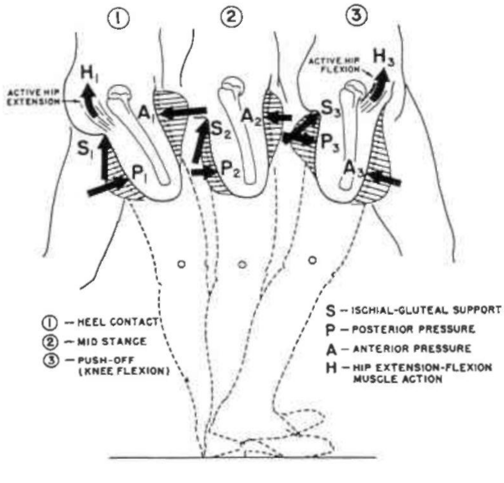

Fig. 9. Force patterns generated in the sagittal plane during gait (Radcliffe, 1970).

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 commun-ication 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

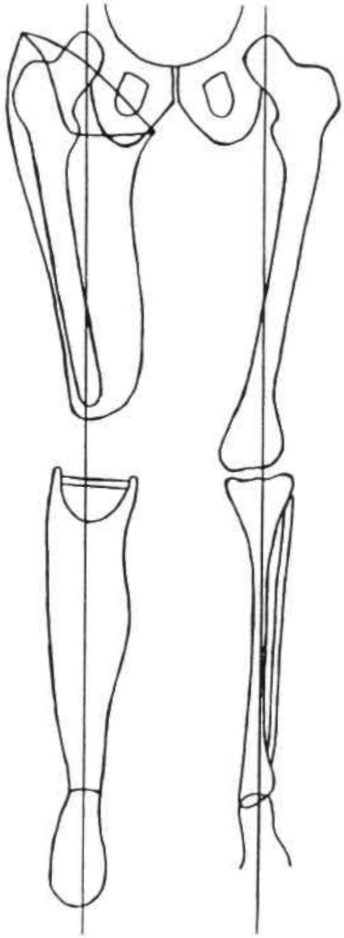

Fig. 10. Long's Line from: fabricating the Long's line above-knee prosthesis. (Long, 1985.)

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 this 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.

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

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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.

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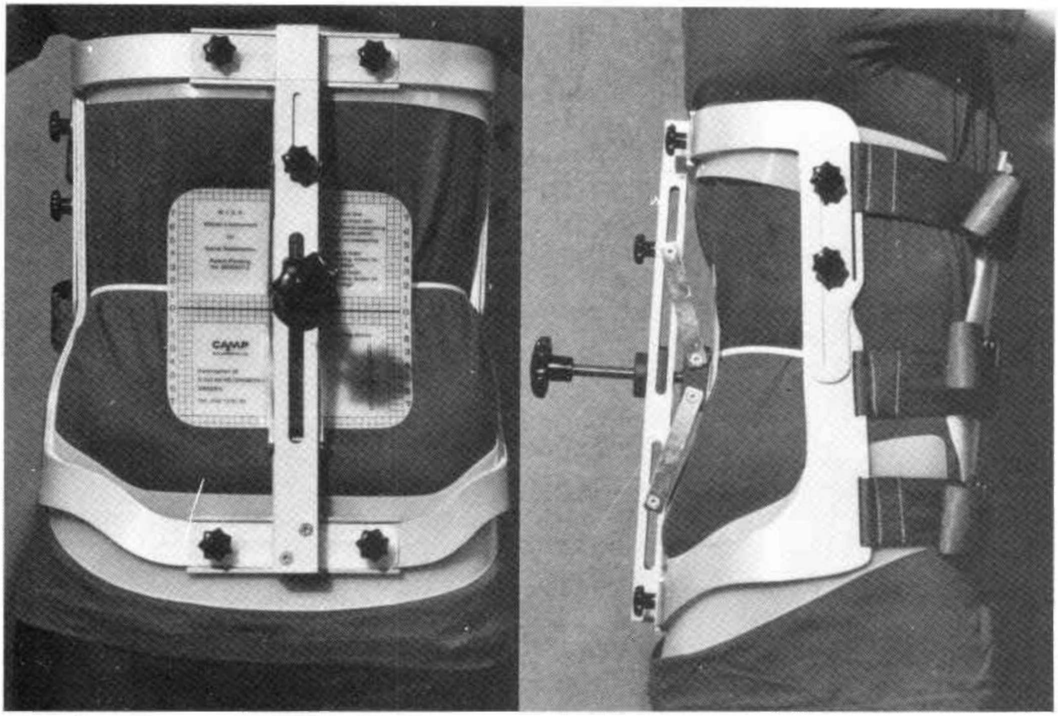

Fig. 1. Test instrument. Left, posterior view. Right, lateral view.

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.

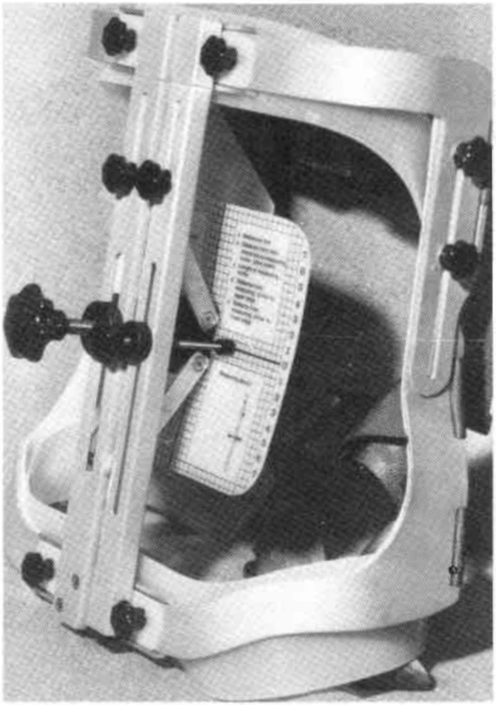

Fig. 2. The measuring device fixed on the test instrument.

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 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

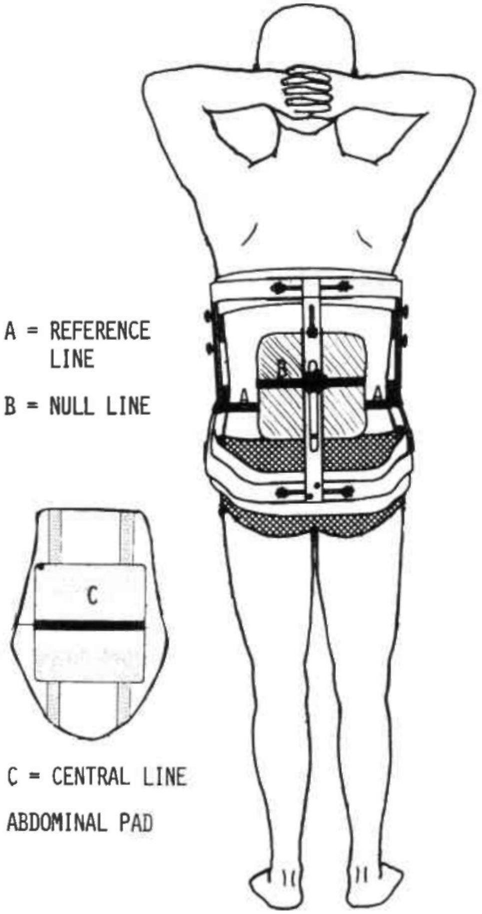

Fig. 3. Reference lines to allow the information observed in the test instrument to be transferred to the brace to be manufactured.

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

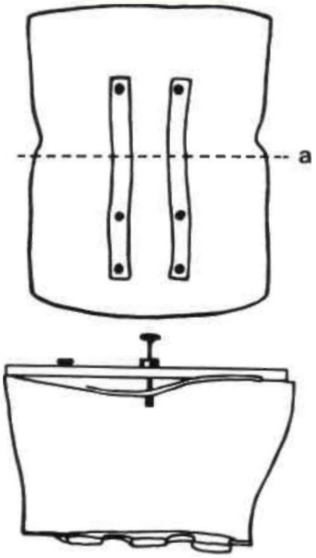

Fig. 4. Top, marking of the reference line (a) on the prefabricated module of a rigid brace. A hole is drilled at the level of the null line. Bottom, measuring device is placed on the module and the screw passes through the hole at the level of the null line.

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 be repeated twice or even three times.

Patient tests

Between 1986-87, 88 consecutive cases which had been referred to the Spinal Unit in Malmö 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 (Willner, 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.

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

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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-Møller 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 et al. 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-Sørensen 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 (Table1). They were all fully rehabilitated and trained to use long leg braces and forearm crutches. Participant No. 1

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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-Møller 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

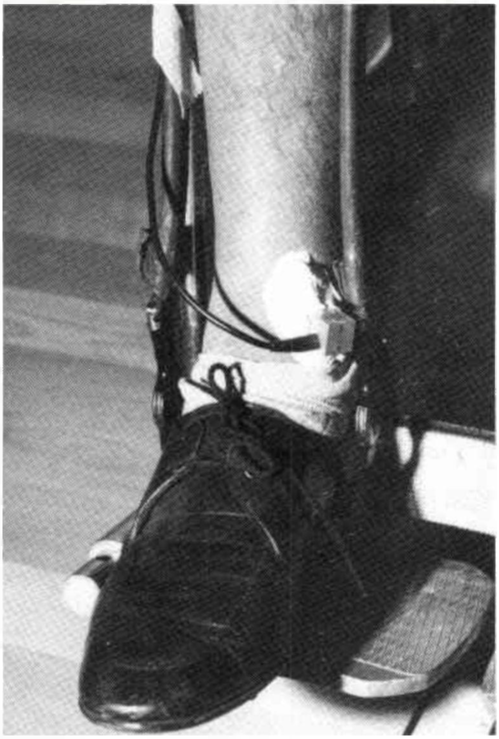

Fig. 1. Accelerometer fixed in a plaster holder and taped to the medial malleolus of the right leg, to create bony contact.

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

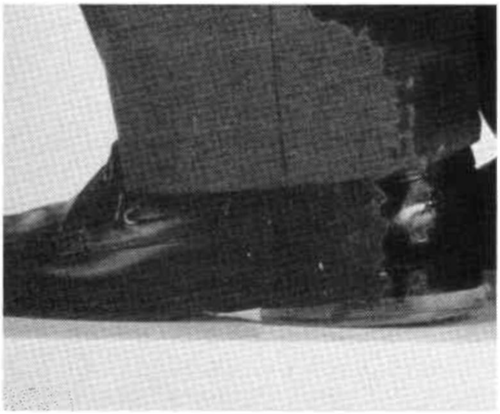

Fig. 3. Shoe with 10mm ethyl vinyl acetate sandwiched into the heel.

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

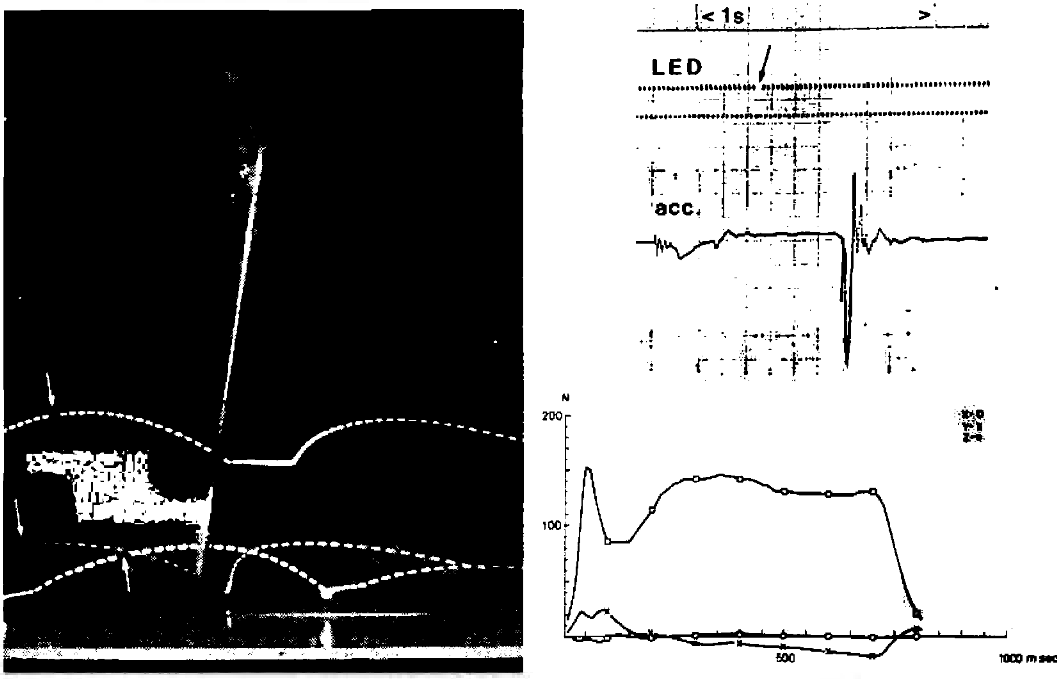

Fig. 4. Left, still photo of paraplegic walking on a walk way with built in force-platform. Light emitting diodes from three light tracks showing the trajectory of the shank, the heel, and the forefoot. Arrows indicate the missing signal and thereby the time link between the three tracks. Top right, oscillogram showing the 50Hz signal for the light emitting diodes (LED) and the accelerometry (acc.) 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.

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

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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.

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

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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 the 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 (McIlwain 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 the 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 testing. 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 (McIlwain 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

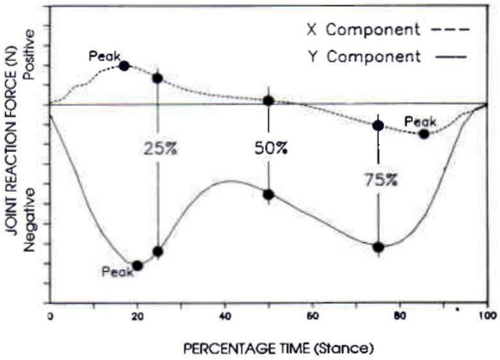

Fig. 1. Joint reaction force dependent variables.

coefficient, was used to quantify the degree of lower limb symmetry.

A 2 × 8 (subject × 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 (McIlwain 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

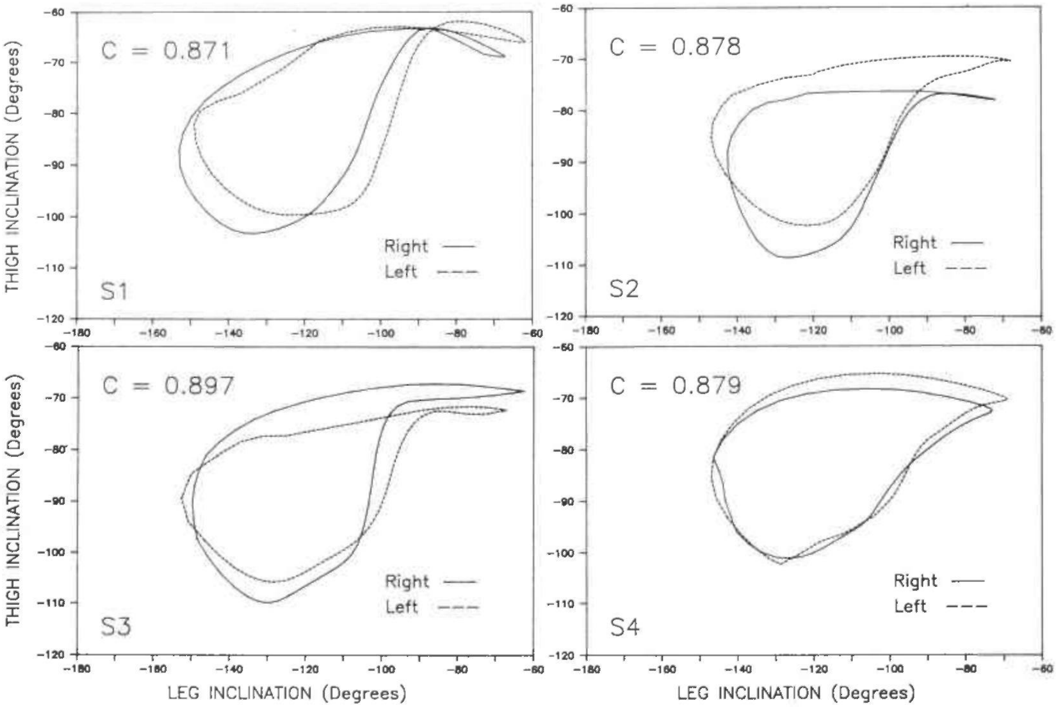

Fig. 2. Bilateral angle-angle plots of leg and thigh for one stride by non-amputees S1, S2, S3 and S4.

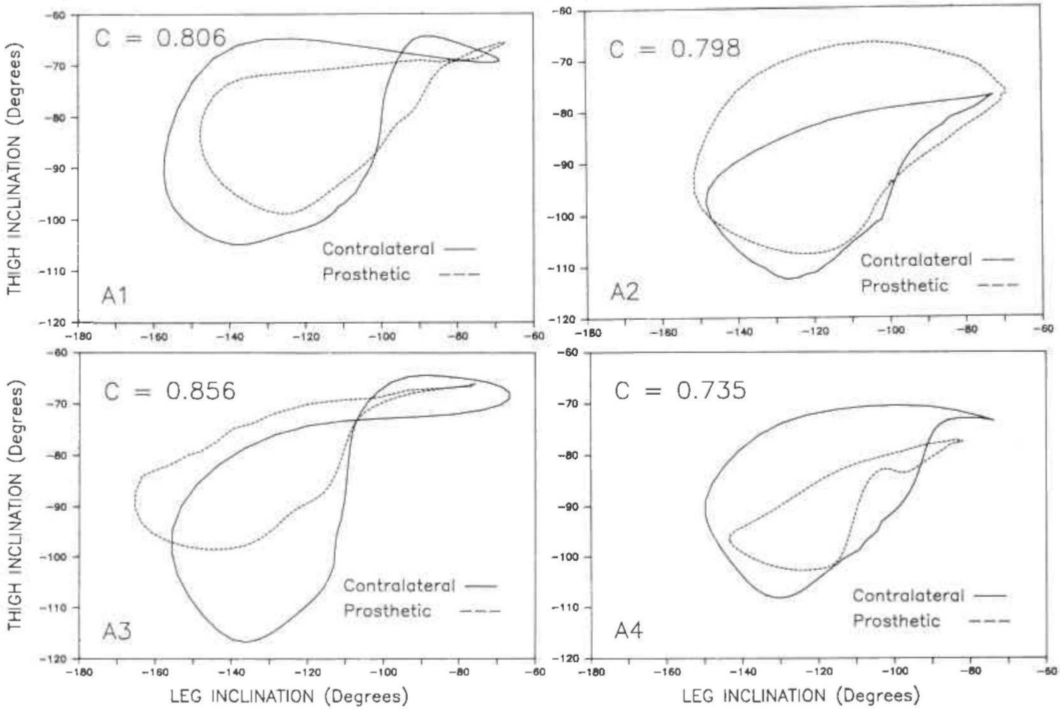

Fig. 3. Bilateral angle-angle plots of leg and thigh for one stride by amputees A1, A2, A3 and A4.

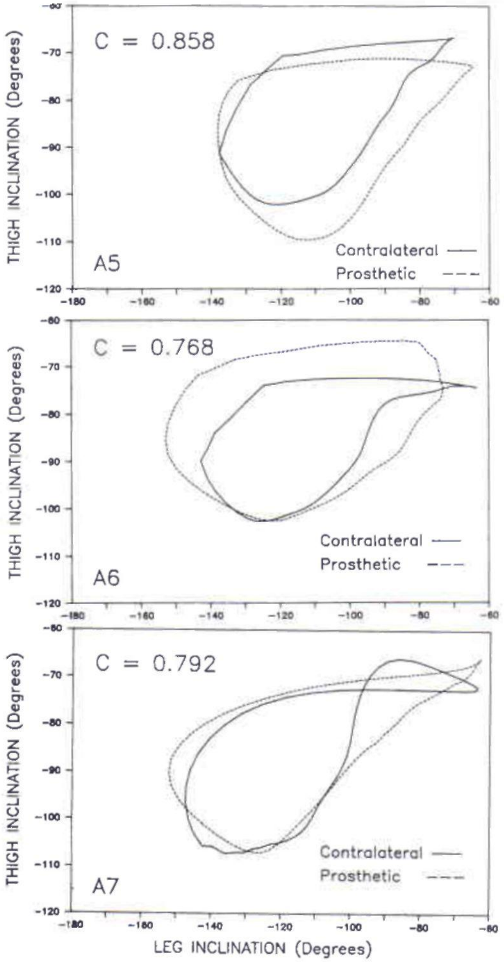

Fig. 4. Bilateral angle-angle plots of leg and thigh for one stride by amputees A5, A6 and A7.

with a value of 1.0 indicating perfect congruity and a value of 0.0 indicating absence of congruity between patterns, (McIlwain 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 S1 through S4 represent the non-amputee group and A1 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 × 8 (subject × 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.

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

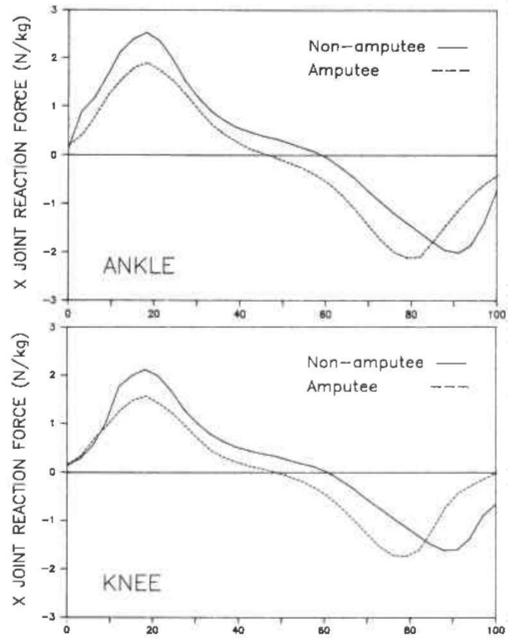

Fig. 5. Top, ankle horizontal joint reaction force, during stance, for an amputee and non-amputee subject. Bottom, knee horizontal joint reaction force, during stance, for an amputee and non-amputee subject.

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.

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Prosthetics and Orthotics International, 1990, 14, 43–44

British Design Award for Blatchford Endolite

Blatchfords Endolite modular system for artificial legs has gained the unique distinction of being the only medical product to win a 1990 British Design Award. This award, achieved in the centenary year of the company, is an appropriate acknowledgement of the high technology now being incorporated in artificial legs to produce varying functions suitable for amputees with differing activity levels in all age groups. Currently 65 per cent of amputees in the UK with modular lower limb prostheses are supplied with the Blatchford Endolite system through the Disablement Services Authority, the principal UK prescriber of the system.

Charles A. Blatchford founded the limb manufacturing business which has carried his name through successive family generations since 1890. At that time artificial legs were skilfully worked from wood but the material placed some limitation on function. His philosophy of "The patient comes first" continues to provide the Blatchford design team with the incentive to improve the function, the reliability and the comfort of lower limb prostheses using the latest advances in technology and materials.

The Blatchford Endolite system represents a major achievement in applying the technology and materials developed initially for space exploration. With their down-to-earth approach the Blatchford design team, led by the late Brian Blatchford MBE and Technical Director, John Shorter, utilised the strength and light weight of carbon fibre composites in the development of Endolite, the first production artificial leg in the world to use this exotic material. The system is designed to be equally suitable for enfeebled geriatric patients as well as for young and active amputees. This all-British development by a family-run business was achieved in the face of authoritative advice from the US that carbon fibre composites, while ideal for artificial legs, would prove too difficult to fabricate. Subsequently the US has become the second largest market for Endolite.

Working to weight limits set by clinicians at an international meeting, the Blatchford team at the company's research and development unit in Basingstoke produced complete prostheses weighing 1kg for below knee (BK) and 2kg for above knee (AK) and 3kg for through-the-hip level. The light weight of these prostheses has been achieved by using carbon fibre for the major load bearing component parts to produce a durable structure within a cosmetic foam fairing and outer "skin" which is difficult to detect from a natural leg. Having achieved satisfactory results for the shin structure, a new flexible foot was developed to simulate the "spring" of the natural foot. To complete the system a flexible ankle joint, known as Multiflex, was designed to enable the patient to walk with confidence on uneven surfaces. This mechanism, together with a range of 15 different knee modules including the Endolite Stabilised Knee (ESK) which locks when weight is applied and releases smoothly when weight is removed, has been designed for the entire spectrum of amputees.

The Blatchford Endolite system is the result of five years of intensive research and development including patient trials. Before the system could be accepted by the National Health Service, the individual modules as well as the complete system underwent rigorous approval testing in its laboratories and in prolonged field trials.

The modular design of the Endolite system enables the volume production of its component parts. This has produced savings in the unit cost of components and permits a high degree of interchangeability of parts to meet the changing needs of individual patients. The modular design also speeds assembly and makes adjustments on the patient easier and quicker. These factors together with the functional advantages of Endolite have made the system extremely cost effective. This has been acknowledged by health authorities in the UK and in many overseas countries. West Germany, for example is currently a major growth market. Also, the company is involved in equipping and staff training for a new artificual limb factory in the People's Republic of China.

Only properly trained prosthetists and technicians are allowed to prescribe and fit the Blatchford Endolite system. The company runs regular training and updating courses for all its agents and staff to ensure that they are fully aware of the latest developments and practices. Valuable feedback for these courses and for the ongoing process of design improvement is provided by the recently opened Blatchford Prosthetic Research Centre at Basingstoke District Hospital. This provides a central retrieval and dissemination centre for all matters relating to prosthetics in general and the Blatchford Endolite system collated from international sources as well as the company's 13 limb fitting centres throughout the UK.

The 1990 British Design Award for the Blatchford Endolite system is the second award received from The Design Council by Chas. A. Blatchford & Sons Ltd. In 1976 the company won a Design Council Award and a Queen's Award for Technical Innovation. In the same year the late Brian Blatchford received The Duke of Edinburgh's Designers Prize.

Prosthetics and Orthotics International, 1990, 14, 45–46

International Newsletter
Spring 1990

The excitement of the World Congress in Kobe has inspired ISPO members worldwide to pursue a wide variety of professional endeavours.

Canadian National Society will host a one-day seminar immediately preceding the Canadian Association of Prosthetics and Orthotics Convention 1990 at the Winnipeg Convention Centre. The June 10, 1990 programme will focus on the application of CAD-CAM techniques to the field of prosthetics and orthotics. Speakers will offer guidance regarding which of the systems on the market one should buy. International authorities will address the application of CAD-CAM to all levels of prosthetics and orthotics. The War Amputations of Canada is the generous sponsor of the programme.

Soviet Union will coordinate a Soviet-American exhibition and seminar, INVALTECH-90 April 16-21 in Moscow. The new programme will feature technology facilitating the treatment and everyday life of handicapped and disabled persons. Seminar topics include methods of rehabilitation; medical, biomechanical, psychological and technical aspects of modern prostheses; research and development in prosthetics; demonstration of prosthetic manufacturing; demonstration of surgical methods to prepare individuals for prosthetic use; audioprosthetics; equipment for the visually impaired. Exposition topics range from prosthetics hardware to equipment for independent living, sports and vocational activities; training of the handicapped; and sensory aids. The programme is under the aegis of the Ministry of Social Security with many other governmental and nongovernmental bodies cooperating.

Australian National Society is sponsoring a five-day course, Upper Extremity Prosthetics and Orthotics, May 21-25, 1990 at the Central Development Unit, Repatriation General Hospital, Heidelberg, Victoria. Topics include basic biomechanics of below and above-elbow prostheses, prosthetic componentry, harnessing, prescription, check-out, training and management of brachial plexus lesions. The programme includes a large practical component so that participants will become proficient in the fitting, harnessing and trimlines of interim and definitive prostheses; appropriate prescription componentry, harnessing, check-out of operational and functional training of below and above-elbow prostheses. Lectures, demonstrations and group discussions are planned, with a final multiple choice examination.

United States National Member Society is enjoying a cooperative relationship with the National Headquarters of the American Orthotics and Prosthetics Association, American Academy of Orthotists and Prosthetists, and the American Board for Certification in Prosthetics and Orthotics, Inc. Indicative of the relationship is the space allocated to ISPO in the monthly AOPA Almanac. Bruce McClellan, CPO, US ISPO Executive Board member, reported in the January issue that ISPO is a unique multidisciplinary association of professional workers who have a common interest in prosthetics and orthotics, unlike other professional organizations devoted to practitioners in a single discipline. Recently, US ISPO has co-sponsored two programmes regarding new developments, namely evaluation of the ischial containment above-knee prosthetic socket, and the continuing evolution of Computer Assisted Design/Computer Assisted Manufacture (CAD/CAM) in prosthetics and orthotics. US ISPO teamed with the Texas Chapter of the American Academy of Orthotists and Prosthetists to sponsor a programme on emergency medical management. The programme reviewed medical, ethical and legal aspects of crises which might occur during the care of individuals using prostheses or orthoses. The course was extremely well received and is planned to be offered again in several other areas of the country. US ISPO was responsible for the opening afternoon programme of the annual meeting of the American Academy of Orthotists and Prosthetists in Phoenix, Arizona. The keynote speaker was Dr. George Neff, orthopedist from Germany. ISPO President Willem Eisma also addressed the convention. Members reported on orthotic and prosthetic practice and needs in Costa Rica, Russia, Saudi Arabia, New Guinea and Armenia. In addition, the functions and purposes of the International Standards Organization were described. Planning for the VII World Congress in Chicago, June 28-July 3, 1992 intensifies.

US ISPO is a co-sponsor of ACOPPRA "90" Congress for the multidisciplinary rehabilitation team to be held in San Jose, Costa Rica, March 26-28, 1990. The programme will include speakers from Mexico, Costa Rica and the United States. Topics range from social and psychological issues to the hip disarticulation prosthesis, care of the insensitive foot, fracture management, care of patients with AIDS and other infectious diseases, postsurgical treatment of amputees, upper-limb orthotics, bone lengthening, scoliosis, tone reducing orthoses, prosthetic feet, mobilization devices for children with spina bifida, cerebral palsy, upper-limb prosthetics, cosmetic restorations and optimal weight of lower-limb appliances.

The Fourth Biennial Pacific Rim Prosthetic/Orthotic Conference will take place April 24-28, 1990 on the Kohala Coast, Hawaii. A full scientific programme is planned, including exhibition of new components.

Joan Edelstein
Editor

Addendum—First meeting of the three German speaking ISPO National Societies

On February 23 and 24, the three National Societies of the totally or at least partially German speaking countries such as Germany, Austria and Switzerland organized a joint meeting in the Rehabilitation Centre of the Swiss Workers' Compensation at Bellikon, Switzerland. The Executive Board of ISPO was represented by its newly elected member Dr. Jean Vaucher from Geneva who welcomed the almost 200 participants. The meeting included interdisciplinary lectures on lower limb amputation and prosthetics. Emphasis was put on new trends and first results with CAD-CAM-manufactured sockets, the CAT-CAM-type socket and with partial foot amputations. Special prosthetic devices for various sporting activities with understandable emphasis on alpine sports were discussed.

It will now be the turn of Austria and Germany to continue this successful first experience, particularly in view of the possibilities of extending ISPO activities into the German Democratic Republic.

R. Baumgartner
Munster

Book Review

The Functional Foot Orthosis.
J. W. Philps.
Churchill Livingstone, 1990.
ISBN 0-443-04058-3.
180pp. £14.95.

There are a tremendous number of shoe inlays produced the world over, and a significant percentage of them do not act effectively. Many are in fact, disfunctional. This book can be used to advantage in improving this situation.

The author describes how to construct a functional foot orthosis to comply with the biomechanical needs of the lower limb by holding the foot as near to its optimal functioning position as possible. To obtain a correct understanding of the book, it is necessary to have at one's disposal a clear comprehension of the biomechanics of the lower limb. The author assumes this knowledge in the readers.

It is crucial for a correct construction of functional foot orthoses to assess correctly the angular position of the foot segments at the main joints and the angular relation of the segments to the ground reference line. The joint angles should be understood as a measure of angular displacement from the neutral position at the joint in question. These angles, the tibial angle, subtalar joint angle, rearfoot angle, midtarsal joint angle, and forefoot angle, are clearly defined and additionally explained by means of well illustrated examples.

The author presents a fascinating method of investigating the alignment of these important segments to promote the biomechanical analysis required for the construction of a functional foot orthosis. This method gives a well arranged overview of how the relative alignment of the segments can be used to define their relation to the ground in the form of a reference line. All angles are represented by their projection on the frontal plane of the foot.

The book gives a well structured, systematic presentation of the construction and production of foot orthoses. It is specially useful that it gives guidelines for analysis of different pathological deviations. By explaining the reasons for the pathological situation and showing the theoretical background regarding the main goal of the foot orthosis, a solid basis is already established to construct an effective foot orthosis.

The book is built up on a step by step progression, from the choice of material to the checkout of the finished orthosis. There is also a clear description of the properties, production, handling and selection of the actual materials. Even the suitable dimensions are described to cover each special need.

Advice is given regarding the use of tools and machinery to obtain the best results. By showing the most common construction faults of orthoses, and the most common failures in using machinery, the reader also obtains valuable advice about what should be avoided.

A special chapter covers unconventional shapes of orthoses used for defined severe pathological situations. The biomechanical analysis is clearly given for these cases, but because of the complexity, an understanding requires adequate previous knowledge of foot biomechanics. This emphasises that to achieve a good result demands not only practical skills, but also a deep knowledge from the orthotist of the complex structure and function of the foot.

Besides black and white photographs a large number of simplified line drawings leads to an easy understanding. The well structured layout of the book makes it practical to use.

The book can be classified as one of the best in its field by giving solid, well based introduction. It can be useful to both students and other professionals who wish to obtain high competence in producing effective foot orthoses. It is especially valuable for the reader who is interested not only in "how" but also in "why".

The book is supplemented with a "trouble shooting" scheme for the most common problems. The large list of literature is valuable for further studies.

The author creates in his own special way of handling the text an atmosphere of a living teaching experience.

George Veres
Principal
National College of Prosthetics
Delo, Norway

Prosthetics and Orthotics International, 1990, 14, 48–51

Calendar of events

20–25 May, 1990
5th European Regional Conference of Rehabilitation International, Dublin, Ireland.
Information: Olive Rhodes, Conference Secretariat, National Rehabilitation Board, 24–25 Clyde Rd., Dublin 4, Ireland.

22–24 May, 1990
3rd S. M. Dinsdale International Conference on Rehabilitation, Ottawa, Canada.
Information: Education Dept., The Rehabilitation Centre, 505 Smyth Rd, Ottawa, Ontario K1H 8M2, Canada.

4-8 June, 1990
Nordic Congress on Orthopaedics, Helsinki, Finland.
Information: Nordisk Ortopedisk Forening, Prof. P. Rokkanen, Tolo sjukhus, Topeliusgaten 5,00260 Helsinki, Finland.

5-7 June, 1990
Annual Scientific Meeting of the International Society of Paraplegia, Tel Aviv, Israel.
Information: The Honorary Secretary, IMSOP, National Spinal Injuries Centre, Stoke Mandeville Hospital, Aylesbury, Bucks HP21 8AL, UK.

7-10 June, 1990
Southern Orthopaedic Association Meeting, Hawaii, USA.
Information: AAOS, 222 S. Prospect Ave., Park Ridge, IL 60068, USA.

10-13 June, 1990
8th Nordic Meeting on Medical and Biological Engineering, Aalborg, Denmark.
Information: Dept. of Medical Informatics and Image Analysis, Badehusvej 23, DK-9000 Aalborg, Denmark.

11-14 June, 1990
American Orthopaedic Association Meeting, Colarado, USA.
Information: Ms. S. Mclellan, American Academy of Pediatrics, PO Box 927, Elk Grove, IL 60007, USA.

13-16 June, 1990
8th International Symposium on Bioengineering and The Skin, Stresa, Italy.
Information: S. Sacchi, Clinica Dermatologica, Universita di Pavia, Policinicio S. Matteo, P.le Golgi, 17100 Pavia, Italy.

13-17 June, 1990
International Society for the Study of the Lumbar Spine Meeting, Boston, USA.
Information: Prof. A. N. Nachemson, Dept. of Orthopaedics, Sahlgren Hospital, S-413 45 Goteborg, Sweden.

15-20 June, 1990
13th Annual RESNA Conference, Washington, U.S.A.
Information: Susan Leone, RESNA, 1101 Connecticut Ave. NW, Suite 700, Washington, DC 20036, U.S.A.

17–22 June, 1990
6th World Congress of the International Rehabilitation Medicine Association, Madrid, Spain.
Information: Secretary, IRMA VI, PO Box 61274, 28080 Madrid, Spain.

18–22 June, 1990
3rd International Physiotherapy Congress, Hong Kong.
Information: Miss Doreen Bauer, Scientific Convener, 3rd International Physiotherapy Congress, c/o Victoria Branch of the Australian Physiotherapy Association, PO Box 226, Collingwood, Victoria, Australia.

24-28 June, 1990
Annual Conference of the American Physical Therapy Association, Anaheim, U.S.A.
Information: Information Dept., APTA, 1111 N. Fairfax St., Alexandria, Virginia 22314, U.S.A.

8–11 July, 1990
7th European Society of Biomechanics Congress, Aarhus, Denmark.
Information: 7th Congress of the European Society of Biomechanics, Biomechanics Laboratory, Orthopaedic Hospital, Randersvej 1,8200 Aarhus N., Denmark.

6-10 August, 1990
Gordon Research Conference on Bioengineering and Orthopaedic Science, New Hampshire, USA.
Information: Alan J. Grodzinsky, Room 38-377, M.I.T., Cambridge, Massachussets 02139, USA.

26-28 August, 1990
East Coast Conference on Biomechanics, New York, USA.
Information: Prof. H. S. Ranu, Dept. of Biomechanics, Nycom, New York Institute of Technology, Old Westbury, New York, NY 11568, USA.

26-31 August, 1990
1st World Congress of Biomechanics, San Diego, U.S.A.
Information: Prof. G. W. Schmid-Schonbein, AMES-Bioengineering M-005, Univ. of California, San Diego, La Jolla, CA 92093, U.S.A.

29-31 August, 1990
6th IMEKO Conference on Measurement in Clinical Medicine and 8th Hungarian Conference on Biomedical Engineering, Sopron, Hungary.
Information: Méréstechnikai és Automatizalasi Tudomanyos Egyesulet, POB 457, H-1372 Budapest, Hungary.

6-7 September, 1990
Pediatric Orthopaedic Society Meeting, Montreal, Canada.
Information: POS, 222 S. Prospect Ave., Park Ridge, IL 60068, USA.

8-15 September, 1990
SICOT 90-18th World Conference, Montreal, Canada.
Information: SICOT 90, Sorrelcom Inc., 1425 Boul Dorchester West, 8th Floor, Montreal, Quebec H3G 1T7, Canada.

11-16 September, 1990
American Orthotic and Prosthetic Association Annual National Assembly, Boston, USA.
Information: AOPA, 717 Pendelton St., Alexandria, VA 22314, USA.

12-14 September, 1990
British Orthopaedic Association Scientific Meeting, Birmingham, England.
Information: BOA, 35–43 Lincoln's Inn Fields, London WC2A 3PN, England.

19-21 September, 1990
30th AGM of the Biological Engineering Society, Durham, England.
Information: Mrs. B. Freeman, BES, 35 Lincoln's Inn Fields, London, England.

22-27 September, 1990
Scoliosis Research Society Meeting, Hawaii, USA.
Information: SRS, 222 S. Prospect Ave., Park Ridge, IL 60068, USA.

27-29 September, 1990
Annual Congress of the Chartered Society for Physiotherapy, Bournemouth, England.
Information: Chairman, 1990 Programme Organising Committee, c/o Events Organiser, Chartered Society of Physiotherapy, 14 Bedford Row, London WC1R 4ED, England.

3-6 October, 1990
44th Annual Meeting of the American Academy for Cerebral Palsy and Developmental Medicine, Orlando, USA.
Information: AACPDM, PO Box 11086, Richmond, VA 23230, USA.

10-12 October, 1990
Clinical Orthopaedic Society Meeting, Houston, USA.
Information: COS, 222 S. Prospect Ave., Park Ridge, IL 60068, USA.

11-13 October 1990
1st IFMBE Far Eastern Conference on Medical and Biological Engineering, Tokyo, Japan.
Information: Prof. Yoshimito, West 12, South 17, Sapporo 064, Hokkaido, Japan.

14-18 October, 1990
Western Orthopaedic Society Meeting, San Antonio, USA.
Information: WOA, 2975 Treat Boulevard, Concord, CA 94518, USA.

15-19 October, 1990
Conference and Scientific Symposium of the International Federation of Multiple Sclerosis Societies, Dublin, Eire.
Information: Conference Secretariat, POB 5, Dun Laighore, Co. Dublin, Eire.

17-21 October, 1990
Eastern Orthopaedic Society Meeting, Southampton, NY, USA.
Information: EOA, 301 8th St.-Suite 3 F, Philadelphia, PA 19106, USA.

22-24 October, 1990
International Cerebral Palsy Symposium, Prague, Czechoslovakia.
Information: I. C. P. Symposium, c/o Assoc. of the Czechoslovac Medical Societies, J. E. Purkyne, tr. Vitezneho Unora 31, 120 26 Praha, 2, Czechoslovakia.

26-30 October, 1990
9th Asia/Pacific Rehabilitation International Conference, Beijing, China.
Information: J. Morrison, RADAR, 25 Mortimer St., London W1N 8AB, UK.

29-31 October, 1990
Functional Electrical Stimulation; Practical Aspects for Clinicans, Glasgow, Scotland.
Information: Dr. C. A. Kirkwood, Bioengineering Unit, Wolfson Centre, 106 Rottenrow, University of Strathclyde, Glasgow G4 0NW, Scotland.

5-8 November, 1990
European Conference on the Advancement of Rehabilitation Technology, Maastricht, The Netherlands.
Information: ECART, Congress Organization Services, Van Namen and Westerlaken, PO Box 1558,6501 BN Nijmegen, The Netherlands.

19-22 November, 1990
2nd North Sea Conference on Biomedical Engineering, Antwerp, Belgium.
Information: Luk Pauwels, Technologisch Instituut-KVIV, Desguinlei 214,2018 Antwerpen, Belgium.

6–8 December, 1990
6th International Conference on Biomedical Engineering, Singapore.
Information: Secretary, 6th Biomed. Conference 1990, Dept. of Orthopaedic Surgery, National University Hospital, 5 Lower Kent Ridge Rd, Singapore 0511, Republic of Singapore.

1991
31 January–3 February, 1991
Combined Sections Meeting of the American Physical Therapy Association, Anaheim, U.S.A.
Information: Information Dept., APTA, 1111 N. Fairfax St., Alexandria, Virginia 22314, U.S.A.

19-24 March, 1991
AOPA Annual Meeting and Scientific Symposium, San Diego, U.S.A.
Information: AOPA, 717 Pendleton St., Alexandria, Virginia 22314, U.S.A.

3-6 June, 1991
American Orthopaedic Association Annual Meeting, Palm Beach, USA.
Information: AOA, 222 S. Prospect Ave., Park Ridge, IL 60068, USA.

22-26 June, 1991
14th Annual RESNA Conference, Kansas City, USA.
Information: Susan Leone, RESNA, 1101 Connecticut Ave.Nw,Suite 700, Washington DC 20036, USA.

23-27 June, 1991
Annual Conference of the American Physical Therapy Association, Boston, U.S.A.
Information: Information Dept., APTA, 1111 N. Fairfax St., Alexandria, Virginia 22314, U.S.A.

7–12 July, 1991
16th International Conference on Medical and Biological Engineering, Kyoto, Japan.
Information: Dr. O. Z. Roy, IFMBE, c/o National Research Council of Canada, Room 164, Bldg. M50, Ottawa, Ontario, K1A OR8, Canada.

28 July-2 August, 1991
11th Congress of the World Confederation for Physical Therapy, London, England.
Information: Secretariat, WCPT, Conference Associates, 27A Medway St., London SW1P 2BD, England.

7-11 August, 1991
Southern Orthopaedic Association Meeting, Colorado Springs, USA.
Information: SOA, 222 S. Prospect Ave., Park Ridge, IL 60068, USA.

1–6 October, 1991
American Orthotic and Prosthetic Association Annual National Assembly, California, USA.
Information: AOPA, 717 Pendelton St., Alexandria, VA 22314, USA.

13-16 October, 1991
7th International Conference on Mechanics in Medicine and Biology, Ljubljana, Yugoslavia.
Information: ICMMB 91, Technical Organiser, CANKARJEV DOM, Cultural and Congress Centre, Kidricev park 1,61000 Ljubljana, Yugoslavia.

9-13 December, 1991
13th International Conference on Biomechanics, Perth, Australia.
Information: 13th ISB Congress Secretariat, Dept of Human Movement, Univ. of Western Australia, Nedlands WA 6009, Australia.

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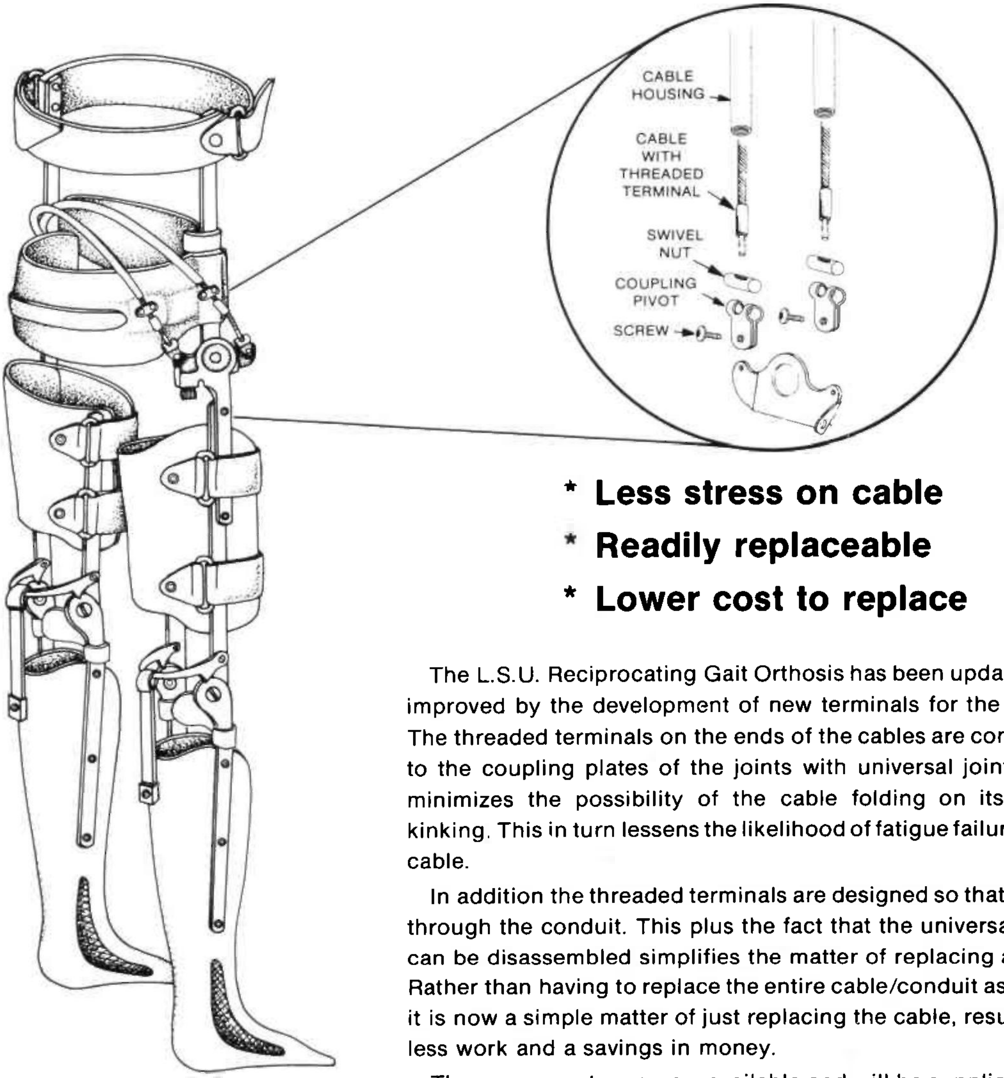

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* **Lower cost to replace**

The L.S.U. Reciprocating Gait Orthosis has been updated and improved by the development of new terminals for the cables. The threaded terminals on the ends of the cables are connected to the coupling plates of the joints with universal joints. This minimizes the possibility of the cable folding on itself and kinking. This in turn lessens the likelihood of fatigue failure of the cable.

In addition the threaded terminals are designed so that they fit through the conduit. This plus the fact that the universal joints can be disassembled simplifies the matter of replacing a cable. Rather than having to replace the entire cable/conduit assembly it is now a simple matter of just replacing the cable, resulting in less work and a savings in money.

These new parts are now available and will be supplied when ordered. Existing stocks will be dropped as they are depleted.

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