The Journal of the International Society for Prosthetics and Orthotics

Prosthetics and Orthotics International

Including Supplement on Seating

April 1992, Vol. 16, No. 1

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Prosthetics and Orthotics International

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ISPO

Elected Members of Executive Board	
W. H. Eisma (President)	Netherlands
M. Stills (President Elect)	USA
S. Heim (Vice President)	Germany
A. Jernberger (Vice President)	Sweden
V. Angliss	Australia
P. Christiansen	Denmark
T. Keokarn	Thailand
J. Vaucher	Switzerland
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E. Lyquist (Past President)	Denmark
G. Murdoch (Past President)	UK
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N. A. Jacobs (Hon. Secretary)	UK
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J. Hughes (Protocol and Nominations)	UK
M. Stills (Congress and Membership)	USA
J. Edelstein (International Newsletter and Journal Promotion)	USA
H. C. Thyregod (Professional Register)	Denmark
H. J. B. Day (Limb Deficient Child)	UK
V. Angliss (Publications)	Australia
B. Klasson (Socket Design)	Sweden
S. Heim/J. Hughes/G. Murdoch (Education)	Germany/UK/UK
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R. Henze (IVO)	Germany
J. Van Rolleghem (INTERBOR)	Belgium
J. N. Wilson (WOC)	UK
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G. Murdoch (1977–1980)	UK
A. Staros (1980–1982)	USA
E. Lyquist (1982–1983)	Denmark
E. G. Marquardt (1983–1986)	Germany
J. Hughes (1986-1989)	UK
Secretary	
Aase Larsson	Denmark

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Editorial

This issue of Prosthetics and Orthotics International contains the financial statement for 1991, and also a review of finances extending over two triennial periods, 1986–1991 (shown overleaf).

In 1990 we had a small positive result on the years finances and the income from membership fees matched the costs of running the organization, publishing reports from conferences and producing our official journal, Prosthetics and Orthotics International. With the sponsorships from the Society and Home for the Disabled (SAHVA) and the War Amputations of Canada a break even was also achieved in 1991. High costs were accrued on Prosthetics and Orthotics International due to a considerable increase in the number of printed pages. It is, however, the opinion of the Executive Board that the special issues highlighting particular fields have improved the scientific value of the journal. It is also strongly believed that an extra effort in attracting more advertising will further reduce costs. There is a slight alteration in the presentation of the accounting with respect to courses and congresses — expenses are not displayed until income appears.

Our primary result has again become negative because of ISPO's engagement in international work. This is the flagship of our organization in influencing education programmes and care of the amputees and the severely disabled, in particular in the developing countries.

Our accounts do not display or set value on the many hours and personal expenses contributed by our board members and officers participating in such work. Neither does it set value on the silent approval by our employers, who allows for many hours of work devoted to ISPO, in particular at the National Centre in Glasgow. Furthermore SAHVA continues to provide office facilities in Copenhagen without costs for the organization.

The 6-years key-figures show that we have increased our assets by nearly 12 per cent, mainly through careful investment policies. This is in spite of high expenses due to international work and the arrangement of conferences and work-shops. The costs for our secretariat has increased by 28 per cent reflecting inflation and a small increase in work-load, but the expenses reimbursed to the Executive Board and officers have been largely unchanged over the past 5 years.

In 1990 the membership fee was raised from 400 DKK to 450 DKK per year, but the fee for members in developing countries was not adjusted. The Executive Board has decided to keep the fee unchanged for 1993 at 450 DKK for 25 countries from Organisations for Economic Co-operation and Development (OECD) and other high income areas, as defined by the World Bank. With the purpose of evoking further interest in ISPO it has been decided that members from all other countries can enrol at half that fee, i.e. 225 DKK per year, although this means increased fees for a relatively small number of members.

We all look forward to meeting at the Seventh World Congress in Chicago, 1992.

J. Steen Jensen
Treasurer

ISPO Accounting 1986 – 91

	1986	1987	1988	1989	1990	1991
Income	**850.747**	**811.872**	**865.823**	**1.034.989**	**1.150.259**	**1.129.242**
– members	716.323	709.289	758.759	899.545	1.017.829	991.546
– sponsors	134.424	102.583	107.064	135.444	132.430	137.696
Meetings with other orgs.	**–5.167**	**–61.953**	**–149.785**	**–135.565**	**–147.719**	**–209.681**
Confs./Workshops	**–7.691**	**–536.777**	**–39.476**		**–204.220**	**31.106**
Courses			**–6.066**			
Congresses	**325.751**	**–408**	**–4.477**	**–117.523**	**82.370**	
Journal	**–12.075**	**–83.474**	**–78.214**	**–65.242**	**–84.584**	**–266.641**
– income	209.518	283.101	282.723	306.270	311.618	414.744
– expenses	–221.593	–366.575	–360.937	–371.512	–396.202	–681.385
Professional register				**–8.681**	**–32.934**	**–3.953**
Publications	**–59.961**	**7.326**	**12.212**	**–106.018**	**3.695**	**–12.979**
– income	3.206	7.326	12.212	10.820	20.320	22.755
– expenses	–63.167			–116.838	–16.625	–35.734
Activity result	**1.091.604**	**136.586**	**600.017**	**601.960**	**766.867**	**667.094**
Administration	**–580.283**	**–705.106**	**–802.521**	**–998.111**	**–858.387**	**–833.988**
– secretariat	–429.265	–432.773	–486.652	–555.263	–536.342	–548.992
– board	–110.609	–226.906	–283.307	–392.056	–272.017	–253.705
– meeting expenses	–40.409	–45.427	–32.562	–50.792	–27.835	–20.213
– International Comm.					–22.193	–11.078
Primary result	**511.321**	**–568.520**	**–202.504**	**–396.151**	**–91.520**	**–166.894**
Capital yield	**539.433**	**173.544**	**634.260**	**209.424**	**258.371**	**387.244**
– interest. maturity yield	539.433	297.544	430.356	294.728	324.595	322.960
– changes in value		–124.000	203.904	–85.304	–66.224	64.284
Years result	**1.050.754**	**–394.976**	**431.756**	**–186.727**	**166.851**	**220.350**
Assets	**3.758.222**	**3.354.659**	**3.722.592**	**3.732.140**	**3.829.983**	**4.190.597**

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ISPO Statement of Accounts, 1991

Auditor's Report

The financial statements for the year 1991 have been audited by the undersigned.

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

The financial statements have been prepared in accordance with statutory requirements, and the constitution of the Society and generally accepted accounting principles. In my opinion the financial statements give a true and fair view of the state of the association's affairs as of December 31, 1991 and of the result for the year then ended.

Copenhagen, February 17, 1992.

Søren Wonsild Glud
State Authorised Public Accountant

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

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

Income Statement for the Year 1991

SUMMARY		**1991**	**1990**
Society membership fees (note 1)		991.546	1.017.829
Sponsorship (note 2)		137.696	132.430
Meetings with other organisations (note 3)		(209.681)	(147.719)
Conferences, courses etc (note 4)		31.106	(204.220)
World congress 1989		–	82.370
Prosthetics and Orthotics International (note 5)		(266.641)	(84.584)
Professional register		(3.953)	(32.934)
Publications (note 6)		(12.979)	3.695
Activity result		657.604	776.867
Administration expenses (note 7)		(833.988)	(858.387)
Primary result		(166.894)	(91.520)
Interest (note 8)		316.321	318.345
Dividend (note 8)		1.504	1.755
Maturity yield (note 8)		5.135	4.495
Change in value of securities (note 8)		64.284	(66.224)
Result for year	DKK	220.350	166.851

Balance sheet as of December 31, 1991

	1991	1990
ASSETS		
Cash	964.215	567.577
Accrued interest	68.468	70.155
Advertising receivable	64.202	81.182
Prepayment, World Congress, Chicago	142.616	130.080
Advance funding of world congress 1980	87.437	87.437
Receivables	362.723	368.854
Securities (note 9)	2.863.660	2.869.239
Office equipment (note 8)	0	24.313
Total assets	4.190.598	3.829.983
LIABILITIES AND CAPITAL		
Accrued expenses	69.898	50.023
Accrued printing costs	–	120.000
Prepaid membership fees	34.100	6.591
Prepaid advertising income	6.174	2.115
Prepaid subscription income	96.600	88.000
Prepaid conference fees	200.221	–
Short-term liabilities	406.993	266.729
Provision World Congress 1980	87.437	87.437
Equity capital January 1, 1991	3.475.817	3.308.966
Result for the year	220.350	166.851
Equity capital December 31, 1991	3.696.167	3.475.817
Liabilities and Capital	DKK 4.190.597	3.829.983
Contingent liabilities (note 10)		

Notes to the Financial Statements

1. Society membership fees
Membership fees consist of payments from 2236 listed members, excluding 45 honorary members.

	1991	1990
2. Sponsorship		
Contribution from: the War Amputations of Canada	37.696	32.430
SAHVA	100.000	100.000
	DKK 137.696	132.430
3. Meetings with other organisations		
AOPA	62.259	61.397
OT Berlin	54.449	–
Asian Conference Japan	28.040	–
American Academy of Orthotists and Prosthetists	26.750	14.664
Education Commission	11.159	20.513
World Orthopaedic Concern	9.493	4.640
Interbor Brussels	8.013	–
UN Vienna	6.674	–
Other meetings 1990	–	41.571
ACCOPRA	2.845	4.934
	DKK 209.682	147.719

4. Conferences, courses etc			
Amputation surgery – profit		(31.106)	–
Consensus Conf Oct 1990 – cost		–	204.220
	DKK	(31.106)	204.220
5. Prosthetics and Orthotics International			
Advertising		189.069	145.619
Subscriptions		225.675	165.999
		414.744	311.618
Printing and mailing		(603.820)	(365.422)
Production editor		(52.640)	(16.524)
Meeting expenses		(24.925)	(14.256)
		(681,385)	(396.202)
Net result	DKK	(266,641)	(84.584)
6. Publications			
Booksales		22.755	20.320
CAD CAM report		(19.614)	–
Amputation surgery consensus report		(16.120)	–
Other reports		–	(16.625)
Total cost	DKK	(12.979)	3.695
7. Administrative expenses			
Executive board and office:			
Travel and hotel cost		253.705	272.017
Meeting expenses		20.213	27.835
International committee		11.078	22.193
	DKK	284,996	322.045
Secretariat, Copenhagen:			
Fees other organisations		–	4.209
Staff salaries		307.696	283.190
Labour tax		12.968	12.446
Data service		6.386	7.587
Meeting expenses		16.181	52.134
Bank expenses		6.863	7.710
Postage		52.403	73.056
Telephone		5.437	4.591
Stationery		36.232	15.301
Office supplies		6.904	3.841
Auditing		38.526	30.500
Book keeping		19.022	23.690
Sundry		6.570	(6.226)
Society promotion		9.490	–
Depreciation		24.314	24.313
		548.992	536.342
Administrative expenses, total	DKK	833.988	858.387
8. Office equipment			
Computer equipment, at cost		95.347	95.347
Office equipment, at cost		26.220	26.220
		121.567	121.567
Depreciation January 1		(97.254)	(72.941)
Depreciation December 31		(24.313)	(24.313)
		(121.567)	(97.254)
	DKK	0	24.313

9. Securities

		Nominal value	Original cost	Year end value	Interest/ dividend
Bonds					
9% Kred. Danmark 2007		3.043.000	2.834.708	2.926.925	276.661
Shares					
Den Danske Bank		9.400	30.891	28.952	1.504
Total	DKK	3.052.400	2.865.599	2.955.877	278.165

10. Contingent liabilities

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

Prosthetics and Orthotics International, 1992, 16, 7–10

Executive Board Meeting

28 and 29 January 1992

The following paragraphs summarise the major discussions and conclusions of the Executive Board Meeting held in Groningen in January. They are based on the draft minute of that meeting which has yet to be approved by the Board.

International Committee Meeting

The Executive Board discussed the draft agenda for the International Committee meeting to be held in Chicago on 26 and 27 June 1992. The draft agenda included the points which would have been discussed at the interim meeting of the International Committee Representatives had it been held, as well as the matters of business normally conducted by the International Committee. The draft agenda will be circulated to National Member Societies asking for their comments and any further items for discussion.

Standing Committee Chairmen and Task Officer Reports

The Chairman of the Finance Committee presented an overview of the financial results for the past five years as well as a preliminary accounting for 1991. This showed the Society to be in a rather stable position. However, he suggested that efforts to recruit more members to the Society as well as subscribers and advertisers to the Journal would go a long way to improving the financial stability of ISPO. *(Honorary Secretary's Note: An analysis of the finances of the Society over the past six years can be found in the Editorial of this issue of the Journal as well as a full financial statement for 1991).* The Honorary Treasurer outlined a proposed budget for 1992 which showed a slight deficit. However, it did not inlcude the outcome of the coming Congress in Chicago. The Executive Board approved the proposed budget for 1992. The Executive Board agreed that the international fee should remain at 450DKK for 1993 and attempts would be made to hold this rate throughout the coming triennium. The Executive Board agreed to a proposal from the Honorary Treasurer that the 450DKK fee should apply to the 25 countries of the Organisation for Economic Co-operation and Development (OECD) and other high income areas as defined by the World Bank and that all other countries should have a subscription of half that amount, ie 225 DKK per annum.

At the request of the President, the Protocol and Nominations Committee had examined the submissions for the Blatchford Prize. At present no recommendation was made. The procedure to select the prize winner for the Frochheimer Prize was activated and the prize winner will be selected within the next few months. As the Society had no firm guidelines with regard nominations for Honorary Fellowship of the Society, it was agreed that the Protocol and Nominations Committee should prepare a draft of such guidelines and present them to the next Executive Board meeting. A further proposed amendment to the Constitution, Clause 4.5.5 had been received from the UK National Member Society. This was in relation to the composition of the Protocol and Nominations Committee. This proposed amendment was duly published to the International Committee and the Membership in the December 1991 issue of the Journal and will be discussed and voted upon at the International Committe meeting.

The Honorary Secretary reported that there had been a slight decrease in the membership in 1991 to 2,280 as compared to 2,306 in 1990. Subscribers to the Journal had increased during that period from 354 to 386. There are now 22 National Member Societies. Attempts to form National Member Societies in Indonesia, Central America, Agrentina, Chile, Colombia, Thailand and Slovenia were being pursued. The President Elect indicated that the membership brochure was under development, but that there is still the difficulty in obtaining high quality illustrations. However, he will attempt to have it completed in time for the Chicago Congress.

The Honorary Secretary reported on his recent visit to the Tanzanian Training Centre for Orthopaedic Technology (TATCOT) during which he had carried out an inspection of their training

course. A number of changes had been made to the course in accordance with the recommendations made at the last visit. The Executive Board discussed the report and agreed that TATCOT be recognised for a further three years after which a re-inspection should take place. The Honorary Secretary reported that the International Committee of the Red Cross (ICRC) had submitted an outline syllabus of their courses in Mozambique and Afghanistan. The difficulty in recognising these courses lies in the fact that they are of a fixed term, taking one intake of students only, after which the courses are terminated. It was agreed that the Society should explore the possibility of recognising the syllabus and participate in the final examination of these courses. John Hughes reported on the activities of the ISPO/INTERBOR Joint European Education Committee. An analysis of the completed questionnaires on education and training in the different European Community countries had been carried out. A report on this analysis was subsequently submitted to the European Community Action Scheme for the Mobility of University Students (ERASMUS) together with a letter requesting further financial support to continue the studies. The Executive Board discussed a letter, from W. Neumann, President of the American Board for Certification (ABC) in Orthotics and Prosthetics, which referred to the internationalisation of orthotic and prosthetic certification and it was agreed that W. Neumann be invited to make a presentation on behalf of ABC to the next Executive Board meeting.

The Publications Committee was considering a scheme to promote journal subscriptions. Their findings would be presented to the next Executive Board meeting. They had felt that the International Newsletter had not been totally successful and that the views of the International Committee on its continuation should be sought at its meeting in Chicago. The report on the Consensus Conference on Amputation Surgery has been published and distributed to those who took part. The publication of the report for sale is at present being investigated. *(Secretary's Note: Since the meeting, it has been agreed that the report of the Consensus Conference on Amputation Surgery shall be published for sale).* The Executive Board noted that the Society was exploring the possibility of publishing a text book on Amputation Surgery with Butterworth Heinemann's based on the outcome of the Consensus Conference.

It was reported that the International Standards Organisation's Technical Committee 168 Working Group 1 (ISO TC168 WG1) on Nomenclature and Classification in Prosthetics and Orthotics had met recently in Glasgow. The work of the Committee related to prosthetics had been progressing steadily and it had been agreed that proposals should be formulated to expand the work of TC168 to include orthotics. The convenors of WG1 and WG2 (Medical Aspects) will meet to discuss this matter and prepare proposals.

The President reported on the arrangements for the up-date course on Lower Limb Amputation and Related Prosthetics to be held in Groningen. There were at that time approximately 80 registrations for the course exclusive of faculty. In addition, there were 4 exhibitors. A Reception and Buffet has been offered by the University of Groningen and the City of Groningen on the first evening of the course, and a Course Dinner arranged for the final evening. It is anticipated that a profit would be made for the Society on this event. The Board discussed the possibilities of holding similar courses in other parts of the world. It was agreed to explore organising a course in Tanzania from 14th–18th September directly after the Rehabilitation International Congress in Kenya. The possibility of holding a course in Bangkok, Thailand in January 1993 was also being considered, as were other courses in Eastern Europe and Central or South America.

H.J.B. Day's report on the Limb Deficient Child was discussed by the Executive Board and it was agreed that a proposed register of limb deficient children was not a viable exercise and it was felt that the task for the Limb Deficient Child has been completed for the present. The Executive Board thanked Dr. Day for all his efforts in this area.

Bo Klasson described the work done in Sweden with regard socket design. He felt that he was not in a position to make a proposal with regard holding a workshop on the subject to the next Executive Board meeting.

The Executive Board noted Joan Edelstein's resignation as Task Officer for Journal Promotion and as editor of the International Newletter as a result of her new employment. The Board thanked her for all her efforts in these areas.

International Consultants

The President reported that in November he had represented the Society at a Regional Conference in Tokyo on Education in the Developing Countries of South East Asia. Proposals emanated from this meeting to establish a school in prosthetics and orthotics in Indonesia either in or associated with a University. This would produce mid-level professionals for this region. Seishi Sawamura and Eiji Tazawa would be in overall charge of this project and it was suggested that this school might be run in collaboration with ISPO. The Executive Board discussed this proposal and expressed their support in principle. It was hoped that further information would be presented to the next Board Meeting.

Cît Marincek informed the Board of the current difficulties in Yugoslavia. Slovenia was largely unaffected by the conflict, however Croatia has over 700 amputees and 300 paraplegics from the armed forces alone as a direct result of the conflict. ICRC and Handicap International are presently active, however, when the conflict is over there will be a need for training to allow the development of services for that region.

International Organisations

Discussions are underway with INTERBOR to put forward a project proposal on CAD CAM Evaluation for Phase II of the European Community programme, Technology Initiative for Disabled and Elderly Persons (TIDE). It is anticipated that a second bid will be made with ISPO and INTERBOR as full partners.

It was reported that arrangements were being made to participate in the RI World Congress in Nairobi, Kenya from 7–11 September 1992. Harold Shangali and Wilfred Raab of TATCOT were preparing an exhibt that would show the work of ISPO, TATCOT and the German Agency for Technical Co-operation (GTZ). In addition, they were organising a session on prosthetics and orthotics and it is hoped that ISPO will make a presentation in Plenary Session.

The Honorary Secretary reported that he had represented the Society at a consultative meeting of non-governmental organisations at the UN Office in Vienna on 2 December 1991. He had made a presentation on ISPO activities over the past year and emphasised the need to produce trained personnel in prosthetics and orthotics in order to provide satisfactory services in the developing world. The Executive Board discussed the Society's Official Representatives to the UN and it was agreed that the Honorary Secretary, Sepp Heim and Jean Vaucher be appointed to represent the Society in the Geneva and Vienna Offices and the President Elect and Joan Edelstein to represent the Society in the New York Office.

The President reported that both he and the Honorary Secretary had met with the President of the World Rehabilitation Fund (WRF) and that it was a friendly meeting, during which future collaboration was discussed, although no arrangements were finalised. The Board agreed that this matter should be pursued.

Congresses

Dudley Childress reported on progress regarding the Chicago Congress. Arrangements for the Exhibit were going well and to date 165 booths have been sold. Arrangements for the programme were also going well. The deadline for papers has been extended to 1 February and to date 250 abstracts have been received. The committees which have been established were working hard. Arrangements with regard a social programme were well underway. It was anticipated that the Congress will be successful both as far as scientific programme and participation are concerned. *(Honorary Secretary's Note: Since the meeting, it has been reported that over 400 abstracts for presentations have been reviewed and over 180 booths for the exhibit have been sold. Please see the World Congress Update printed in this issue).*

Valma Angliss reported on the arrangements being made for the 1995 Congress in Melbourne. She has been nominated Secretary General and a planning schedule had been prepared. She anticipated that an outline programme would be prepared and presented at the International Congress Committee Meeting to be held in Chicago. The logo for the VIII World Congress had been finalised and work was started on preparing fliers for distribution at the Congress in Chicago.

The Executive Board discussed the selection of future venues for Congresses and the possibility of sharing profits with National Member Societies who organise Congresses. It was agreed that these matters should be discussed at the International Committee Meeting in Chicago.

Nominations for the Executive Board 1992

The Honorary Secretary informed the Executive Board of the arrangements which had been made for nominating the Executive Board for the Triennium 1992–1995. He had written to the International Committee Representatives, outlining the Board's Slate of Nominations and calling for further nominations. Since then one further nomination had been received from the UK National Member Society, namely, David N. Condie. The closing date for nominations is 28 February after which a ballot will take place.

(Honorary Secretary's Note: By the 28 February, no further nominations to the Executive Board had been received.

The nominations are as follows:

President:	Melvin Stills	USA	Orthotist
President-Elect:	Seishi Sawamura	Japan	Orthopaedic Surgeon
Vice-Presidents:	Per Christiansen	Denmark	Prosthetist/Orthotist
	Jean Vaucher	Switzerland	Orthopaedic Surgeon
Members:	Hans Arendzen	Netherlands	Rehabilitation Doctor
	David N. Condie	UK	Rehabilitation Engineer
	Margaret Ellis	UK	Occupational Therapist
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	Harold Shangali	Tanzania	Prosthetist/Orthotist
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As there are five nominations for the four Member vacancies, a postal ballot is now being conducted).

Honorary Fellowship

The President was pleased to announce that the nomination of John Hughes by the UK National Member Society for an Honorary Fellowship had been approved by the International Committee. He thanked John Hughes for all the work he has contributed and continues to put into the Society and congratulated him on being awarded this honour.

Norman A. Jacobs
Honorary Secretary

The Society is pleased to announce the formation of a New Regional Member Society in the Caribbean.

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

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Abstract

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

Introduction

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

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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

away from the femur except at the most distal point. When the femur exerts force against the lateral wall in weightbearing, the quadrilateral socket moves laterally immediately, because the ischium has no effect on stopping this shift. With the more narrow socket and increased A-P, the ischium is inside the socket, preventing lateral shifting of the socket during weight-bearing."

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

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

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

The solution

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

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

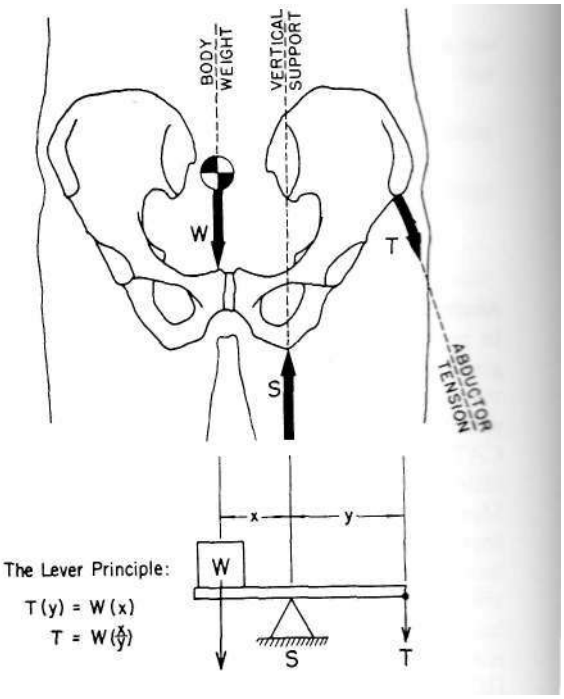

Fig.l. Moments acting about the support point in the frontal plane in lateral stabilization of the pelvis (Radcliffe, 1955).

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

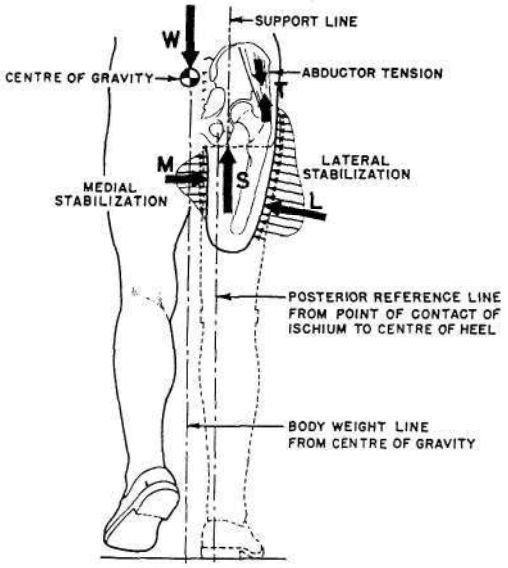

Fig. 2. Lateral stabilization of the pelvis (Radcliffe, 1970).

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

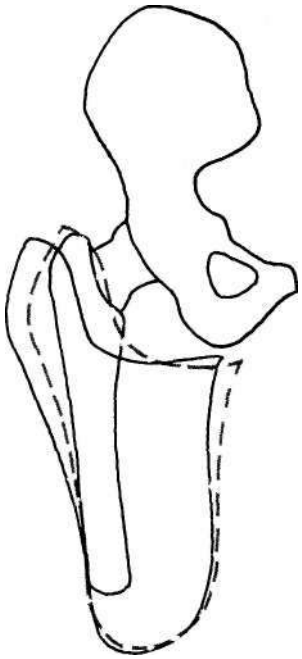

Fig. 3. Socket displaced laterally (solid) from its original resting position (dotted). This can result in a shearing force being exerted against the tissues between the medial bones of the pelvis and the medial brim of the socket.

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

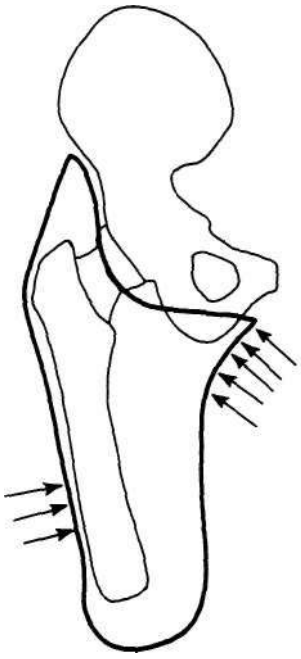

Fig. 4. Medial forces borne by bones of the pelvis and soft tissues, the principle of ischial containment.

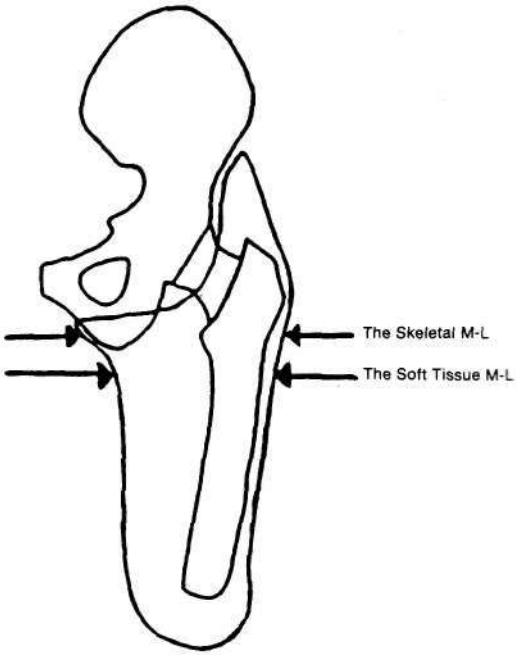

Fig. 5. Skeletal and soft tissue M-L dimensions of the Ischial Containment Socket (Hoyt et al, 1985).

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

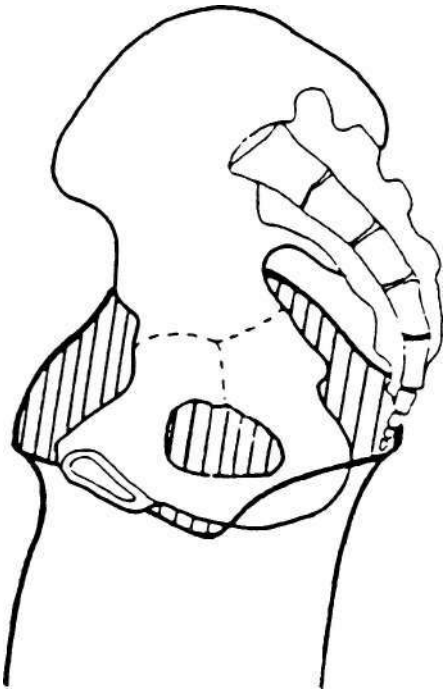

Fig. 6. Medial view of the Ischial Containment Socket in the sagittal plane showing relationship between the proximal edge of the medial wall and the bones of the pelvis (Hoyt et al, 1985).

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.

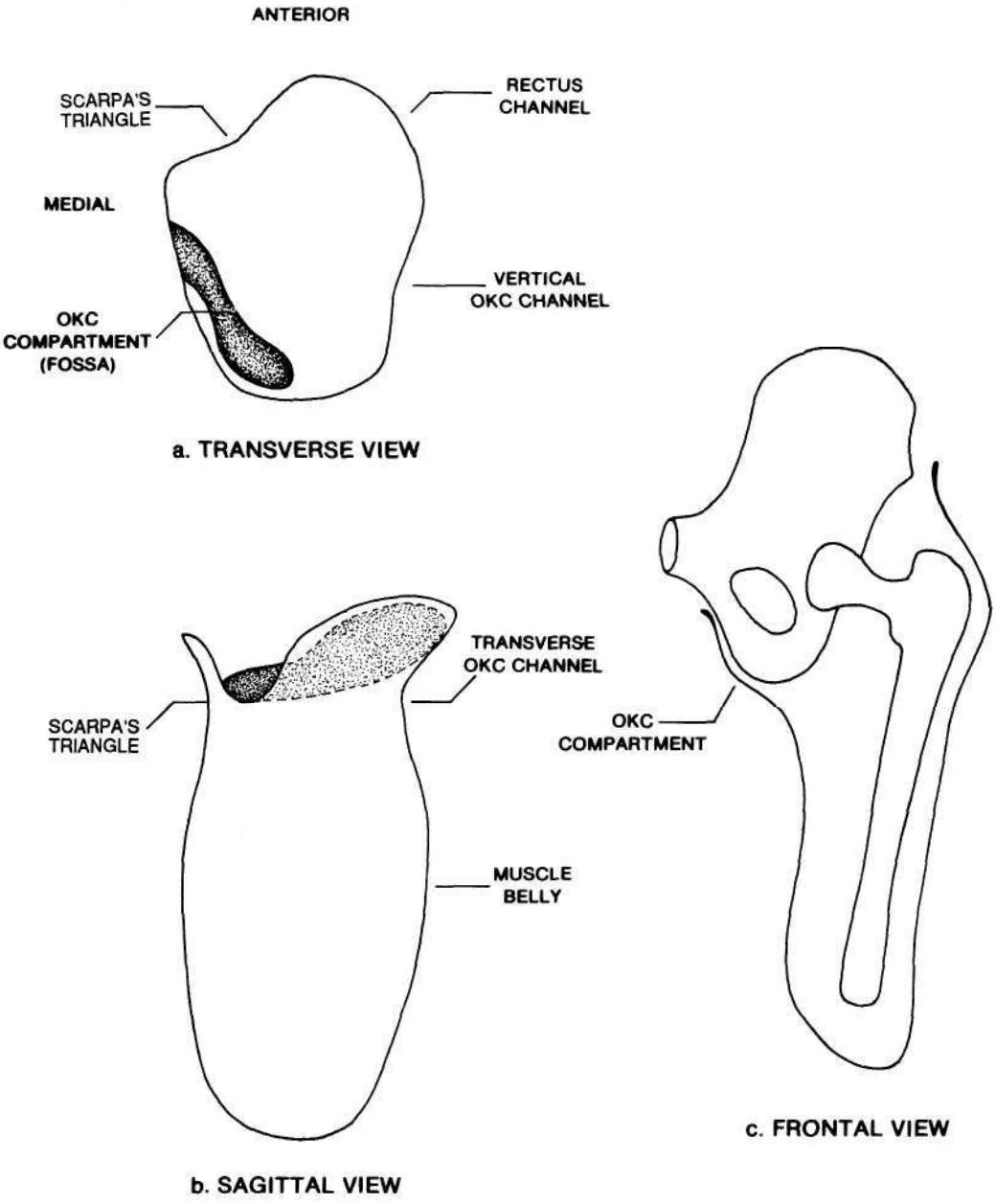

Fig. 7. The SCAT-CAM socket as described by Sabolich. Redrawn from an original by him to clarify some points. According to him some contours are exaggerated for emphasis, but not by much.

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

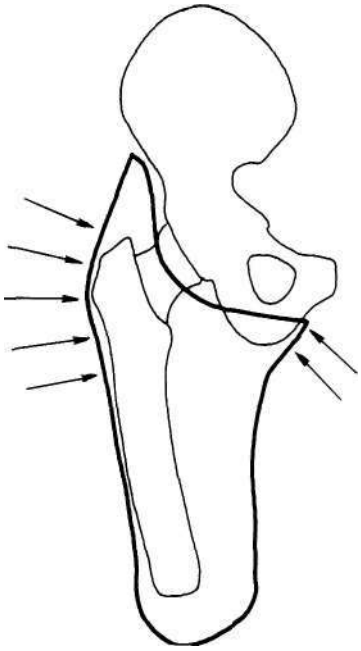

Fig. 8. Counterpressure generated by the lateral wall.

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

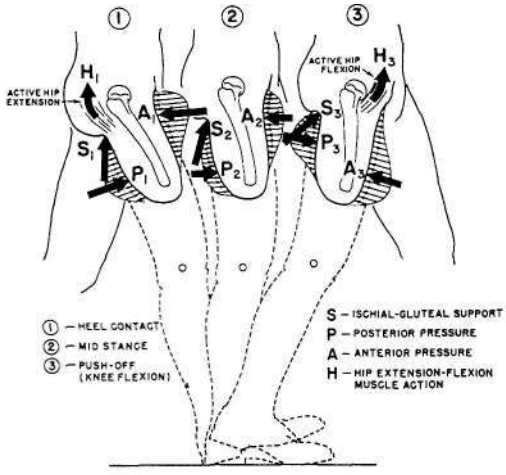

Fig. 9. Force patterns generated in the sagittal plane during gait (Radcliffe, 1970).

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 communication he states—

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

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

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

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

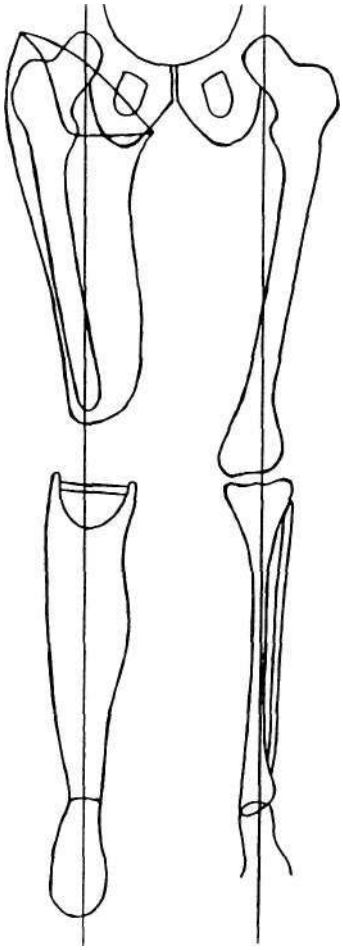

Fig. 10. Long's Line from: fabricating the Long's line above-knee prosthesis. (Long, 1985.)

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

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

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

Conclusion

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

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

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

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

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REFERENCES

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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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.

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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.**

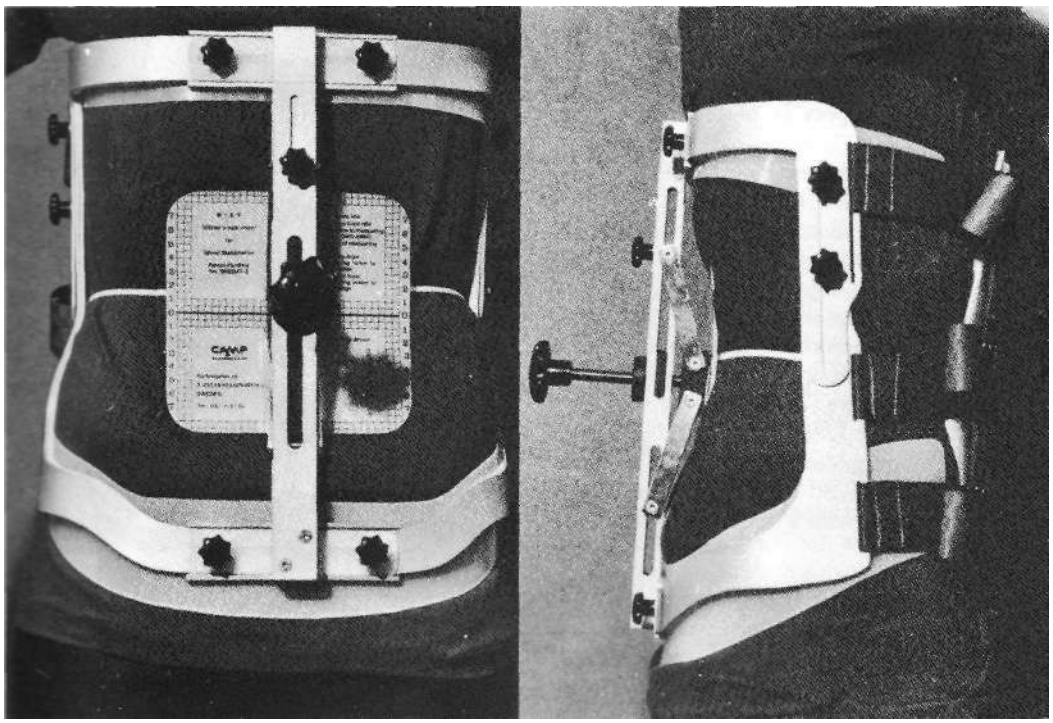

Fig. 1. Test instrument. Left, posterior view. Right, lateral view.

Method

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

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

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

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

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

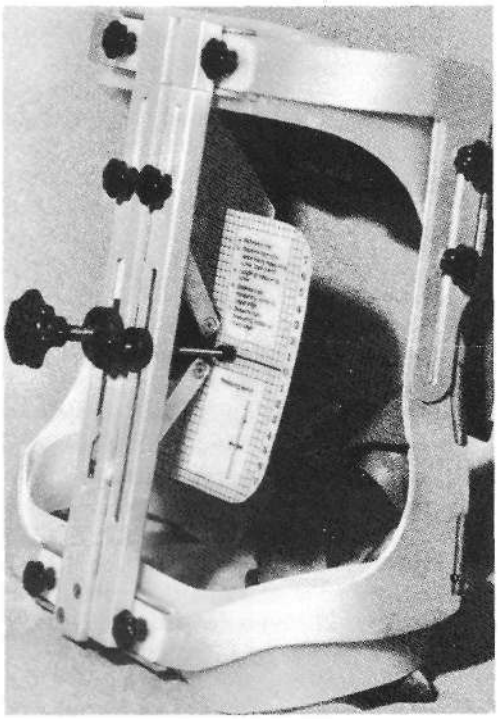

Fig. 2. The measuring device fixed on the test instrument.

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

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

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

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

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

(a) The reference line—the line joining the upper palpable corners of the iliac crest.
(b) The null line—the horizontal line on the back support marking the level of the maximal dorsal support.
(c) The central line—the horizontal line marked on the abdominal pad, the level for the maximal thickness of the pad.

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

(1) The height of the brace to be fabricated.
(2) The level of the null line in relation to the upper and lower edges of the brace.
(3) The range of lordosis. This is measured by screwing the screw of the measuring stick until the top of the screw touches the back support at the level of the null line. The distance from the top of the screw to the measuring stick is recorded.

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

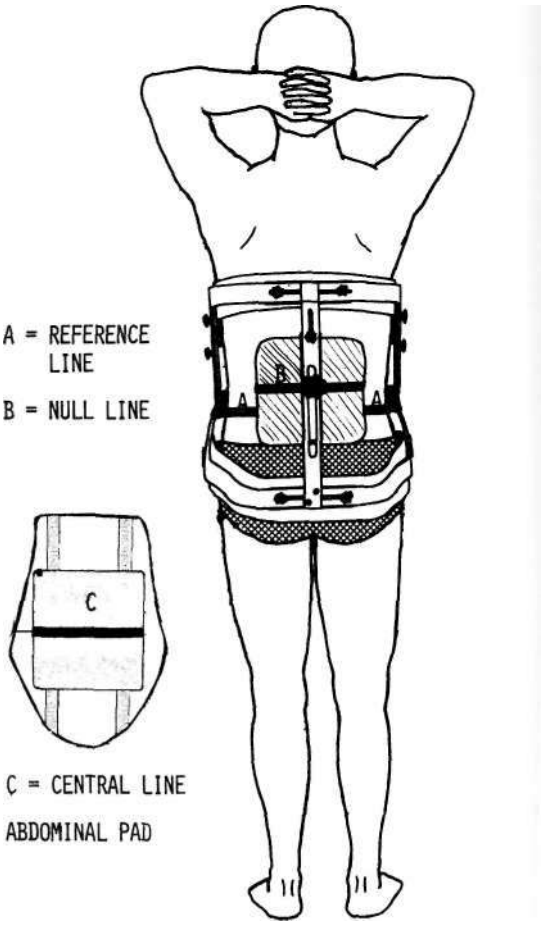

Fig. 3. Reference lines to allow the information observed in the test instrument to be transferred to the brace to be manufactured.

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

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

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

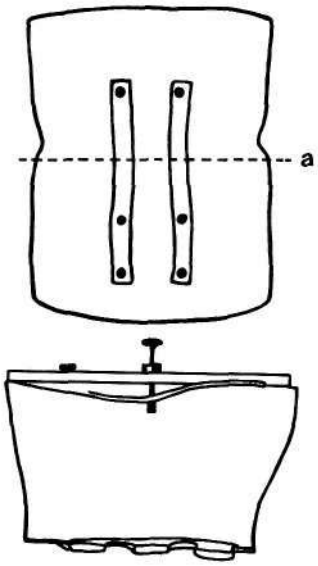

Fig. 4. Top, marking of the reference line (a) on the prefabricated module of a rigid brace. A hole is drilled at the level of the null line. Bottom, measuring device is placed on the module and the screw passes through the hole at the level of the null line.

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

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

Patient tests

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

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

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

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

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

Results

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

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

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

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

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

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

Discussion

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

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

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

Based on these observations the test instrument described was developed.

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

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

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

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

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

Conclusion

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

REFERENCES

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

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

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

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

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

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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-M0ller 1983, Light et al. 1980, Wosk & Voloshin 1985), and furthermore such soles have been found to be of advantage in relation to low back pain, and foot fatigue and stiffness (Dyer 1983, Wosk & Voloshin 1985), and may improve comfort and provide pain relief (Clark etal. 1989).

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

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

Participants and methods

Participants

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

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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-M0ller 1983) and still photography were used in combination with accelerometry and a force-time recording from a force-platform (AMTI^{R}). Both feet were placed on the platform while the crutches were outside. The LEDs were positioned on the leg brace at mid shank, at the heel and at the forefoot. The diodes were fed by a 50 Hz signal from which, however, one impulse was omitted each second. The 50 Hz signal was further registered on the oscillogram together with the signal from the accelerometer with the missing flash forming an exact time link between the recordings and the photography (Fig. 4, left and top right).

Procedures

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

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

The participants were allowed 5 min. to get

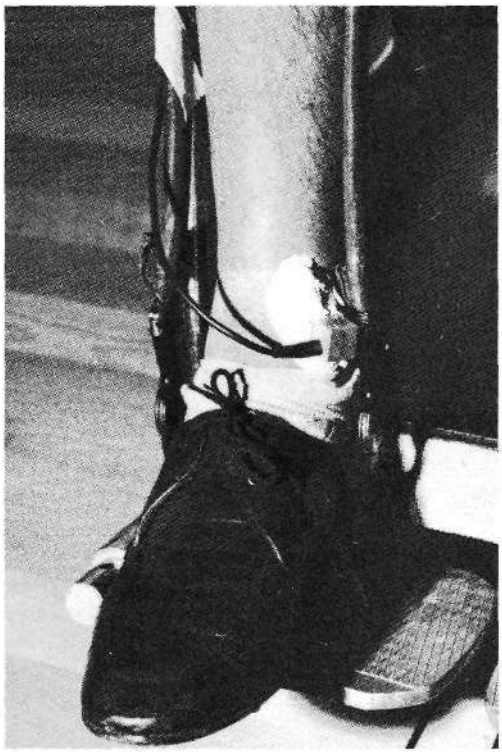

Fig. 1. Accelerometer fixed in a plaster holder and taped to the medial malleolus of the right leg, to create bony contact.

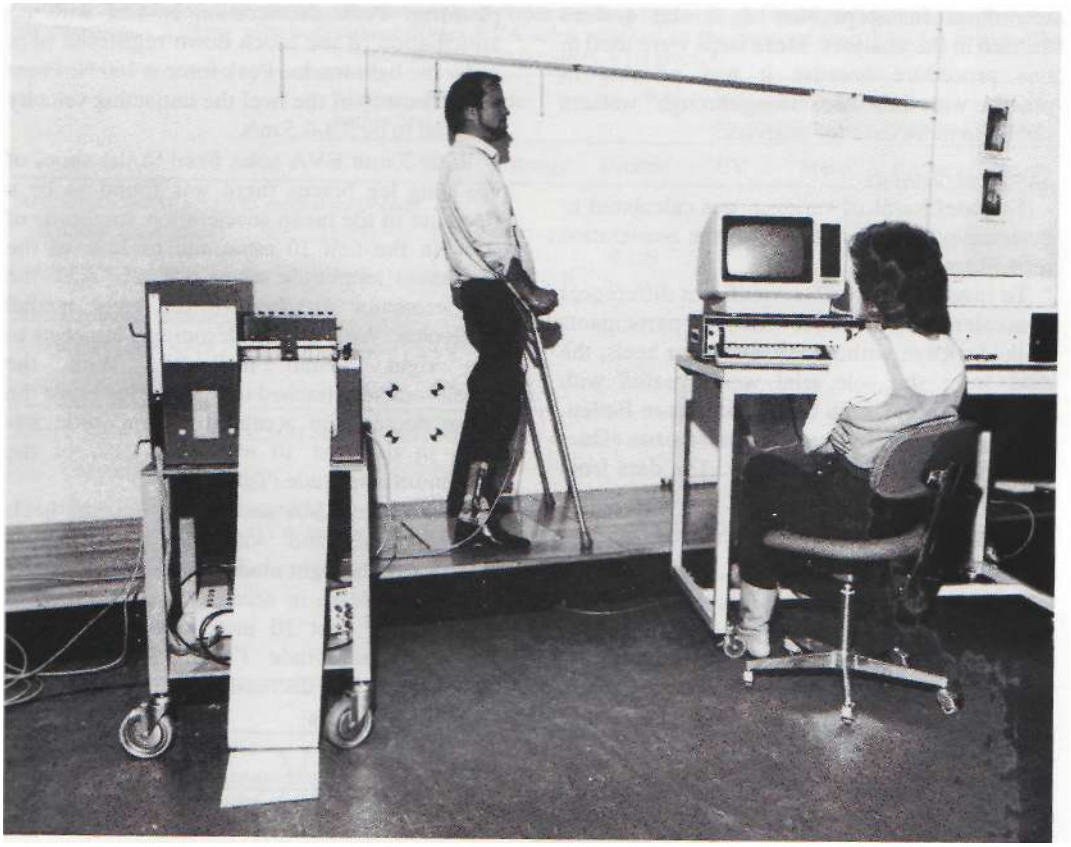

Fig. 2. Paraplegic walking with long leg braces and forearm crutches along a wooden walkway during an experiment with accelerometer attached to the medial malleolus.

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

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

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

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

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

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

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

Statistical methods

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

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

Results

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

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

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

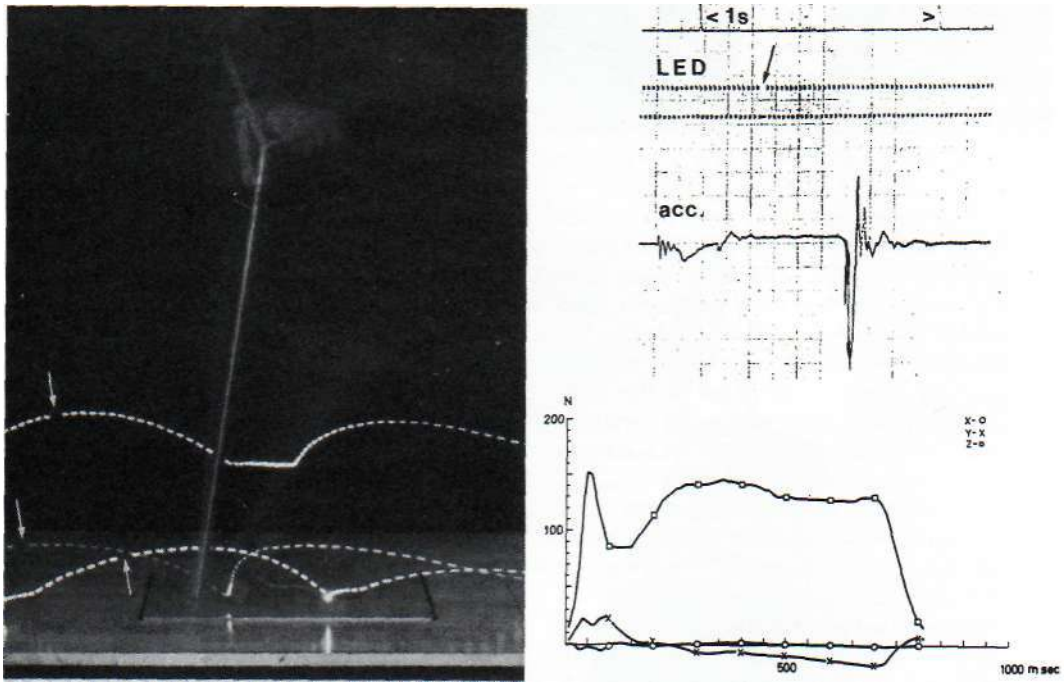

Fig. 4. Left, still photo of paraplegic walking on a walk way with built in force-platform. Light emitting diodes from three light tracks showing the trajectory of the shank, the heel, and the forefoot. Arrows indicate the missing signal and thereby the time link between the three tracks. Top right,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

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

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

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

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

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

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

significant (in all instances $p<0.01$}.

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

Discussion

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

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

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

Considering the sole trial with the major reductions in accelerations registered at the caliper it is noticeable how large were the reductions found in the accelerations recorded from the medial malleolus in the EVA heel study. Thus the results indicate that an even

larger reduction in accelerations up through the long leg braces might be obtained by building EVA into the heels.

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

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

REFERENCES

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

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

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

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

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

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

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

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

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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.

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Introduction

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

Amputee gait has been evaluated both qualitatively (Gonzalez et al., 1974; Urban 1973; Waters et al, 1976; Kegel et al., 1981) and quantitatively in the literature. Quantitative research can be further subdivided into kinematic and kinetic studies. Many studies have presented descriptions of amputee gait based on kinematic measures (Enoka et al., 1982; Hannah and Morrison, 1984; Zuniga et al., 1972). James and Oberg (1973) and Murray et al. (1981), in similar studies of above-knee amputees, found

significant differences in velocity, cadence, gait cycle, and stride length when their study group was compared to normal subjects. These authors noted that the step length of the prosthetic leg tended to be longer than that of the contralateral leg. Breakey (1976), in studies of below-knee amputees, reported that the stance phase of gait was longer in the normal limb and shorter in the amputated limb. Robinson et al. (1977) obtained time distance and accelerometer data from below-knee amputees. The subjects took longer steps more quickly on their prosthetic side and the resulting gait was described as asymmetric. Hershler and Milner (1980) also found asymmetry between the unaffected and the amputated side when looking at the variation of hip angle and knee angle throughout all phases of gait in above-knee amputees. Skinner and Effeney (1985) noted that this asymmetry of motion increases the excursion of die centre of mass during each cycle and thereby increases the energy cost of ambulation. These kinematic studies suggest that amputees demonstrate an asymmetric gait pattern. However this observation has not been verified using quantitative methods to determine the degree of symmetry.

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

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

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

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

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

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

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

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

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

Methodology

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

Data collection

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

A four camera WATSMART kinematic data acquisition system was used to acquire the two-dimensional positions of five anatomical landmarks on the left/contralateral lower limb and three anatomical landmarks on the right/prosthetic side (50 hertz sampling rate). Eight one centimetre diameter disks containing 3 infra-red emitting diodes (IREDs) were placed over the anatomical landmarks. These landmarks approximated the positions of the anatomical joint centres of the hip, knee and ankle, and the proximal and distal ends of the foot segment on the left/contralateral lower extremity. IREDs were placed over the anatomical joint centres of the hip, knee and ankle of the non-amputee's right lower limb. The amputee's prosthetic lower limb was treated in a similar manner with the

anatomical joint centres of the hip and knee located and a third IRED placed distally bisecting the longitudinal axis of the prosthetic shank in the sagittal plane. The cameras were placed perpendicular to the sagittal motion of the lower limbs. The subject was familiarized with the testing area in order to promote natural performances during data collection. Walking trials lasted 6-7 seconds and approximately 20 trials per subject were carried out.

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

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

Statistical analysis

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

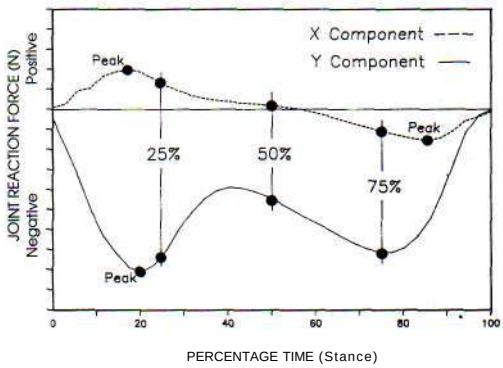

Fig. 1. Joint reaction force dependent variables.

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

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

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

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

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

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

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

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

Discussion

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

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

Lower limb symmetry

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

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

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

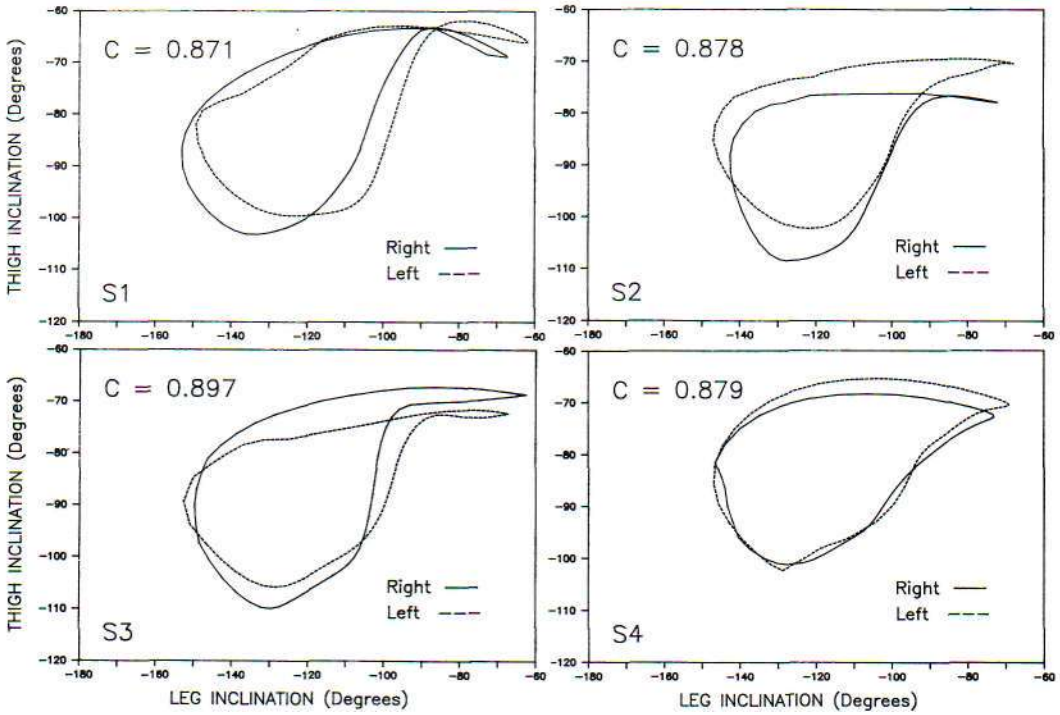

Fig. 2. Bilateral angle-angle plots of leg and thigh for one stride by non-amputees SI, S2, S3 and S4.

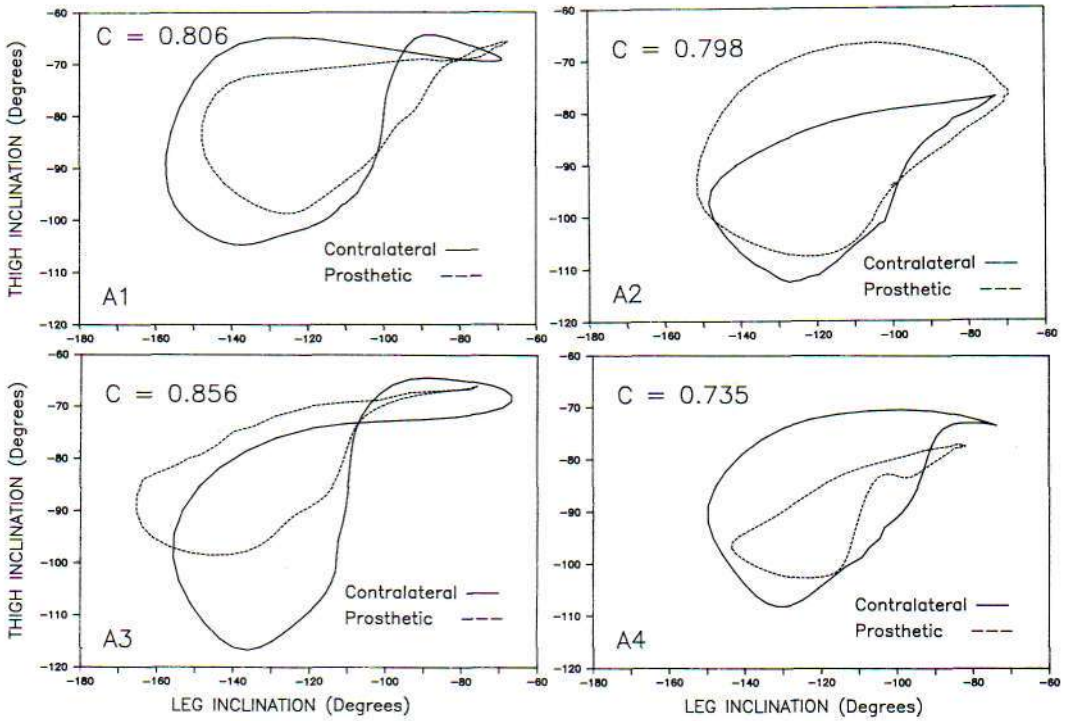

Fig. 3. Bilateral angle-angle plots of leg and thigh for one stride by amputees A1, A2, A3 and A4.

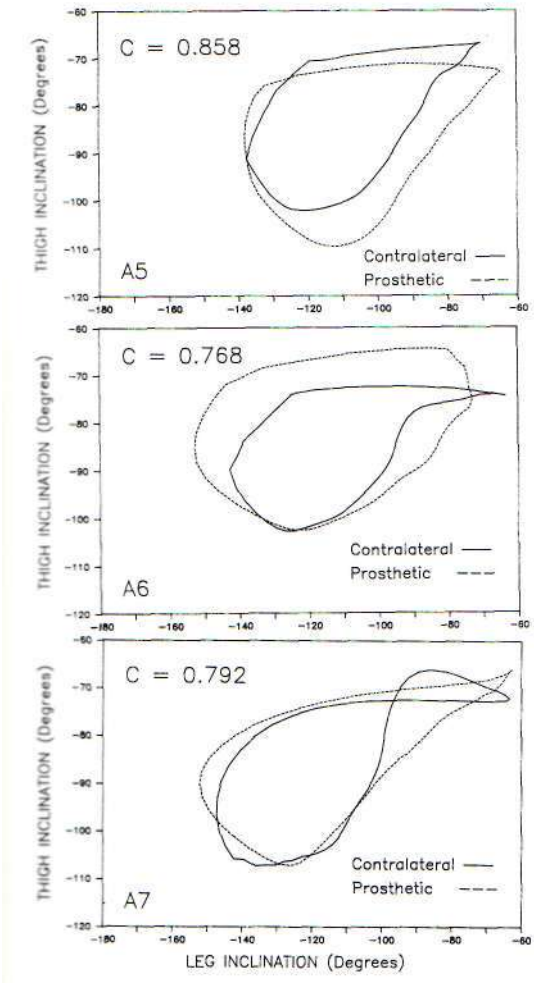

Fig. 4. Bilateral angle-angle plots of leg and thigh for one stride by amputees A5, A6 and A7.

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

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

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

Joint kinetics

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

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

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

Mean = 0.833 S.D. = 0.032

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

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

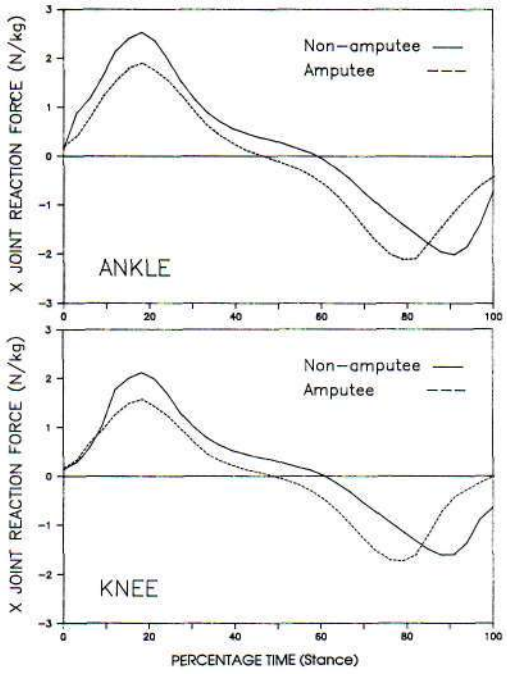

Fig. 5. Top, ankle horizontal joint reaction force, during stance, for an amputee and non-amputee subject. Bottom, knee horizontal joint reaction force, during stance, for an amputee and non-amputee subject.

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

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

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

Mean = 0.799 S.D. = 0.054

TABLE 6 — Walking Velocity (ms^{-1}).

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

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

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

Conclusions

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

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REFERENCES

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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Lower limb amputee survival

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Abstract

A total of 1710 primary amputees have been studied over a 25 year period and their survival time has been calculated. These were all consecutive primary lower limb amputees admitted to the Dundee Limb Fitting Centre during the period 1965–1989. Overall, the median survival was 4 yr 9 mth for the below-knee amputee (1019 patients) and 4 yr 3 mth for the above-knee amputee (586 patients). The vascular related amputees had an overall median survival of 4 yr.

In the two decades 1970–1979 and 1980–1989 there were significant differences between the survival time of the below-knee and above-knee amputee. The survival of the amputee has increased during the two decades from 3 yr 6 mth to 6 yr 6 mth ($p>0.001$). For the first decade male above-knee and male below-knee amputee median survival was 3 yr 1 mth and 3 yr 11 mth respectively and for the second the survival was 5 yr 9 mth and 6 yr 11 mth for these levels of amputation.

For 1970–1979 no significant differences were found between male and female peripheral vascular disease (PVD) and diabetes mellitus related amputee survival. For 1980–1989 significant differences were found between PVD related male above-knee amputees (3 yr 10 mth) and male below-knee amputees (6 yr 7 mth) ($p>0.01$). Similar results were found for the female patients.

Operative mortality was found to be 5% over the period 1975–1989 which compared favourably with previous studies.

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Table 1. Population included in the study (1710 patients)

Year of amputation	1965–1969	1970–1979	1980–1989
Female	94	263	339
Male	148	396	470
Total	242	659	809
Diagnosis leading to amputation			
PVD	127	403	489
DM	66	185	194
Other e.g., Tumour, trauma infection	49	71	126
Level of amputation			
HQ	0	0	2
TH	6	7	11
AK	111	190	261
TK	29	33	21
BK	81	392	491
Symes	14	29	23
Partial foot	1	8	—

Overall prosthesis fitting 87%.
Wheelchairs supplied 22%.

Table 2. Tayside survival figures for 70 year old people.

	1974	1978	1980	1988
Male	9.37 yr	9.70 yr	9.46 yr	10.34 yr
Female	12.57 yr	12.88 yr	12.93 yr	13.40 yr

Source: Registrar General for Scotland (Personal Communication).

Note: Figures unavailable before 1974 — new district boundaries introduced in 1973–74.

Introduction

The majority of amputations performed in western society are on elderly people. The main condition leading to amputation is some form of peripheral vascular disease (PVD) (Table 1). The average age of such an amputee is reported as about 70 years old (Murdoch, 1988).

The survival of an average 70 year old male from Dundee is 10.34 years (Registrar General for Scotland, 1988) (Table 2) but the survival of an amputee from the similar group has been reported from two years to five years (Barnett *et al.*, 1976; Whitehouse *et al.*, 1968) (Table 3). The survival of patients after amputation surgery was not reported to have changed over the past thirty years but is markedly less than that of their peers, whose life expectancy has increased.

Table 3. Amputee mortality

Silbert (1952)	3 yr 61% survival
Smith (1956)	5 yr 60% mortality
Hansson (1964)	2 yr 58% mortality
Whitehouse *et al.* (1968)	5 yr 50% mortality
Kolind-Sørensen (1974)	5 yr 50% mortality
Barnett *et al.* (1976)	2 yr 50% mortality (98 cases)
Ebskov and Josephsen (1980)	4 yr 22.5% survival (2,029 patients)
Finch *et al.* (1980)	2 yr 45% mortality 4 yr 25% mortality
Mandrup-Poulsen and Jensen (1982)	1 yr for AK amputation — 54% mortality 1 yr for BK amputation — 17% mortality
Jamieson and Ruckley (1983)	No 5 yr survivors (73 cases)
Cumming *et al.* (1987)	22 month mean survival BK amputees
Harris *et al.* (1988)	1966–1971 3 yr 2 mth 1978–1982 3 yr 11 mth considered N/A (52% = 3 yr) 116 cases (1966–71) 189 cases (1978–82)

Review of the literature shows that operative mortality, defined as death within one month of amputation surgery, varies from 1.7% to 31% (Table 4). Some of the publications also highlight that mortality following above-knee (AK) surgery is higher than surgery following below-knee (BK) amputation.

The limited survival of these amputees emphasises the importance of early fitting and speedy rehabilitation. It is important that the clinic team should make every effort to make these patients mobile and independent as soon as possible so that the remaining short life of an amputee is comfortable and worth living. This is best achieved by developing a comprehensive amputee service (Murdoch, 1988).

This paper studies 1,710 primary amputees admitted to Dundee Limb Fitting Centre (DLFC) over a twenty-five year period and reviews the overall survival patterns. The study also compares the differences between decades 1970 to 1979 and 1980 to 1989 in detail.

Table 4. Operative mortality within 1 month of operation

KCH — 2% in the past 15 years DLFC — 3%	} 5% overall
Shumacker and Moore (1951)	6.6%
Silbert and Haimovici (1954)	7%
Smith (1956)	12%
Schlitt and Serlin (1960)	13.6%
Otteman and Stahlgren (1965)	29.7%
Olejniczak (1967)	24.1%
Burgess *et al.* (1971)	8%
Persson and Sunden (1971)	11.9%
Kahn *et al.* (1974)	9%
Barnett *et al.* (1976)	15%
Finch *et al.* (1980)	15%
Houston *et al.* (1980)	15–37% g (Lit. Review)
Barber *et al.* (1983)	(31% AK; 7.2% BK)
Jamieson and Ruckley (1983)	20%
Mann and Bisset (1983)	1.7%
Pohjolainen and Alaranta (1988)	(27%–3 mth)
Harris *et al.* (1988)	(5% AK; 6% BK)
Pohjolainen *et al.* (1989)	(25.5%–2 mth)

KCH — Kings Cross Hospital, Dundee — The Operating Hospital.

DLFC — Dundee Limb Fitting Centre, Dundee, the rehabilitation facility, to where patients were transferred within an average of 7 days post-operatively.

Method

All patient records are kept in the Centre and are updated regularly.

Prospective information on all primary amputees admitted to DLFC has been recorded, initially on specially designed charts (1965–1981) and since 1982 on an enhanced data sheet. The information has been transferred to an Olivetti M24 PC using dBase III+ (1985).

The information for operative mortality, defined as death within one month of surgery in Tayside, was collected from Scottish Medical Records Form 1 (SMR 1) and from the DLFC data base. The SMR1 form is completed by secretarial staff for al patients on discharge from Scottish hospitals. The SMR1 form, along with other information records the following:

1. reason for admission;
2. date of admission;
3. date of discharge.

The record of the date of death has been obtained from a variety of sources which includes patients' relatives, review of the local newspapers, contact with the general practitioner or hospital doctor looking after the patient prior to death, the Primary Care Division of Tayside Health Board and occasionally by visiting the Registrars Office in Edinburgh where death certificates are recorded for the whole of Scotland. A check was made in June 1990 and survival status in this study was recorded at that date. Those who have moved away from this catchment area which includes Tayside and North Fife region of Scotland or have been lost to follow-up have been excluded from the study.

Survival distributions were compared graphically between subgroups by plotting the survival functions against time. This plot shows the proportion still alive at subsequent times, out of an initial cohort of patients. Information from 'censored observations', that is, patients who are still alive at the date of the study and whose survival times are known only up to that point, was incorporated using the Kaplan-Meier method (see for example Pocock (1983)). A test of the differences between survival distributions in different subgroups of patients used the logrank test. Statistical analysis of data was carried out using Nanostat (1989) computer software.

Results

Some 1,710 patients who had primary lower limb amputations between 15.5.65 and 31.12.89 were included in the study, all being admitted to the DLFC following surgery either under the Tayside Amputation Service, or the surgical units in North Fife. The average age of a patient undergoing amputation was 70 years.

Table 1 indicates the population studied. This includes the sex of the patients in the study, the principal cause of the amputation and the ultimate level of the amputation.

Table 2 shows Tayside survival figures for males and females from 1974 — 1988. The survival has continued to increase from 1974 to 1988, in males from 9.37 yr to 10.34 yr and in females from 12.57 yr to 13.40 yr.

Table 3 shows amputee mortality as reported in the literature. This ranges from 54% mortality for AK amputees at 1 year, to 60% mortality at 5 years.

Table 4 lists Tayside's operative mortality along with other reports from the literature. Dundee mortality at 5% compares favourably with that reported in several publications and is better than some at 31% for AK patients in one paper (Barber *et al.*, 1983).

Table 5 presents the median survival of the whole group studied over 25 years irrespective of causes of amputation. Significant differences were found between BK (4 yr 9 mth) and AK (4 yr 3 mth) ($p<0.004$). Comparing all amputees in the two decades, significant differences were found with 1970–79 (3 yr 6 mth) and 1980–89 (6 yr 6 mth) ($p<0.001$). This shows that the

Table 5. Median survival of the whole 25 year study all diagnoses.

Above-knee		
Male	3 yr 6 mth	4 yr 3 mth
Female	4 yr 8 mth	
Below-knee		
Male	4 yr 8 mth	4 yr 9 mth
Female	5 yr 8 mth	
Below-knee	4 yr 9 mth	$p < 0.004$
Above-knee	4 yr 3 mth	

1970–1979 3 yr 6 mth
1980–1989 6 yr 6 mth } $p < 0.001$

Overall operative mortality (within 1 month of amputation) 5%.

Table 6. Median mortality figures for two decades.

All diagnoses	1970–1979 3 yr 6 mth		1980–1989 6 yr 6 mth	
Male	3 yr 1 mth		5 yr 9 mth	
Female	3 yr 11 mth		6 yr 11 mth	
	DM	PVD	DM	PVD
Above-knee				
Male	2 yr 4 mth	2 yr 6 mth	4 yr 6 mth	3 yr 10 mth
Female	11 mth	3 yr 3 mth	74% (5 yr survival)	6 yr 5 mth
Below-knee				
Male	2 yr 10 mth	3 yr 10 mth	4 yr 0 mth	6 yr 7 mth
Female	2 yr 5 mth	4 yr 0 mth	53% (5 yr survival)	6 yr 11 mth

For 1970–1979 no significant differences were found between male and female PVD and diabetes mellitus or levels of amputation with regard to survival.

For 1980–1989 significant differences were found between PVD related male above-knee and male below-knee amputees ($p < 0.01$). No differences found between the level or sex of diabetic amputees, nor PVD related female cases.

overall survival of the amputees had significantly increased over the period. The BK patients survived longer than the AK amputees.

Table 6 shows the median survival of two decades 1970–1979 and 1980–1989 in relation to level of amputation and main causal conditions, i.e. PVD (without diabetes) and diabetes related amputations. As stated above the survival of all patients has been found to have increased significantly over the study period, but in addition there was found significant differences between the PVD related male AK amputees and the male BK amputees, ($p<0.01$) with survivals of 3 yr 10 mth and 6 yr 7 mth respectively in the latter decade.

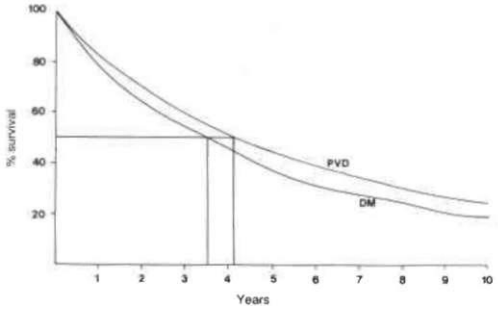
Table 7 shows median survival for PVD with and without diabetes mellitus related amputation. Statistically significant differences are seen between PVD patients without diabetes and those with diabetes mellitus related amputation respectively (4 yr 2 mth) and (3 yr 8 mth) ($p<0.006$). The patients with PVD (without diabetes) related amputations surviving longer than those who had a diabetic related amputation. In addition there were significant differences between male AK (3 yr 6 mth) and male BK patients (4 yr 8 mth) ($p<0.03$).

No significant differences were found between the ages of the patients during the periods of study.

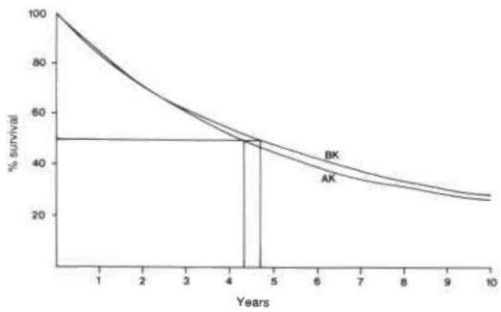
Graph 1 is the survival curve for 10 years of

Table 7. Median survival for the whole 25 year study.

All cases			
	Above-knee male (338)	– 3 yr 8 mth	$p < 0.03$
	Below-knee male (559)	– 4 yr 8 mth	
All vascular cases (1464 overall)		– 4 yr 0 mth	
	Above-knee	– 3 yr 6 mth	
	Below-knee	– 4 yr 2 mth	
	Vascular (non-diabetics) (1019)	– 4 yr 2 mth	$p < 0.006$
	Diabetic cases (445)	– 3 yr 8 mth	



Graph 1. Survival curve for 10 years of 25 year study for PVD with and without diabetes mellitus ($p < 0.006$).



Graph 3. Survival curve for 10 years of 25 year study divided between below-knee and above-knee.

the the 25 year study divided into PVD with and without diabetes mellitus related amputations, showing the increased survival of the PVD amputee over the diabetic patient.

Graph 2 shows the survival curve for 10 years of the 25 year study, for all cases, divided into two decades 1970–1979; 1980–1989. It clearly shows the increased longevity over the two decades.

Graph 3 gives the survival curves for 10 years of the 25 year study of BK and AK amputees. It shows the greater survival of the BK amputee over the AK case.

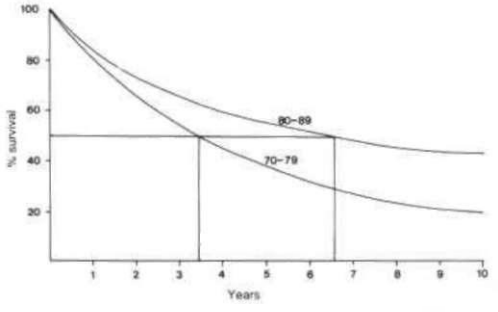
Graph 4 covers the two decades showing the BK and AK amputees for all cases, demonstrating the increased survival for the two decades for both levels of amputation, the BK amputees having a greater survival than the AK in both decades.

Discussion

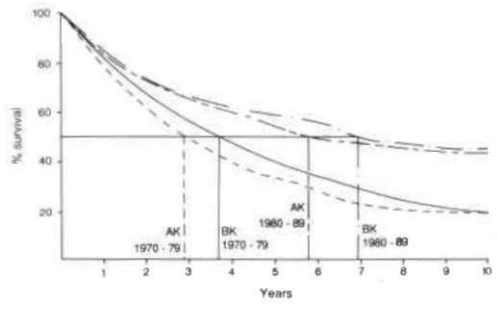
The survival of lower limb amputees has been reported to be from two to five years by various authors (Barnett *et al.*, 1976; Kolind-Sørensen, 1974; Whitehouse *et al.*, 1968) (Table 3). Jamieson and Ruckley (1983) even reported no survivors at 5 years. This study reports on the survival of 1,710 lower limb amputees admitted to DLFC over the last 25 years (Table 1). This prospective study presents the overall survival rate of all patient groups.

The overall survival of patients who had an amputation as a result of PVD was found to be 4 yr (50% survival) (Table 7). When the vascular cases are divided into, those who had diabetes and those who did not, then the survival rate is found to be 4 yr 2 mth for PVD and 3 yr 8 mth for diabetes mellitus (Graph 1).

This difference in survival between the two groups of amputees was significant ($p<0.006$) and is similar to the reports by Smith (1956), (diabetic patients only), but less than those reported by Kolind-Sørensen (1974) and Whitehouse *et al.*, (1968). Silbert (1952) reported a 61% survival at 3 years and 41% at 5 years in a study of 294 diabetic amputees. Ebskov and Josephsen (1980) reported 22.5% survival at 4 years for 2,029 patients which is much less than this group, Pohjolainen and Alaranta (1988) reported 27% mortality at 3



Graph 2. Survival curve for 10 years of 25 year study for all cases divided into two decades, 1970–79 and 1980–89.



Graph 4. Survival curve for all cases divided into below-knee and above-knee for the two decades 1970–79 and 1980–89 ($p < 0.004$).

month and later Pohjolainen *et al.*, (1989) reported a 25.5% amputee mortality within 2 month of amputation (Table 4). Cumming *et al.*, (1987) reported for 48 BK amputees a 22 mth survival only. The diabetic patients in this study have been shown to survive a shorter time than those with only PVD but much longer than in many reported studies.

Review of the Tayside amputees in the decades 1970–1979 and 1980–1989 showed very different and significant results. For all diagnoses during the decade 1970–1979 amputees had a 50% survival of 3 yr 6 mth whereas in 1980–1989 they had a survival of 6 yr 6 mth ($p<0.001$) (Graph 2 and Table 6). This increasing pattern is in keeping with that reported by Harris *et al.*, (1988) who reviewed 116 cases from 1966 to 1971 with a mortality of 3 yr 2 mth and 189 cases over a period of 1978–1982 with a mortality of 3 yr 11 mth (Table 3). Although these figures were not reported as being significant there is the suggestion of a trend towards increasing longevity. The increase in longevity reported in this paper is quite marked and dramatic. There was also an increase in longevity of all Tayside 70 year olds (Table 2). For the period 1970–79 no significant difference was found between the suvival of male and female with PVD or diabetes mellitus. There were however differences found in the 1980–1989 group. The change cannot be explained by differences in the age distributions between the study groups alone, since it was found that there was no significant difference between the ages of the patients in the decades.

For the whole 25 year study the BK amputees survived a median time of 4 yr 9 mth whereas AK amputees only survived a median time of 4 yr 3 mth (Graph 3 and Table 5) ($p<0.004$). Reviewing the survival of the principal levels of amputation shows that there was an increase in the survival of the patients over the two decades (Graph 4).

During the period 1980–1989 the BK male vascular non-diabetic amputee survived 6 yr 7 mth whereas the AK male amputee with a similar pathology survived 3 yr 10 mth ($p<0.01$) (Table 6). Overall a similar survival has found in the whole series 1965–1989 with male BK amputees surviving 4 yr 8 mth and male AK amputees surviving 3 yr 8 mth ($p<0.03$) (Table 7).

The difference presumably reflects the degree and widespread nature of vascular involvement, with AK amputees having significantly worse vascular disease than BK, and it might be assumed worse generalised disease. Associated diabetic cases have significantly shorter survival than the pure PVD associated amputee, 3 yr 8 mth as compared with 4 yr 2 mth ($p<0.006$) (Table 7). This is similar to other reported findings.

Operative mortality (Table 4) is reported as high as 31% for AK (Barber *et al.*, 1983) and as low as 1.7% (Mann and Bisset, 1983). In Dundee with the amputation service providing total patient care, the operative mortality has been 5%. This is still too high but possibly related to the health of the elderly population who require amputation. Troup (1976) reported that many patients had a concurrent disease and this reflects the overall ill-health of the patients. In Dundee in 1983 there were 67% of the amputees admitted to the DLFC with at least one other significant pathology (Stewart, 1985). In the literature the incidence of concurrent disease varies from 43% (Moffat *et al.*, 1981) to 100% (Anderson *et al.*, 1967). Associated cardiac disease is also commonly reported, Malone *et al.* (1979) reported this pathology as occurring in 66% of 133 cases and Kavanagh and Shepherd (1973) recorded 100% of 62 cases as having cardio-respiratory disease. In the previously reported Dundee study it was found to be 30% (Stewart, 1985).

The long term survival of Tayside people surviving to 70 years of age has also changed. In 1974 male survival was 9.37 yr while in 1988 survival had risen to 10.34 yr. Female survival has also risen and is considerably more than male peers. In 1974 female survival was 12.57 yr while in 1988 it had risen to 13.4 yr (Table 2), similarly survival of amputees has also increased but this rise was found to be greater than that of their peers.

It was previously thought that the amputee had a significantly reduced survival but it has been found in this study that in the last decade the survival of the amputee has increased significantly. This increase is much more than the general improvement in the corresponding age group as mentioned earlier. The importance of this is that these patients must be fitted with prostheses wherever possible and offered comprehensive rehabilitation so that the quality of life can be maintained. In this

study 87% of amputees have been successfully fitted with prostheses with 22% of them being supplied with a wheelchair (Table 1).

This series also demonstrates that a high proportion of BK amputations is possible with a well organised co-ordinated amputee service (Murdoch and Donovan, 1988). It was found that male BK amputees had a signficantly longer survival than male AK amputees where the causal condition of amputation was non-diabetic vascular disease, but the survival was less than that of his peer. Diabetic amputees were less fortunate. This presumably relates to the severity of the disease and is often associated with the generalised atherosclerosis. It is likely that the longevity of the BK amputee relates to the lesser vascular involvement than those who require an AK amputation.

Resources need to be available to ensure that all these amputees obtain satisfactory prosthetic care to enjoy an active life in society.

Acknowledgements

Thanks are due to the late Miss M. Wood, former Matron of DLFC and Professor G. Murdoch who initiated the data collection system when DLFC was opened in September, 1965. We also thank Mrs. F. Clark for typing these scripts and Mr. S. Scott for producing the graphs, and the Tayside Health Board for providing the research grant to purchase the Olivetti Computer and dBase III+ Software.

REFERENCES

ANDERSON AD, CUMMINGS V, LEVINE SL, KRAUS A (1967). The use of lower extremity prosthetic limbs by elderly patients. *Arch Phys Med Rehabil* **48,** 533–538.

BARBER GG, MCPHAIL NV, SCOBIE TK, BRENNAN MCD, ELISS CC (1983). A prospective study of lower limb amputations. *Can J Surg* **26,** 339–341.

BARNETT AJ, TWIST E, BALFE A (1976). Lower limb amputation in agreed hospital: a comparative review. *Med J Aust* 3 July **79,** 14–18.

BURGESS EM, ROMANO RL (1971). Amputation of leg for peripheral vascular insufficiency. *J Bone Jt Surg* **53A,** 874–890.

CUMMING JAR, JAIN AS, WALKER WF, SPENCE VA, STEWART C, MURDOCH G (1987). Fate of the vascular patient after below-knee amputation. Lancet 12 Sept, 613–615.

DBASE III+ (1985). Ashton – Tate.

EBSKOV E, JOSEPHSEN P (1980). Incidence of reamputation and death after gangrene of lower extremity. *Prosthet Orthot Int* **4,** 77–80.

FINCH DRA, MACDOUGAL M, TIBBS DJ, MORRIS J (1980). Amputation for vascular disease: the experience of a peripheral vascular unit. *Br J Surg* **67,** 233–237.

HARRIS JP, PAGE S, ENGLUND R, MAY J (1988). Is the outlook for the vascular amputee improved by striving to preserve the knee? *J Cardiovasc Surg* **29,** 741–745.

HOUSTON CC, BIVINS BA, ERNST CB, GRIFFEN WO (1980). Morbid implications of above-knee amputations. Report of a series and review of the literature. *Arch Surg* **115,** 165–167.

JAMIESON MG, RUCKLEY CV (1983). Amputation for peripheral vascular disease in a general surgical unit. *J R Coll Surg Edinb* **28,** 46–50.

KAHN O, WAGNER N, BESSMAN AN (1974). Mortality of diabetic patients treated surgically for lower limb infection and/or gangrene. *Diabetes* **23,** 287–292.

KAVANAGH T, SHEPHERD RJ (1973). The application of exercise testing to the elderly amputee. *Can Med Assoc J* **108,** 314–317.

KOLIND-SØRENSEN V (1974). Follow-up of lower knee amputees *Acta Orthop Scand* **45,** 97–104.

MALONE J, MOORE W, GOLDSTONE J, MALONE, S (1979). Therapeutic and ergonomic impact of a modern amputation programme. *Bull Prosthet Res* **10 (32),** 7–17.

MANN RAM, BISSET WIK (1983). Anaesthesia for lower limb amputation: a comparison of spinal analgesia and general anaesthesia in the elderly. *Anaesthesia* **38,** 1185–1191.

MANDRUP-POULSEN T, JENSEN JS (1982). Mortality after major amputation following gangrene of the lower limb. *Acta Orthop Scand* **53,** 879–884.

MOFFAT LEF, RAINEY JB, MCPHERSON AIS (1987). Major Lower Amputation Reassessment. *J R Col Surg Edinb,* **26,** 50–53.

MURDOCH G, DONOVAN RG (1988). Amputation surgery and lower limb prosthetics. Edited by Murdoch G, Donovan RG. — London: Blackwell Scientific Publications, Edinburgh.

NANOSTAT (1989). — London: Alphabridge Ltd.

OLEJNICZAK S (1967). Summary of leg amputations and prosthetic replacement at Wayne County General Hospital during the five year period 1961 through 1965. *Mich Med* **66,** 723–726.

OTTEMAN MG, STAHLGREN LH (1965). Evaluation of factors which influence mortality and morbidity following lower extremity amputation for arteriosclerosis. *Surg Gynecol Obstet* **120,** 1217–1220.

Persson BM, Sunden G (1971). Amputation pa underbenet vid aldersgangran. *Nord Med* **86**, 1045–1049.

Pocock SJ (1983). Clinical trials — a practical approach. — Chichester: Wiley p 222–224.

Pohjolainen T, Alaranta H (1988). Lower limb amputations in Southern Finland (1984–1985). *Prosthet Orthot Int* **12**, 9–18.

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

Registrar General for Scotland — Annual Report (1987). — London: HMSO No 134: (and Personal Communication).

Registrar General for Scotland — Annual Report (1988). — London: HMSO.

Schlitt RJ, Serlin, O (1960). Lower extremity amputation in peripheral vascular disease. *Am J Surg* **100**, 682–689.

Scottish Health Statistics (1988). — Edinburgh: Information and Statistics Division, Common Services Agency for the Scottish Health Service.

Shumacker H, Moore, T (1951). Leg and thigh amputation in obliterative arterial disease. *Arch Surg* **63**, 458.

Silbert S (1952). Amputation of the lower extremity in diabetes mellitus: a follow up study of 294 cases. *Diabetes* **1**, 297–299.

Silbert S, Haimovici H (1954). Criteria for selection of level of amputation of ischaemic gangrene. *JAMA* **155**, 1554–1558.

Smith B (1956). A twenty year follow-up in 50 below knee amputations for gangrene in diabetes. *Surg Gynecol Obstet* **103**, 625–630.

Stewart CPU (1985). An investigation into the conditions affecting the level of activity achieved by primary amputees in the first six months after amputation. MD Thesis. — Dundee: University of Dundee.

Troup IM (1976). The assessment of amputation level on the management of stump environment in peripheral vascular disease. Dissertation for D Med Rehabil (Personal Communication).

Whitehouse FN, Jurgensen C, Block MA (1968). The later life of the diabetic amputee: another look at the fate of the second leg. *Diabetes* **17**, 520–521.

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Running gait impulse asymmetries in below-knee amputees

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Abstract

In running, large gait asymmetry is expected due to the inability of the foot prosthesis to comply with the kinematic demands and produce a powerful plantarflexion moment. In this work, interlimb asymmetry in below-knee (BK) amputee running gait was assessed for one rigid and three flexible keel prostheses, using vertical and anteroposterior ground reaction forces and respective impulses. Nine BK amputees and 6 controls participated in this study. The running speed was monitored by two light sensitive detectors while the ground reaction forces were measured with a Kistler force plate. Between the prosthetic side and the sound limb the impulse indicator showed greater asymmetry than the force. Interlimb asymmetry was very much present in all types of prosthesis tested but is less pronounced in the flexible keel prostheses. In the latter, the asymmetry may be associated with the force-time history modulation rather than its magnitude alone. Generally, the impulses better describe interlimb asymmetry and the forces allow a greater discrimination between prosthetic foot types.

Introduction

Gait asymmetry has been reported during walking in unilateral below-knee (BK) amputees (Breakey, 1976; Culham *et al.*, 1986; Doane and Holt, 1983). It appears to be linked with an overloading of the musculoskeletal system leading to degenerative changes in the lumbar spine and knees (Burke *et al.*, 1978; Perry, 1975). Temporal, kinematic and kinetic asymmetry indices have been developed to describe the differences between the amputee's prosthetic and sound limbs (Hurley *et al.*, 1990; Seliktar and Mizrahi 1986; Skinner and Effeney, 1985; Winter and Sienko, 1988). Among these, Seliktar and Mizrahi (1986) concluded that the peak vertical forces and their ratios are not meaningful indicators in representing locomotor problems in BK walking gait, while the anteroposterior (AP) force perturbations are most useful in reflecting instabilities arising from the prosthesis. The impulses were, however, sensitive to the quality of gait.

With more and more amputees regularly taking part in strenuous recreational sports in which running is often an important activity, Brouwer *et al.*, (1989) suspect larger interlimb asymmetries. This is due in part to the inability of the foot prosthesis to satisfy the kinematic demands accompanying the large ankle excursion as well as to the lack of the powerful plantarflexion moment required for a strong propulsion. Although the running gait patterns of BK amputees have been described by Enoka *et al.* (1982) and Miller *et al.* (1984; 1987), the impulse parameters have not been extensively used to highlight the effect of different types of foot prostheses, on running gait asymmetry.

It was the purpose of this study to

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demonstrate that the impulse parameter was more appropriate than the ground reaction force in describing interlimb asymmetry in BK amputee running gait and to determine the effect of rigid and flexible keel foot prostheses on gait asymmetry. Emphasis was placed on the general functional characteristics of the prosthesis rather than on the individual type of foot.

Subjects and methods

Nine BK amputees and 6 control subjects participated in this study. In 5 cases, the amputation was consequent to either bone cancer or trauma while for the remaining 4 cases, surgery was carried out to correct congenital malformations. There were 3 female and 6 male amputees with a mean age of 16.4 years ±3.8 years and a weight of 58.6 kg ±12.9 kg. The control group, consisting of one female and 5 male subjects had a mean age and mass of 22.2 years ±3.5 years and 72.1 kg ± 12.4 kg respectively. The forces and impulses were corrected to take into account the weight difference between the amputee and control groups.

Four different types of prosthetic feet were tested. Five amputees had a SACH foot representing the rigid keel group. In the flexible keel group, 2 amputees were wearing the Seattle foot while the 2 others had a SAFE foot. Two of them were also fitted with a modified version of the Flat-Spring foot (FSF) prosthesis (Allard *et al.*, 1988). Because of its form, the SAFE foot has been included in the flexible keel group just as Wing *et al.*, (1989) did; although, Michael (1987) does not consider it to be an "energy storing" foot.

The least time since amputation was 2 years and all amputees had been wearing a PTB socket fitted with either a SACH, Seattle or SAFE foot for at least 2 years. The FSF prosthesis was worn only for the duration of the gait trials.

Physical examination revealed no important orthopaedic abnormality other than amputation. Muscle function was normal and all amputees had a healthy stump.

The ground reaction forces were sampled at 600 Hz using a Kistler force plate while the running speed was monitored by two "Speedtract" light sensitive detectors located at 3m on each side of the force plate along the 25m walkway. The subjects, wearing sport shoes, had about 7 trial runs to adjust their running speed to fall between 2.8m/s and 3.2m/s. Five trials were then recorded for the sound and affected limbs.

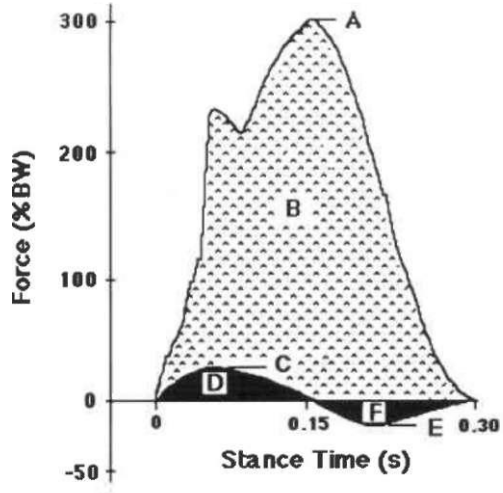

Fig. 1. Below-knee amputee ground reaction forces normalised with respect to body weight as a function of stance time for the prosthetic limb taken at a speed of 3.0m/s. A represents the maximum thrust, B the vertical impulse, C the maximum braking force, D the braking impulse, E the maximum push-off force and F the push-off impulse.

For a BK amputee running at 3.0m/s, Figure 1 illustrates typical vertical and AP ground reaction forces, expressed in percent of body weight (%BW), as a function of the stance time. Six values were extracted from these curves. From the vertical ground reaction force the maximum thrust, A, corresponded to the peak force developed by the amputee during the stance phase of running while the vertical impulse, B, was the total area under this curve. The maximum braking force, C, and impulse, D, associated with the deceleration of the body centre of mass were taken from the AP ground reaction force. The last two parameters were the push-off force, E, and impulse, F, partly representing the acceleration of the body centre of mass in the forward direction.

For these 6 parameters, several one way ANOVAs were used to test the significant difference between prosthetic foot types with respect to the sound or the affected limbs. A confidence level of $\alpha=0.05$ was chosen. Additionally, several asymmetry or symmetry indices were calculated from the force and

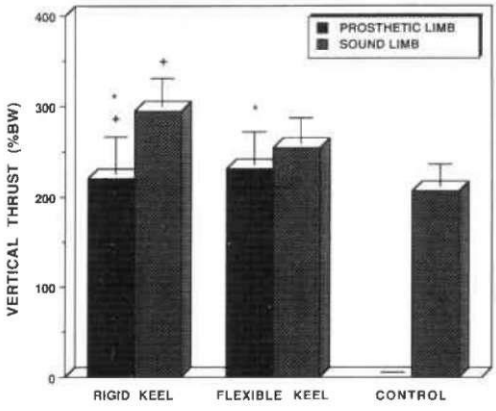

Fig. 2. Maximum vertical thrust for the rigid keel, flexible keel and control groups. Significant differences ($p<0.05$) with the amputee's sound limb are denoted with an asterix (*) while significant differences with the control group are reported with the symbol (+). This applies also to the following figures.

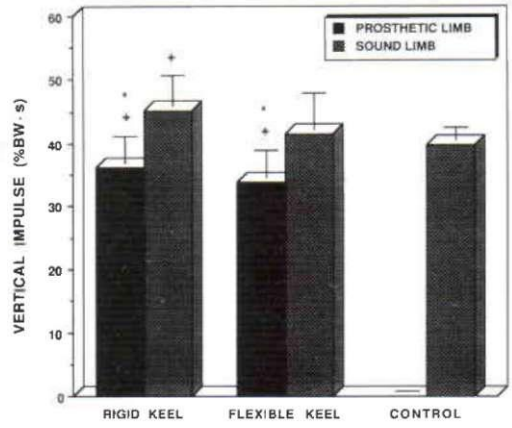

Fig. 3. Vertical impulse for the rigid keel, flexible keel and control groups.

impulse values. These ratios were taken from Seliktar and Mizrahi (1986) and Robinson *et al.* (1987). Among these, the forces and impulses of the prosthetic limb over the corresponding sound limb values were the most consistent ratios.

Results and discussion

The force values have been normalized with respect to body weight and are expressed as percentages. The impulse values have been normalized only with respect to body weight, leaving the time component untouched. The normalized impulse is then expressed in percent body weight-second (% BW· s). Lee *et al.* (1989) normalized both the force and time units of the impulse. This may be justified in normal walking or running if speed is not controlled. Knowing that the amputee spends more time on the sound limb than on the prosthetic limb (Seliktar and Mazrahi, 1986), it is felt that an adjustment on the stance by normalizing it would be to ignore the important time factor in the impulse calculation. The amputee can compensate gait asymmetry by modulating both the force and time parameters rather than one or the other.

The results are reported for the SACH foot or rigid keel, the flexible keel foot prostheses and for values obtained from the right limb of the control group which are included for reference. The maximum forces are presented first, followed by the corresponding impulse values and their respective ratios.

In Figures 2 and 3, the maximum vertical thrust and impulse are given respectively for the two types of prosthetic foot wearers as well as for the control limb. The sound limb of amputees fitted with the rigid keel always shows higher values than that of the control limb, illustrating its preponderant role during the support phase. The maximum thrust asymmetry is more marked for the rigid keel than for the flexible keel foot prostheses. Furthermore, there is no significant difference between the values of the flexible keel and those obtained from the control group.

The vertical impulse values (Fig. 3) show a similar trend. Significant differences are reported between the sound and the prosthetic limbs, regardless of the prosthesis being used. The affected side values are always smaller 36.2%BW· s for the rigid keel and 33.9 %BW· s for the flexible keel prostheses compared to the respective sound limb values which are about 43 %BW· s. The rigid keel prosthesis is different in that the force asymmetry is larger than the impulse asymmetry. This is mainly due to a significantly higher force and impulse being developed by the contralateral sound limb.

The braking force (Fig. 4) and the braking impulse (Fig. 5) present a similar pattern revealing significant interlimb asymmetry in amputee running gait. The SACH foot exhibits the highest force (33.2 %BW) and impulse (1.9 %BW· s) among the prosthetic feet in reducing the amputee's forward momentum. This is

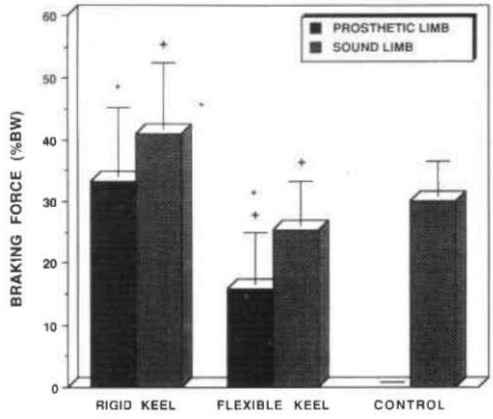

Fig. 4. Braking maximum force for the rigid keel, flexible keel and control groups.

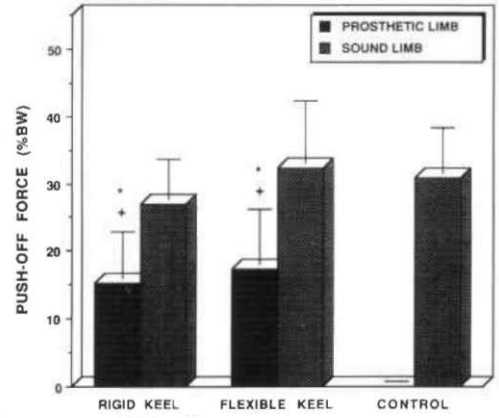

Fig. 6. Push-off maximum for the rigid keel, flexible keel and control groups.

related to the typical BK initial ground contact condition with the knee in total extension to prevent buckling (Enoka *et al.*, 1982).

The braking forces developed by the sound limb of amputees fitted with the flexible keel prostheses (25.3 %BW) are also different from the braking forces reported for the non-amputees (30.0 %BW). The corresponding impulse values on the other hand reveal no significant differences. The impulses reflect a good momentum conservation which could not be predicted from the braking forces alone. It is thought that better braking force and impulse conditions could have been obtained with the flexible keel prostheses if the FSF prosthesis had been fitted with a cushioned heel rather than just a rubber sole glued to its base. Without the FSF, the average braking force and impulse of the flexible keel group would have been much larger (20.2 %BW and 1.59 %BW· s) thus reducing the asymmetry between the affected and sound limbs.

With respect to the push-off force (Fig. 6), the flexible keel prostheses develop about the same propulsion force as the rigid keel; although the values are about 53% of the control limb. The sound limb push-off forces are essentially similar to those of the controls. The push-off impulses (Fig. 7) are significantly greater for the flexible keel foot prostheses (1.32 %BW· s) than for the SACH foot (0.86 %BW· s). The low impulse values for the flexible keel prostheses are attributed to the poor performance of the SAFE foot (0.90 %BW· s) which is comparable to that of the SACH foot. This would support Michael's

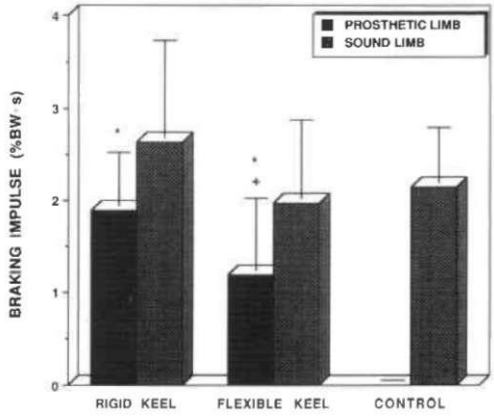

Fig. 5. Braking impulse for the rigid keel, flexible keel and control groups.

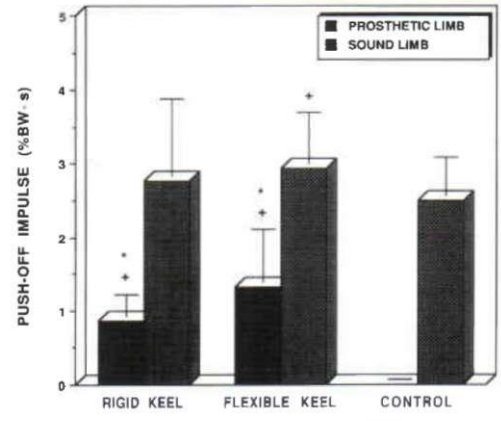

Fig. 7. Push-off impulse for the rigid keel, flexible keel and control groups.

(1987) rationale for not considering the SAFE prosthesis as a so-called energy storing foot. For both prosthetic types, average impulse values for the amputee's sound limb are higher than for the control group, reflecting a form of compensation mechanism for the inefficiency of the prosthetic device. For the rigid keel prosthesis group, this difference is not significant due to its large standard deviation. Results for the flexible keel group are different from the control group due to their lesser variability than that of the rigid keel group.

Different ratios were calculated using vertical, braking and push-off forces and impulses. The symmetry index defined by Robinson *et al.* (1987) was applied successfully by Herzog *et al.* (1989) in 34 ground reaction force variables to assess asymmetry in normal human gait. Using the same index, an abnormal characteristic trend associated with amputee running gait was observed but no significant differences were found between the two types of prosthetic feet. Only the ratio of the prosthetic forces and impulses over the corresponding sound limb values (Seliktar and Mizrahi, 1986) respectively yielded consistent results (Table 1).

With respect to the interlimb running gait asymmetry, all the impulse ratios were usually lower than the force ratios giving greater empahsis to asymmetry. The asymmetry in these feet may be explained in part by the time response characteristics of the keel, affecting both the force modulation as well as the force amplitude. This is mostly manifested during the critical push-off period where the amputee must assure his forward displacement and prepare himself for the following step while maintaining a steady state velocity. Difference between braking forces and impulses are less pronounced. This can be attributed to the relatively passive function of the braking phase when the prosthetic limb strikes the ground. The lack of muscle action and the use of cushion heel on both types of prostheses result in a similar force and impulse relationship pattern. The vertical force and impulse ratios show an opposite trend for the rigid keel prostheses. This difference can be attributed to the higher than normal values for the sound limb of amputees fitted with the SACH foot. Zahedi *et al.* (1987) reported that kinetic measurements are variable in assessing amputee locomotion and prosthetic alignment. The authors' results using ground reaction forces only, confirm their findings. However, impulse values which were not discussed by Zahedi *et al.* (1987) displayed a greater interlimb asymmetry than the forces.

Table. 1. Force and impulse ratios obtained from rigid keel (RK) and the flexible keel (FK) groups expressed in percentage.

	Vertical		Braking		Push-off	
	RK	FK	RK	FK	RK	FK
Forces (% BW)	75	91	81	62 (80)*	57	78
Impulses (% BW· s)	80	81	72	60 (80)*	42	45

(*) Values without the FSF prosthesis.

The effect of prosthetic type on running gait asymmetry is well discriminated by the force and impulse ratios. The flexible keel prostheses display less asymmetry than the rigid keel group relfecting their dynamic elastic characteristics. The vertical and push-off force ratios are closer to normal when the flexible keel prostheses are used. A similar trend is also noticed with the impulse ratios but, the differences are less apparent. It can still be assumed that the flexible keels are better than the rigid ones in respect of asymmetry, but further improvements are warranted to reduce the running gait asymmetry.

Conclusions

In this work interlimb asymmetry in BK amputee running gait was assessed for rigid and flexible keel foot prostheses using vertical and AP ground reaction forces and corresponding impulses. The impulse indicator showed greater asymmetry between the prosthetic side and the sound limb than the force parameters; whereas the impulse parameter was more consistent.

Interlimb asymmetry, evidenced by force and impulse ratios, is very much present in both types of prosthesis. The force ratios better differentiate between prosthetic foot types than the impulse ratios. The asymmetry in flexible keel prostheses may be more associated with the force profile modulation than its magnitude. Notwithstanding the present limitations of flexible keel prostheses, the resulting asymmetry is relatively lower than with the SACH foot. It appears that the elastic characteristics of the flexible keel prostheses

are mainly responsible for the decreased asymmetry in running gait. In general, the impulses better describe interlimb asymmetry and the forces allow a greater discrimination between prosthetic type.

Acknowledgement

The authors express their gratitude to the National Health Research and Development Program (NHRDP) of Canada for supporting part of this work. The Formation de Chercheurs et d'Aide à la Recherche (FCAR) fund is also acknowledged for the Ph.D. scholarship awarded to Mr. Fançois Prince. We wish to extend our gratitude to Mr. Jacques Ozie for his technical assistance.

REFERENCES

ALLARD P, KOFMAN J, LABELLE H, LÉVESQUE C, DUHAIME M (1988). Vorlaufige bewertung eines optimalen blattfeder-kunstfubes fur kinder (preliminary evaluation of an optimally designed flat-spring foot prosthesis for children). *Orthop Tech* **39**, 261–268.

BROUWER BJ, ALLARD P, LABELLE H (1989). Running patterns of juveniles wearing SACH and single-axis foot components. *Arch Phys Med Rehabil* **70**, 128–134.

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

BURKE MJ, ROMAN V, WRIGHT V (1978). Bone and joint changes in lower limb amputees. *Ann Rheum Dis* **37**, 252–254.

CULHAM EG, PEAT M, NEWELL E (1986). Below-knee amputation: a comparison of the effect of the SACH foot and single axis foot on the electromyographic patterns during locomotion. *Prosthet Orthot Int* **10**, 15–22.

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

ENOKA RM, MILLER DI, BURGESS EM (1982). Below-knee amputee running gait. *Am J Phys Med* **61**, 66–84.

HERZOG W, NIGG BM, READ LJ, OLSON E (1989). Asymmetries in group reaction force patterns in normal human gait. *Med Sci Sports Exerc* **21**, 110–114.

HURLEY GRB, MCKENNEY R, ROBINSON M, ZADRAVEC M, PIERRYNOWSKI MR (1990). The role of the contralateral limb in below-knee amputee gait. *Prosthet Orthot Int* **14**, 33–42.

LEE AG, ENGSBERG JR, HARDECK JA (1989). Force platform analysis of gait of children with a below the knee amputation. Horizontal components. In: Proceedings of the ASME Winter Meeting, p111–112.

MICHAEL JW (1987). Energy storing feet: a clinical comparison. *Clin Prosthet Orthot* **11**, 154–168.

MILLER DI, PASSER MW, BURGESS EM (1984). Development and improvement of running skill in unilateral below-knee amputees. (VA Contract no. V663P-1254). — Seattle, WA: University of Washington, p243.

MILLER DI (1987). Resultant lower extremity joint moments in below-knee amputee during running stance. *J Biomech* **20**, 529–541.

PERRY J (1975). Pathological gait. In: Atlas of Orthotics: Biomechanical Principles and Applications. — American Academy of Orthopaedic Surgeons: St-Louis: CV Mosby Co. p144–168.

ROBINSON RO, HERZOG W, NIGG BM (1987). Use of force platform variables to quantify the effects of chiropractic manipulation on gait symmetry. *J Manipulative Physiol Ther* **10**, 172–176.

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

SKINNER HB EFFENEY DJ (1985). Gait analysis in amputees. *Am J Phys Med* **64**, 82–89.

WING DC, HITTENBERGER DA (1989). Energy-storing prosthetic feet. *Arch Phys Med Rehabil* **70**, 330–335.

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

ZAHEDI MS, SPENCE WD, SOLOMONIDIS SE, PAUL JP (1987). Repeatability of kinematic and kinetic measurements in gait studies of the lower limb amputee. *Prosthet Orthot Int* **11**, 55–64.

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The social and economic outcome after upper limb amputation

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Abstract

This paper reports a study of 66 upper limb amputees in County Funen, Denmark who were visited in their homes by the author. The purpose of the study was to evaluate for the same period of time the social and economic outcome for a population of upper limb amputees compared with the normal population.

The number of amputees investigated corresponds to the annual number of persons becoming upper limb amputees in Denmark. The aetiology of registered amputees also corresponds to the Danish Amputation Register.

The amputees had become "better placed" in the social system after amputation independent of prosthetic use. The mean age of the amputees corresponded with the age where people reach their best social grouping. Their social migration quotient was higher than the background population and reflected the amputees better income and housing conditions. The reasons for these surprising results must be the high grade social system in a sophisticated industrial country. None of the amputees were required to pay for rehabilitation or prosthetic supply.

A lower divorce rate for the amputees was explained by a symbiosis between the amputees and their partners to protect their future existence. Only 14% lived alone.

Those who had their sexual debut after amputation were 3 years later in sexual experience than the rest of the amputees because of difficulties during the maturing process.

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Introduction

The loss of an upper limb results in a major and sudden restriction of function, sensation and of cosmesis and can be a great socio-economic catastrophe for the individual.

The ratio of upper and lower limb amputees differs between countries. War has a great influence (Steinbach, 1986).

Denmark is a well developed country with a high grade social system. In Denmark upper limb amputations constitute 3% of all amputations. The annual number of upper limb amputations is 75 (62—82) compared to 2164 lower limb amputations (Ebskov, 1986; Andersen-Ranberg and Ebskov, 1988).

The primary purpose of this study was to evaluate the social and economic outcome for a population of upper limb amputees compared with the normal population in the same period of time.

Method

The number of upper limb (UL) amputees in County Funen, Denmark was not known. The Amputation Register (Ebskov, 1986) could only count back to the year 1972. Therefore the material was collected from all the hospitals and prosthetic centres in the county. All registers were examined. The material therefore includes all amputees who are being or have been treated in the County of Funen in the period 1 January 1900 to 31 December 1987.

All amputees were visited in their homes by the author. This gave the opportunity to see and register their daily surroundings and function and meet their families.

Two sets of questionnaires were constructed. One for amputees under 18 years and one for amputees over 18 years.

The amputees were characterised as active

Table 1. Total number of upper limb amputees registered.

Cause of amputation	F	M	n	%
Trauma	4	49	53	50
Congenital	9	5	14	13
Brachialplexus lesion	1	5	6	6
Vascular disease	13	8	21	20
Tumour	4	7	11	11
Summation	31	74	105	100

Table 2. Number of upper limb amputees visited.

Cause of amputation	F	M	n	%
Trauma	3	40	43	65
Congenital	6	4	10	15
Brachialplexus lesion	1	5	6	9
Vascular disease	3	1	4	6
Tumour	1	2	3	5
Summation	14	52	66	100

users (i.e. active prosthesis more than 8 hours a day), partially active users (i.e. active prosthesis less than 8 hours a day), passive users (i.e. only users of a passive prosthesis regardless of time) and prosthetic non-users (i.e. no use of prosthesis at all).

The amputees were classified in social groups after the terms of the Danish Institute for Social Research (Enevoldsen *et al.*, 1980) and compared with the social groups of the background population. Children were placed in the social groups of their parents, and pensioners as a function of their education and work at retiral.

The mean time lapse from amputation to completion of the investigation was 20.6 years. The year 1968 corresponds to the mean year of amputation. Therefore the commencement year 1968 and the year of completion 1988 were selected as the basic investigation years.

As the background population for the basic investigation years two comparable groups were selected. The two groups were respectively the Danish population and the age group in the population corresponding to the mean age of the amputees at amputation and at review.

Results

Some 105 UL amputees were registered; 32

Table 3. The personal characteristics of the amputees and their school education and social education and social grouping at review.

Number	26	6	16	18
Characteristic	Active	Part. active	Passive	No prosthesis
Mean age in years at review	29.2	35.3	70.0	44.1
Mean time in years since amputation	13.8	8.3	46.5	12.3
Loss of dominant hand (%)*	68.4	83.3	43.8	40.0
Loss of elbow (%)**	15.4	0.0	31.3	38.9
School education mean years at review	9.8	8.8	6.5	7.9
Social grouping at review (%)				
I	0.0	0.0	0.0	0.0
II	8.0	17.0	13.0	11.0
III	50.0	17.0	31.0	39.0
IV	19.0	17.0	31.0	17.0
V	13.0	50.0	25.0	33.0

* Congenital amputees not included.
** All congenital amputees had a 8–12 cm stump below elbow.
Social group classification:
I Very large self-employed and top salaried employees
II Major self-employed and major salaried employees
III Minor self-employed and middle salaried employees
IV Wage earners, skilled and minor salaried employees
V Wage earners, unskilled

were dead and 7 would not participate (3 were in conflict with the hospital system, one had psychiatric reasons and 3 did not give any reasons) (Table 1).

Sixty-six amputees were visited in their homes by the author (Table 2). The mean age at amputation was 24.5 years (0—72 years). Mean age at review was 45.1 years (4—83 years). The mean time lapse from amputation to review was 20.6 years (0—63 years). Twenty-six were active prosthetic users, 6 were partially active prosthetic users, 16 were passive users and 18 did not use a prosthesis at all. Table 3 shows the personal characteristics of the amputees and their social grouping. Loss of dominant hand did not include those with deficiencies present at birth who are for convenience described as congenital amputees. There were no significant differences in the social grouping between the 4 functional prosthetic groups.

Amputation was traumatic in 65% of the cases. As the persons with plexus lesions are seen as traumatic amputations (Kejlaa *et al.*, in press), 74% of the amputations were the consequence of trauma. Most accidents occurred in the industry involving unskilled or skilled workers and in the traffic affecting youngsters without education. This explains why 70% of the amputees were placed in the two lower social groups when amputated (Fig. 1).

At review 20 years later the amputees had placed themselves in a better position in the social group system (Fig. 2). Compared to the background population the amputees had a higher social migration quotient for the same period of time. None of the amputees was placed in the social group 1 either at amputation or at review.

The employment status at amputation and at review for the amputees is shown in Figure 3. With time more amputees had become pensioners. Of the pensioners 65% were entitled to pension because of age, the rest were pensioners for other reasons not related to amputation (early retirement pension). There were no pensioners in the congenital and plexus groups. In the tumour and illness groups 86% (n=6) were pensioners. The background population included 18% pensioners (of which 73% were retired because of age, 27% had early retirement pension) at time of review.

The unemployment rate for the amputees was twice the unemployment rate in Denmark both in regard to the total population and the

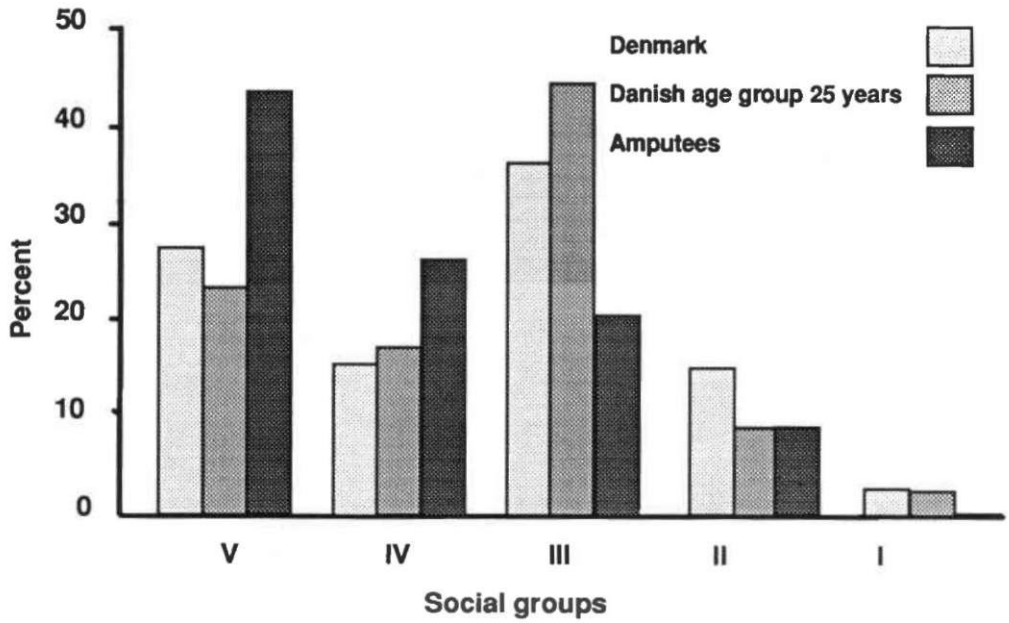

Fig. 1. The social grouping of amputees compared to the Danish population and the corresponding age group fitting the mean age for the amputees at amputation in percent (1968) (2.1% of the Danish population and 2.8% of the Danish corresponding age group could not be classified).

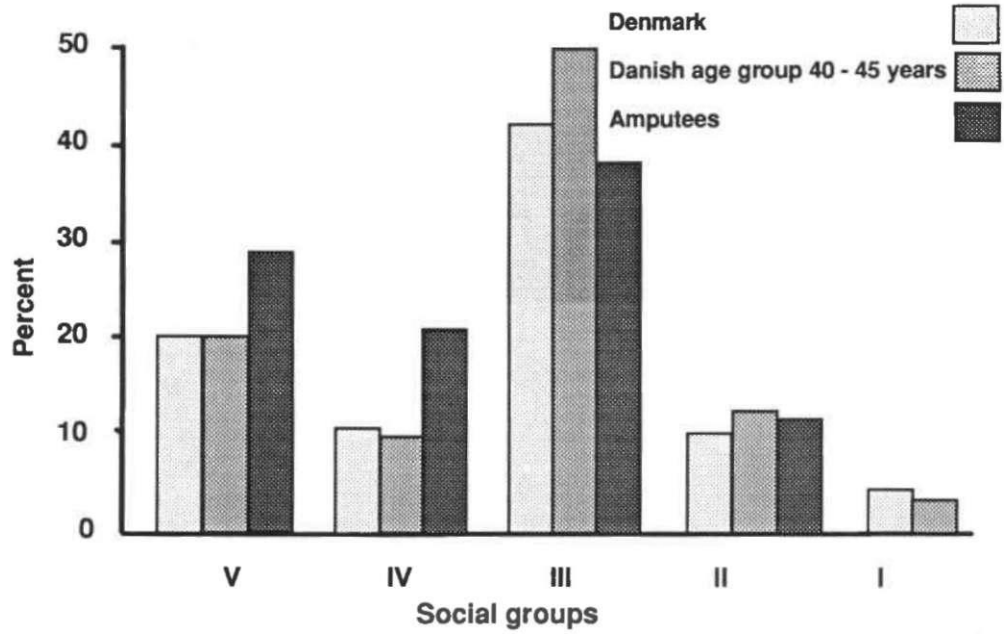

Fig. 2. The social grouping of amputees compared to the Danish population and the corresponding age group fitting the mean age for the amputees at review in percent (1988) (12% of the Danish population and 3.5% of the Danish corresponding age group could not be classified).

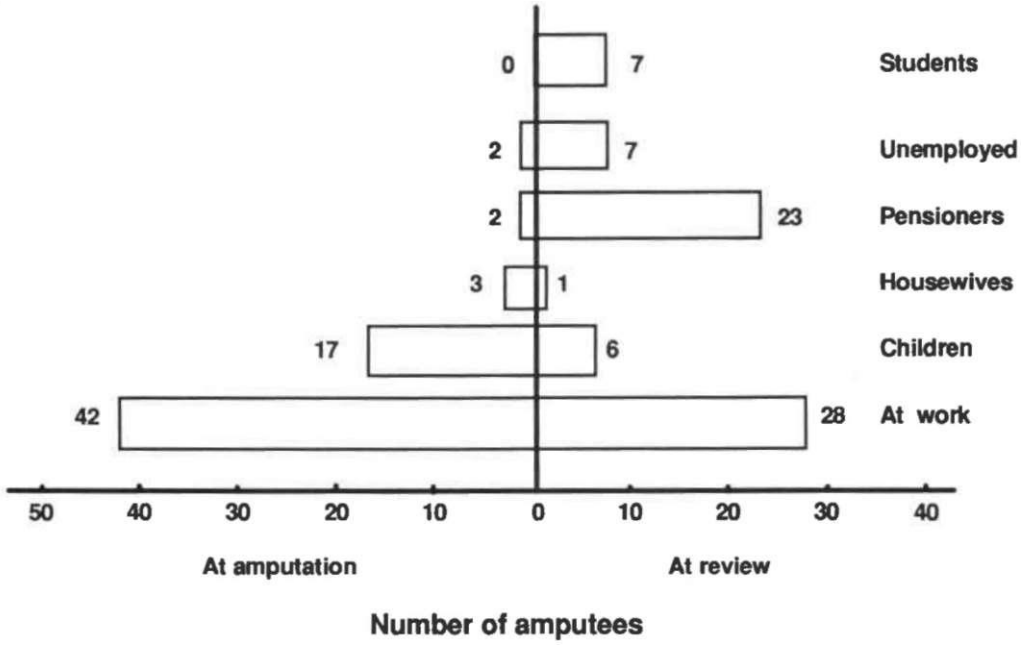

Fig. 3. The employment status of the amputees at amputation and at review.

total labour force. There has been an increase in the average unemployment status in Denmark from 0.7% to 8.7% from 1965 to 1988. This corresponds to the unemployment increase for the amputees.

The mean gross income for persons in Denmark was 128,800 Danish Kroner (DKK) (lower quartile 57,200, upper quartile 222,100 DKK) in 1986. The mean gross income for the amputees at review was 137,900 DKK. (39,000–480,000 DKK).

Seventy-six percent (n=37) of those who had a traumatic cause of amputation were entitled to an insurance payment. The amount became index regulated in 1964. The mean amount was 220,000 DKK (36,000–540,000 DKK). The insurance amount was in most instances paid out in instalments until the amputee became a pensioner and then the balance was paid.

Dwellings, households and persons by type of building and number of rooms are shown in Figures 4 and 5. A greater number of amputees than the background population lived in one-family houses and their dwellings had more rooms. The mean floor space for the amputees was 115 square metres (12–460 square metres) against 102 square metres (39–>300 square metres) for the background population. Of the amputees 77% owned their houses against 56% of the background population.

Most families in Denmark have 2–3 children. Only 2% have more than 5 children. There were fewer amputee families with 2–3 children, but more with 5–6 children. The number of occupants of dwellings were the same for the background population and the amputees.

The divorce rate in Denmark is today about 50%. The mean divorce rate in Denmark concerning the period from the earliest marriage for the amputees to completion of the investigation was 23% (7–46%). The divorce rate for the amputees was 11% (4 out of 37 marriages).

Only 14% of the amputees lived alone against 28% in the background population. Six were children. The first sexual experience was in mean 3 years later (1.8–6.4 years) for 12 amputees who had their sexual debut after amputation compared with 48 amputees who had their first sexual experience before amputation.

Discussion

The number of investigated amputees corresponds to the annual number of persons

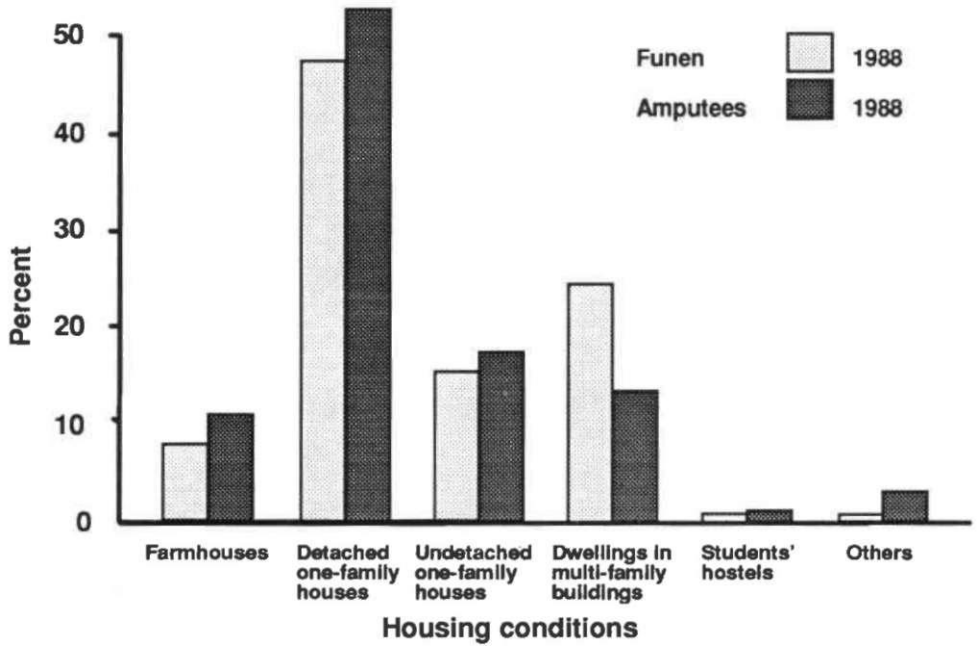

Fig. 4. Dwelling stock by type of building in percent. Amputees compared with the condition for the County Funen.

who become UL amputees in Denmark. The aetiology for the registered amputees corresponds well with the distribution of the Amputation Register (Ebskov, 1986; Andersen-Ranberg and Ebskov, 1988).

It is of interest that the amputees independent of prosthetic use had become better placed in the social system than before amputation. In this regard it is important to remember that the majority will not reach their best social group until the age of forty. This corresponds well with the mean age of 45.1 years in this investigation.

The social laws have changed with the development of society and have become more profitable for the individual. The good social system in Denmark has been of great importance for the amputees. All have had the opportunity to become rehabilitated. For the amputees there has been no cost in regard to rehabilitation or prosthetic supply. Another important factor is the education level in Denmark. Most amputees had an education of a minimum of 7 compulsory school years. It has been shown (Andersson and Berg, 1975) that low age and good education were positive factors for successful rehabilitation. This corresponds well with this investigation.

The insurance amount has been of supporting value for those who became amputees in an accident or at work. This gave the amputees a guarantee for rearrangement of their lives and therefore a good outcome of rehabilitation.

The higher social migration quotient for the amputees must be seen as a consequence of the above mentioned facts.

The higher income for the amputees must be explained by their better social grouping and their mean age at time of investigation, which again explains their better housing conditions. The high number of pensioners is a consequence of age and the higher number of persons with illnesses and tumours in relation to the background population.

The unemployment increase for the amputees was a reflection of the current unemployment increase in Denmark.

The low divorce rate must be understood as a symbiosis between the amputees and their partners producing a dependence secured by a stable relationship to protect future existence. This corresponds with the fact that only 14% of the amputees lived alone.

Sexual debut was 3 years delayed as a consequence of amputation and this delay must be explained by a difficult age with insecurity and fear, which disappear with the ripening process not only of the amputee, but of the counterparts.

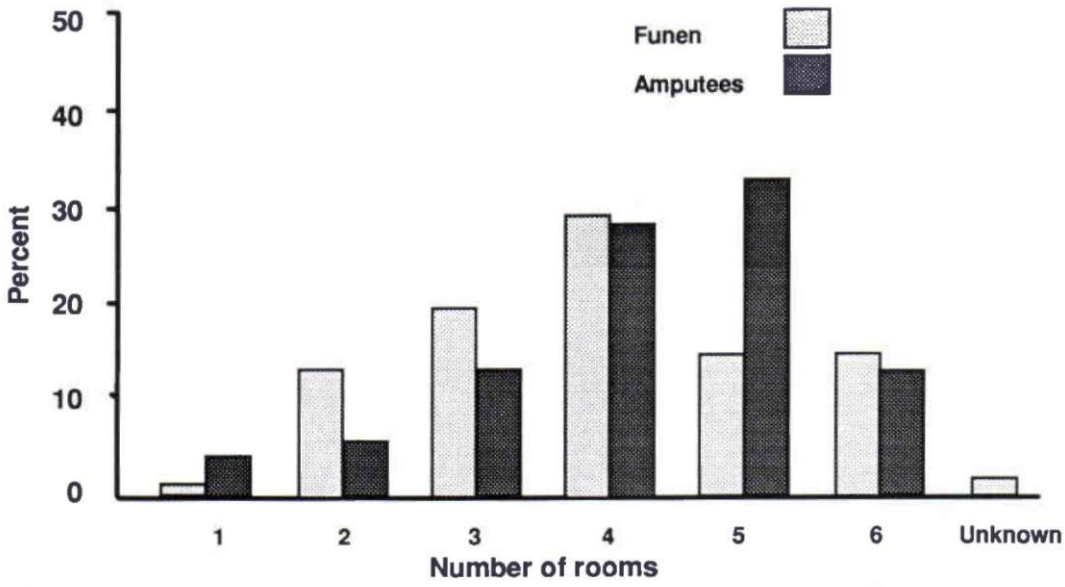

Fig. 5. Dwelling stock by number of rooms in percent. Amputees compared with the condition for the County Funen.

REFERENCES

ANDERSEN-RANBERG F, EBSKOV B (1988). Major upper extremity amputation in Denmark. *Acta Orthop Scand* **59,** 321–322.

ANDERSSON K, BERG S (1975). The relationship between some psychological factors and the outcome of medical rehabilitation. *Scand J Rehabil Med* **7,** 166–170.

DANMARKS STATISTIK: STATISTICAL YEARBOOK (1971). Copenhagen.

DANMARKS STATISTIK: STATISTICAL YEARBOOK (1989). Copenhagen.

EBSKOV B (1986). The Danish amputation register 1972–1984. *Prosthet Orthot Int* **10,** 40–42.

ENEVOLDSEN B, MICHELSEN N, FRIIS-HASCHE E, KAMPER-JORGENSEN F (1980). Social classifications. II. Svalastoga's subdivision according to social status rank and the social grouping employed by the Institute for Social Research. *Ugeskr Læger* **142,** 544–550.

KEJLAA G, BARFRED T, SIMFSEN K (in press). Amputation and shoulder arthrodesis after brachial plexus lesions. *Ugeskr Læger.*

STEINBACH T-V (1978). Upper limb amputation. *Prog Surg* **16,** 224–248.

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Functional benefit of an adaptive myoelectric prosthetic hand compared to a conventional myoelectric hand

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Abstract
Eight patients with a traumatic unilateral upper limb amputation, who used conventional myoelectric prostheses, were also fitted with a commercially available myoelectric prosthetic hand with an adaptive grip, in order to compare the functional benefit of the two types of prostheses.

Comparisons were made regarding width of grip, force of grip, scores in a standardised grip function test and prosthesis preference. The conventional prosthesis showed significantly better results regarding these parameters. The adaptive hand does not appear to be fully developed for practical use in prosthetic rehabilitation.

Introduction

Myoelectric prostheses were developed about 1960 and have become useful in prosthetic rehabilitation (Schmidl, 1973). At the authors' centre, the conventional type of myoelectric prosthetic hand is used in below-elbow amputation. It is also used in above-elbow amputation if the length of the stump is sufficient for the use of a body-powered elbow. In these patients the functional results are often good enough to provide a reason for regular use of the prosthesis. In a few patients with a high above-elbow amputation some functional improvement has also been observed from a conventional myoelectric hand, in systems in which the more proximal functions of the prostheses seemed to be the limiting factors (Thyberg and Johansen, 1985; Johansen *et al.*, 1986).

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Until now the prosthetic hand has a non-adaptive grip and one trend in prosthetic research has been to construct a prosthetic hand with a grip more like the human hand (Kato, 1978). Different prototypes have been developed and for some years the adaptive myoelectric ES hand has been commercially available (Boenick and Becker, 1980; Roesler, 1982). The authors' aim was to study the usefulness of this prosthesis, compared to a conventional non-adaptive myoelectric prosthesis, in rehabilitation of patients with a traumatic unilateral upper limb amputation.

Material

Patients

Eight consecutive patients attending the prosthetic clinic, who reported regular use of their conventional myoelectric prostheses, were offered a trial of the new prosthesis with an adaptive grip. All patients accepted. Patient data are given in Table 1. All were men with a unilateral traumatic upper limb amputation and no additional impairment.

Six patients were fully employed, 3 in practical work, 3 in mainly desk work, and 2 were studying.

All patients were trained to use myoelectric prostheses during their initial rehabilitation programmes, and they all reported a daily use which tallied with the observed need of frequent technical service.

Prostheses

Each patient was fitted with one myoelectric adaptive hand (ES Hand, Protesindustri AB) and one myoelectric prosthetic hand of conventional type (Otto Bock 8E38=7 3/4) at the same time and with identical sockets (Fig. 1). The prostheses were adjusted at delivery

Table 1. Patient Data

Patient No	1	2	3	4	5	6	7	8	Median
Age	41	38	27	63	58	27	31	58	39.5
Time since amputation (years)	4	2	3	46	39	2	29	20	12
Time since 1st myoelectric prosthesis (years)	2	2	2	5	17	2	9	20	3.5
Amputation level	AE	BE	BE	BE	BE	BE	BE	BE	
Amputation side	R	L	L	R	R	R	L	L	
Dominance before amputation	R	R	R	R	R	R	?	R	

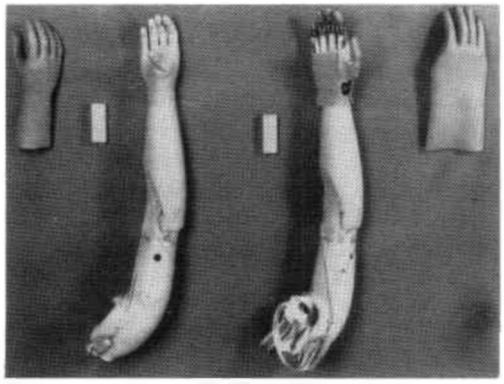

Fig. 1. Adaptive hand (left). Non-adaptive hand (right).

and the patients' ability to control the prostheses was checked.

The adaptive hand (Fig. 2) is produced in one size for adults. It has a motorised adaptive grip (Boenick and Becker, 1980) in which flexion in the second and third digit respectively is continued until further flexion in each finger is stopped by the object. This is achieved by wires arranged like tendons. The fourth and fifth digits are flexed by the same type of mechanism but flexion is stopped automatically when flexion in the second and third digit is stopped. Thus, the grip is not adaptive with regard to all fingers independently. Flexion of the thumb is also motorised and is activated simultaneously with flexion of the fingers. In addition to the usual prosthetic tip pinch or power grip the position of the thumb can be altered to get a pinch grip against the lateral aspect of the second digit. The hand was delivered with a short technical instruction, cables, glove and batteries (Otto Bock type 757B8). Since no wrist unit was available from the manufacturer, an Otto Bock (10SI=50, 10S4, 10S7) wrist unit was added to the system.

Method

All patients were instructed to use the adaptive hand as much as possible, without regard to preference during the first two months after delivery. During the following ten months both types of prostheses could be used and the patients were tested concerning grip function and technical parameters. At follow-up after one year the patients were asked which type of prosthesis they preferred for further

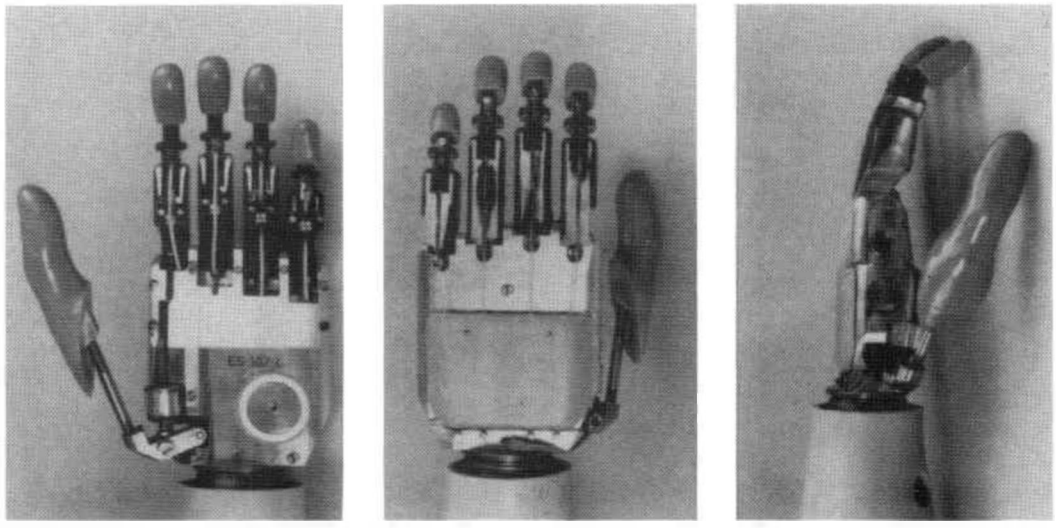

Fig. 2. Adaptive hand.

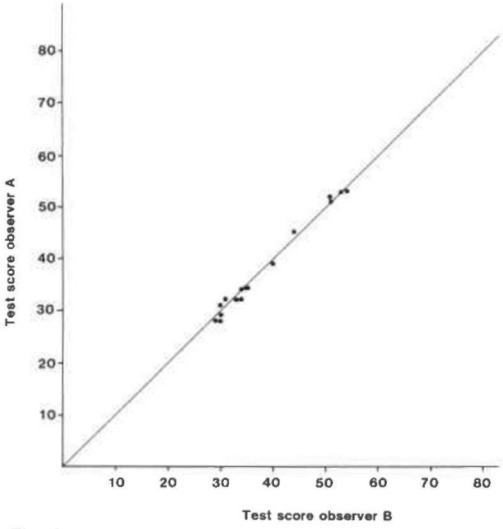

Fig. 3. Interobserver correlation in grip function test (r_s=0.97, p<0.001)

use. In addition their opinion was sought regarding the cosmetic appearance of the prostheses.

The patients were tested with a standardised grip function test (Sollerman, 1980; Wilton, 1990). The test is based on previous studies by Sollerman and Sperling, where the grip pattern of the healthy hand is divided into seven main hand grips related to normal function of the human hand. The test consists of 20 different ADL tasks scoring from 0 to 4 points.

Each patient was assessed at three sessions with each type of prosthesis in an alternating order. The highest score obtained with the adaptive hand was compared to the highest score obtained with the non-adaptive conventional hand.

To assess the reliability of the test, 7 out of 8 patients participating in the main study and 9 additional patients, with unilateral upper limb amputations and fitted with myoelectric non-adaptive prostheses, were tested and scored by two independent observers.

Width of grip, force of grip, weight of hand and maximum circumference of the hand were tested according to previously published test instructions (Ingvarsson *et al.*, 1982).

Width of grip was measured by grasping prisms and cylinders with size intervals of 5 mm and with the hand placed horizontally. Maximum force of grip at 20%, 50% and 80% of maximum gripping width was measured by means of a strain gauged device (AB Detektor, Gothenburg). The mean values of five consecutive tests were compared.

Correlations were described by Spearmans rank correlation coefficient (r_s) and differences were tested with Wilcoxons signed rank test.

Results

The grip function test

When scores by two independent observers

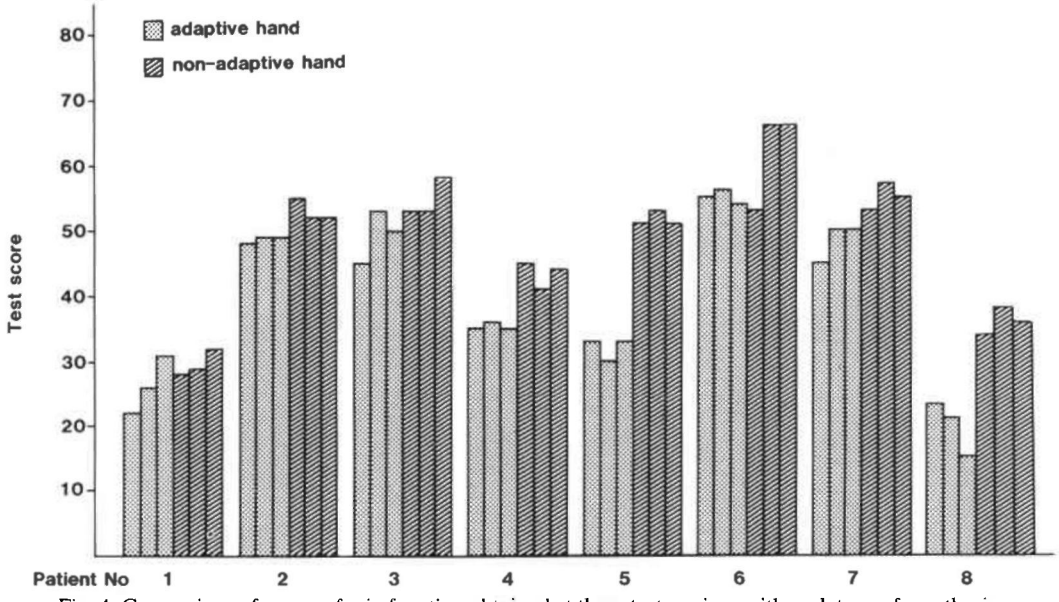

Fig. 4. Comparison of scores of grip function obtained at three test sessions with each type of prosthesis.

Table 2. Mean and range of maximum force of grip at 20%, 50% and 80% of maximum width of grip.

	Width of grip (% of max)	20	50	80
Maximum force of grip (N)	Adaptive hand	24 (15–43)	28 (17–44)	30 (17–45)
	Non-adaptive hand	105 (90–138)	116 (97–151)	129 (102–176)

of the grip function test were compared a good correlation was found (r_s=0.97 and $p<0.001$) (Fig. 3).

Adaptive versus non-adaptive hand

The scores for each type of prosthesis in the grip function test are presented in Figure 4.

In each type of prosthesis the lowest score was usually gained at the first test. The second test tended to give a higher score and the third test tended to give the same score as the second or lower.

The best scores of the non-adaptive conventional hand were significantly better than the best scores of the adaptive hand, ($p<0.01$).

Technical test

Width of grip and force of grip were significantly greater for the non-adaptive conventional hand (Tables 2 and 3).

The maximum circumference (closed hand) was 270 mm for the adaptive hand and 260 mm for the non-adaptive conventional hand.

The adaptive hand weighed 595g and the non-adaptive conventional hand weighed 505g.

Prosthetic preference

Both types of prosthesis were available for practical use and after one year the patients were asked which hand they preferred for further use. All patients preferred the non-adaptive conventional hand. The cosmetic appearance of the adaptive hand was considered not to be satisfactory by seven patients. One patient thought the cosmetic appearance of the adaptive hand was good.

Discussion

The consequences of upper limb amputation may be described in terms of impairment, i.e. level of amputation or loss of hand function. It may also be described in terms of disability, i.e. loss of ability to perform certain activities. A third alternative is to describe the consequences in terms of handicap, i.e. the disadvantage in relation to a specific environment or social role of a patient (WHO, 1980).

Table 3. Width of grip (mm) Mean and Range

Type of object	Cylinder	Prism
Adaptive hand	64 (55–75)	59 (40–70)
Non-adaptive hand	94 (80–100)	90 (70–100)

The aim of rehabilitation is to reduce the consequences of amputation and the effect of rehabilitation may be evaluated in relation to the above-mentioned aspects.

One important aim in prosthetic rehabilitation is to restore a degree of grip function, and if the improvement is relevant in the perspective of disability and handicap a prosthesis may be accepted by the patient. The acceptance of a prosthesis may be regarded as an indication of the benefit for the individual patient, but additional assessment of grip function and the ability to perform relevant, standardised activities may be helpful in the analysis of the more general benefit of the prosthesis. Most standardised indices of activities of daily living (Barer, 1989) relate to activities which are not relevant in unilateral amputation of the upper limb. To describe the disability of these patients, more sensitive and specific tests are required (Stein and Walley, 1983). The test of grip function described by Sollerman (Sollerman, 1980) is a standardised test which is representative for activities of daily living both with regard to the dominant hand and the non-dominant hand.

In patients with different impairments of hand function, Sollerman found a good correlation when the test results of two independent observers were compared (Sollerman, 1980), and this tallies with the results of the present study. Sollerman found a good correlation between the results of two consecutive testing procedures. In this study there was some intra-individual variation in three consecutive tests (Fig. 4) and in 11 of 16 cases the score in the second test was higher than in the first test. With regard to the observed intra-individual variation the best scores for each type of prosthesis, in each patient, were used for comparison.

The difference between the two types of prostheses, in the Sollerman test, was statistically significant. Whether this difference is clinically significant cannot be concluded from this study but it seems to be a relevant contributing factor regarding the preference of prosthesis.

The acceptance or rejection of a prosthesis (Childress, 1973) depends on the balance between the benefit and the trouble associated with the use of the prosthesis (Roeschlein and Domholt, 1989). The benefit may depend on the improvement of grip function or ability to perform manual activities and on cosmetic aspects, and in each case this benefit will depend on the environment and the social role of the patient. In general the grip function of a prosthesis is influenced by the technical properties of the prosthesis, the socket fabrication, and the training programme. The observed differences in width and force of grip were statistically significant. Although it is still debatable which technical parameters are clinically significant, the results in this project from the technical test also tallied with the patients' choice of prosthesis. In order to focus on grip function, identical sockets were used for the two types of prostheses. All patients were trained to use a conventional myoelectric prosthesis during their initial rehabilitation programme and their ability to control the adaptive prosthesis was checked at delivery. In order to minimize influence from different individual needs and social roles, each patient was used as his own control when the results were compared.

The usefulness of a prosthetic system, in rehabilitation, will also depend on whether technical service from the manufacturer is available or not. Regarding prototypes and small series, lack of service may limit the clinical usefulness of a prosthesis, despite good results concerning the discussed parameters.

Beside grip function, the subjective benefit of a prosthesis may depend on cosmetic factors. Regarding this aspect most of the patients also preferred the conventional prosthesis. A quantitative comparison of grip function and cosmetic aspects is difficult.

In conclusion, the particular type of adaptive hand that was studied did not appear to increase the functional benefit compared to a conventional myoelectric prosthesis. Thus, it could not be verified that an adaptive prosthetic hand would be the best technical solution. If a prosthetic system is to be clinically useful, it must provide good grip function and still be simple and reliable enough to use without the facilities of a development laboratory. In order to achieve this balance, a close contact between technical development and clinical rehabilitation may be one of the most important factors.

Directory of suppliers

ES Hand (Een and Holmgren Systemteknik Hand), Protesindustri AB, Box, 67 S-751 03 Uppsala, Sweden

Otto Bock Scandinavia AB, Box 623, S-601 14 Norrköping, Sweden

REFERENCES

Barer D, Nouri F (1989). Measurement of activities of daily living. *Clin Rehab* **3,** 179–187.

Boenick U, Becker FF (1980). Der derzeitige Entwicklungsstand adaptiver Kunsthände mit elektromechanischem Antrieb (Teil 1) (The present state of development of electrically driven adaptive hands (Part 1)). *Ortop Tech* **31,** 85–87.

Boenick U, Becker FF (1980). Der derzeitige Entwicklungsstand adaptiver Kunsthände mit elektromechanischem Antrieb (Teil 2) (The present state of development of electrically driven adaptive hands (Part 2)). *Ortop Tech* **31,** 97–100.

Childress DS (1973). Powered limb prostheses: their clinical significance. *IEEE Trans Biomed Eng* **20,** 200–207.

Ingvarsson B, Karlsson I, Ottoson LG, Thyberg M (1982). Test instructions for the technical testing of mono-functional myoelectrically-controlled prosthetic hands, proposal (technical note). *Prosthet Orthot Int* **6,** 41–42.

Johansen PB, Breitholtz M, Cavrini R, Johansson B, Schmidl H, Thyberg M (1986). Prosthetic rehabilitation in bilateral high above elbow amputation. *Scand J Rehabil Med* **19,** 85–87.

Kato I (1978). Trends in powered upper limb prostheses. *Prosthet Orthot Int* **2,** 64–68.

Roeschlein R A, Domholt E (1989). Factors related to successful upper extremity prosthetic use. *Prosthet Orthot Int* **13,** 14–18.

ROESLER H (1982). Entwicklung, Stand und Perspektiven kunstlicher Hände (Development, present state and perspectives of artificial limbs). *Ortop Tech* **33**, 101–105.

SCHMIDL H (1973). The INAIL-CECA prostheses. *Orthot Prosthet* **27**, 6–12.

SOLLERMAN C (1980). Grip function of the hand. Thesis. –Göteborg, Sweden: University Medical Dissertations.

STEIN RB, WALLEY M (1983). Functional comparison of upper extremity amputees using myoelectric and conventional prostheses. *Arch Phys Med Rehabil* **64**, 243–248.

THYBERG M, JOHANSEN PB (1985). Prosthetic rehabilitation in unilateral high above-elbow amputation and plexus lesion: case report. *Arch Phys Med Rehabil* **67**, 260–262.

WORLD HEALTH ORGANISATION (1980). International classification of impairments, disabilities and handicaps (ICIDH) a manual of classification relating to the consequences of disease. Geneva: WHO.

WILTON JC (1990). Hand function evaluation: a comparitive study between an American and a Swedish instrument. In: proceedings of 10th International Congress of the World Federation of Occupational Therapists. Melbourne 2nd – 6th April, 1990. p 365–368.

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Prosthetics and orthotics: a survey of centres in the Kingdom of Saudi Arabia

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Abstract

This paper reports the results of a survey carried out to evaluate existing prosthetic and orthotic facilities and programmes of education, training, and research and development in the Kingdom of Saudi Arabia. One hundred and twenty hospitals and medical rehabilitation centres were each circularised with a questionnaire requesting information that mainly concerned (i) types of prostheses/orthoses, (ii) area of facility, (iii) personnel number and qualifications, and (iv) problems encountered and suggested solutions. The completed questionnaires revealed that in the final analysis of data there were only ten prosthetic/orthotic facilities.

The survey provided useful data on the personnel, equipment, and facilities available in each hospital or medical rehabilitation centre, together with details of the services to prospective referring clinicians. Two centres were found to provide high quality services by qualified personnel. There were no formal prosthetic/orthotic training programmes and there was only one prosthetic/orthotic research and development centre.

The respondents generally felt that there were three major problems: (i) lack of qualified personnel, (ii) lack of materials and components, and (iii) lack of continuing education and training programmes. It is hoped that presentation of these results will provide facts for both health-care providers and educators which may be used as a basis for development in this important area of health-care.

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Introduction

In the Kingdom of Saudi Arabia, there is a sufficient number of disabled people, resulting from birth defects, non-communicable diseases and accidents to warrant implementation of rehabilitation programmes. Accordingly, the government has introduced many rehabilitation activities through the Ministries of Health, Social Welfare, Education and Defence.

The population of Saudi Arabia is estimated to be about 10 million, living over an area of approximately 900,000 square miles. One of the greatest benefits resulting from Saudi Arabia's new wealth is the enormously improved system of health-care which it has been possible to develop. Free medical care is provided by the state for Saudi citizens, non-Saudi residents (with very few restrictions), and pilgrims to the Holy Places of Islam. Specialised hospitals have been built and equipped with the world's most up-to-date medical instruments. King Faisal Specialist Hospital in Riyadh provides treatment for selected patients and King Khalid Eye Specialist Hospital in Riyadh is considered as one of the best equipped hospitals for eye care in the world. The Armed Forces Hospitals for the Armed Forces personnel and their dependents, the National Guard Hospitals, and the University Hospitals also provide specialty treatments for heart disorders and kidney failure, along with a host of other diseases. The Saudi Arabian Fourth Development Plan (1985–1990) aimed at raising the number of hospitals to 283, while hospital beds would be annually increased by 14% to reach 55,600 beds by the year 1990. This is considered vital if the government is to provide proper medical care for its people. The plan also aimed at raising the number of primary health-care centres to 15,000.

The country is divided into eleven health regions, each headed by a regional health director who is responsible for preventive and curative health services. The Ministry of Health provides almost 60% of the health services (67% of the hospital beds, 59% of physicians and 54% of the nurses). About 25% of the services are provided by more than 10 different governmental agencies and 15% of the services are provided by the private sector. They constitute a pooling of resources which demands solid planning and co-ordination. A study sponsored by the Ministry of Planning found that there is inadequate collaboration between health providers. Only to a certain extent does the Ministry of Health receive information from other health suppliers. The inefficient information system makes the planning and evaluation of health services a rather difficult task. There are no clear indications as to why, how, or by whom statistics of health services and their activities are to be collected, compiled, and analysed.

Despite the extensiveness of the medical statistics compiled by the Ministry of Health for its annual reports, no regular reports were available on the number and types of disabilities present, nor on the size and geographical distribution of the disabled population. There were no general statistics on the number and type of prostheses/orthoses prescribed in the Kingdom. In view of the lack of adequate information regarding the number, condition and geographical distribution of disabled persons, accurate correlative study between available resources and actual needs is not possible. All of this information is required for planning prosthetic/orthotic services and the evaluation of future needs in terms of personnel, facilities and funds.

At the time that this study was initiated, little information existed that defined the current status of the prosthetic/orthotic facilities and the needs of the disabled. Observations made during field visits by an ad hoc committee to various medical rehabilitation institutions have brought to light a variety of deficiencies in prosthetic/orthotic care that warrant further inquiry. In order both to correct these deficiencies and to sharpen the focus on the prosthetic/orthotic education and training, research and development and manpower development programmes, it was decided that it would be useful to determine, by means of a survey, exactly what programmes were actually available. The real need of improving these services or introducing new ones as perceived by the practitioners working in the field, could then be assessed.

This investigation describes the prosthetic/orthotic units in the hospitals and medical rehabilitation centres surveyed with respect to number of patients treated annually, types of orthoses and prostheses, area of facility, number of staff and their qualifications, materials and components, as well as the problems encountered and suggested solutions. It proceeds by enumerating and describing those facilities and the problems in prosthetic/orthotic care noted by principal workers themselves. The recommendations and conclusion of this report apply generally to the Ministry and non-Ministry of Health existing facilities.

Origins and objectives

This work was initiated out of field visits to various medical rehabilitation institutions by a committee consisting of an orthopaedic surgeon, a physiatrist, a physical therapist, a bioengineer and prosthetic/orthotic consultants. Their observations have brought to light a variety of deficiencies in rehabilitation services that warrant further inquiry. A comprehensive survey was, therefore, recommended to evaluate the existing facilities and patient needs and define the main problems of these services.

The principal aims of the survey were to review:

- the existing prosthetic and orthotic facilities;
- the programmes of education and training in prosthetics and orthotics;
- the prosthetic and orthotic research and development facilities; and
- the manpower development programmes in the field of prosthetics and orthotics.

Finally, the survey aimed to ascertain the personal opinions of the principal workers in the field as to the current status of the prosthetic/orthotic services, and in particular, their comments and remarks towards how the services, in their area, should develop in the future.

Methodology

In order to achieve these aims, an evaluation was carried out by means of questionnaires. The questionnaire consisted of eight major sections. The first section asks if there is already prosthetic/orthotic services available in the hospital or not. The second section, which is the clinical services section, aims at obtaining information on: the total number of patients annually seen at the department/unit; the types of patients seen; and the number of fittings carried out annually of upper and lower limb prostheses, upper and lower limb orthoses, spinal orthoses, orthopaedic shoes, wheelchairs and finally car adaptation. The third section, dealing with the total area of the facility and areas of each section separately, was included to obtain background information on the ratio of staff and/or patients to the area. The fourth section is concerned with personnel numbers and qualifications to obtain accurate information on the qualified and non-qualified personnel and their nationalities. The next three sections are concerned with the need for establishing prosthetic/orthotic services and asks more specific questions concerning the opinions of the hospital staff as to whether a workshop producing plastic splints only is considered enough for their requirement.

The final section of the questionnaire deals with comments and remarks regarding future plans for improving the existing facilities or providing new services in order to ascertain the respondents' personal views on the prosthetic/orthotic requirements in accordance with their local needs based on their experience.

Results

Results received from eighty out of the one hundred and twenty questionnaires circulated to hospitals and medical rehabilitation centres were divided into two groups based on the availability of medical rehabilitation services. The results of the respondents with one or more of the four major fields of medical rehabilitation (Prosthetics/Orthotics, Physical Therapy, Occupational Therapy, and Speech and Hearing Therapy) were analysed by dividing them into two groups based on the existing prosthetic/orthotic facilities. The following are hospitals and centres providing prosthetic/orthotic services.

A. Ministry of Health existing facilities:

1. Riyadh Medical Rehabilitation Centre (RMRC) was established in 1974 and serves as the principal supplier of prostheses/orthoses to patients from the Kingdom as well as neighbouring countries. The medical service is provided by a full-time expatriate lady physiatrist and two orthopaedic surgeons who attend weekly out-patient clinics. There is no in-patient facility. Only 6 out of the 12 qualified technicians and 20 assistant technicians are Saudi citizens (Table 1).
2. Makkah Medical Rehabilitation Centre was established in 1978 to serve as a main supplier to the Western Region. This centre is fully equipped and staffed by 18 expatriate technicians.
3. Abha Medical Rehabilitation Centre has a small prosthetic unit with 2 technicians, no Saudis. The unit is located within Abha General Hospital compound.
4. King Fahad Hospital – Madinah: the prosthetic/orthotic unit was established in 1986 in a remodelled building close to the hospital. It is suitably equipped and staffed by 4 technicians who are expatriates. Currently it is operating at a reduced level dut to shortages in materials and com-ponents.
5. King Fahad Hospital – Gizan: a prosthetic/orthotic unit was set up in 1984. It is well equipped and staffed with one well-trained technician, head of unit, and 4 qualified technicians, all expatriate. Some of the equipment must be modernised, and also there is need for plastic material and upper limb components.

B. Non-Ministry of Health facilities:

These provide quality services to certain groups of people and are as follows:

6. King Faisal Specialist Hospital – Riyadh: it has a relatively active prosthetic/orthotic unit staffed with 2 certified prosthetists/orthotists and one technician, who are all expatriates.
7. Armed Forces Hospital – Riyadh: it has a fully equipped large prosthetic/orthotic unit staffed by 2 qualified prosthetists/orthotists and 6 technicians who are all expatriates.

Table 1. Shows the distribution of prosthetic/orthotic facilities and the personnel number.

Centre Number	Name of Centre/Unit	Location	Personnel		
			National	Exp.	Total
1	Riyadh Medical Rehabilitation Centre (MOH)	RIYADH *(Central)*	6	26	32
2	Makkah Medical Rehabilitation Centre (non-MOH)	MAKKAH *(Western)*	—	18	18
3	Abha Medical Rehabilitation Centre (MOH)	ABHA *(Southern)*	—	2	2
4	King Fahad Hospital (MOH)	MADINAH *(Western)*	—	4	4
5	King Fahad Hospital (MOH)	GIZAN *(Southern)*	—	4	4
6	King Faisal Specialist Hospital (non-MOH)	RIYADH *(Central)*	—	3	3
7	Armed Forces Hospital – Riyadh (non-MOH)	RIYADH *(Central)*	—	8	8
8	Armed Forces Hospital – Al-Hada (non-MOH)	TAIF *(Western)*	—	5	5
9	King Khalid University Hospital (non-MOH)	RIYADH *(Central)*	—	2	2
10	National Guard King Fahad Hospital* (non-MOH)	RIYADH *(Central)*	—	—	—
Total		No.	6	72	78
		%	7.69	92.31	100

*Fully equipped workshop but not operational yet.

8. Armed Forces Hospital – Al-Hada (Taif): it has well equipped workshops and highly qualified personnel. This large prosthetic/orthotic unit which is staffed by 5 qualified expatriate technicians produces all types of prosthetic/orthotic devices with the exception of upper limb prostheses.
9. King Khalid University Hospital – Riyadh: the orthotic unit is composed of a small workshop that delivers simple orthoses. It is staffed by 2 technicians, who are expatriates.
10. National Guard, King Fahad Hospital – Riyadh has a fully equipped prosthetic/orthotic workshop, but it is not operational as yet.

There is only one 3-year on the job technical training programme for 6 assistant technicians at Riyadh Medical Rehabilitation Centre. This programme is not recognised by the civil service bureau. This lack of recognition has resulted in difficulty in attracting candidates with reasonable levels of education which would allow them to continue their education and training in this field. Since the current trainees have only elementary school certificates as their basic entry requirement, intensive teaching and supervision have been provided to bring their standard to the appropriate level.

The Joint Centre for Research in Prosthetics/Orthotics (King Saud University and Ministry of Health) is the only research and development facility in the Kingdom. It was established in 1987. It operates fully equipped orthopaedic assessment, biomechanics, and gait laboratories.

Clinical Data

During 1987 – 1988 period, the total number of patients receiving prostheses/orthoses from the nine operational Ministry and non-Ministry of Health centres/units was 8,534 (Table 2). Of these patients, 30% were female and 70% male. Of the total number of orthoses supplied, the commonest variety was shoe adaptation (26%); followed by knee-ankle-foot orthoses and spinal orthoses at 22% each, wrist-hand orthoses (10%), ankle-foot orthoses (8%), hip-knee-ankle-foot orthoses (5%), and surgical (orthopaedic) shoes (3%). Of the 635

Table 2. Shows various type of prostheses/orthoses produced (1987 – 1988).

Type of Appliances \ Centre Number		1	2	3	4	5	6	7	8	9	10	Total No.	Total %
Upper limb prostheses		36	22	13	6	5	6	Information was not readily available	—	—	Not Operational	88	1.03
Lower limb prostheses		206	117	39	23	60	52		50	—		547	6.40
Upper limb orthoses		188	67	18	150	220	73		55	320		1091	12.78
Lower limb orthoses		783	335	56	265	80	685		270	312		2786	32.69
Spinal orthoses		289	794	119	137	70	44		250	55		1758	20.62
Shoe adaptations		773	188	120	100	60	493		100	245		2079	24.37
Surgical shoes		94	88	3	—	—	—		—	—		185	2.16
Total	No.	2369	1611	368	681	495	1353		725	932		8534	100
	%	27.76	18.88	4.31	7.99	5.80	15.85		8.50	10.91			

prostheses delivered, 86.1% were lower limb and 13.9% were upper limb. The commonest variety was below-knee (44%); followed by above-knee (22%), partial foot (10%), partial hand (4.5%) and below-elbow (4%).

A separate questionnaire on the incidence of different cases and the causes of amputation was sent to Riyadh Medical Rehabilitation Centre which is the largest facility of is kind in the Kingdom, supplying 27.76% of the total number of prosthetic/orthotic devices. Results from the questionnaire indicated that poliomyelitis (37%) was the leading condition; followed by congenital deformities (23%) and amputation (16%) (Fig. 1). Since the polio cases reported represent the percentage referred to the centre, this does not necessarily show the true incidence of poliomyelitis in the Kingdom. With regards to causes of amputation, trauma was found to be the leading cause (40.72%), followed by diabetes/gangrene (24.95%), tumour (5.99%) and congenital (4.99%) (Fig. 2). Specific causes of trauma were road traffic accidents (29.94%), occupational (industrial) injuries and burns (9.98%), and snake bite (0.8%). It is striking to note that amputations due to road traffic accidents are the first cause, considering the high standard of road networks throughout the Kingdom.

Discussion

The majority of respondents took the opportunity to express further opinions, particularly with reference to the final section of the questionnaire dealing with comments, and although the quality and depth of the comments varied considerably, only one centre

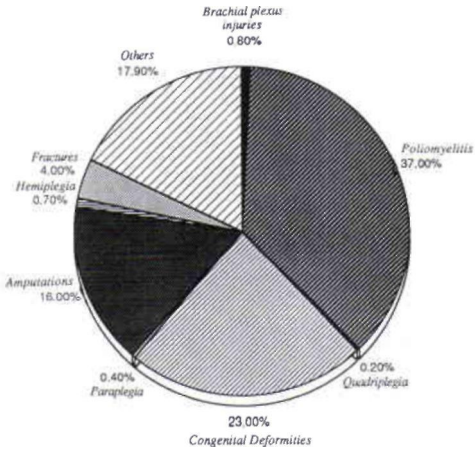

Figure 4. Incidence of Different Cases.

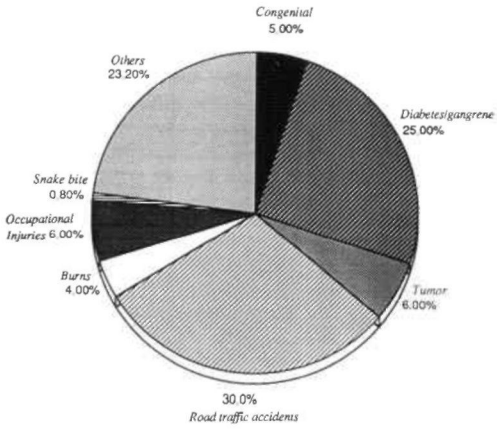

Figure 5. Causes of Amputation

failed to provide any additional remarks. Generally these remarks could be considered under three headings: lack of high qualified personnel, the problems associated with shortages of materials and components, and the requirements for future development.

In order to produce properly fitted and aesthetically acceptable prostheses/orthoses, a clinical prosthetist/orthotist is required, who is capable of interpreting the clinical needs to fulfil the prescription requirements of the clinician. He must be capable of supervising the prosthetic/orthotic technician and overseeing production. At the present time, most of the facilities throughout the Kingdom have only prosthetic/orthotic technicians. There is an urgent need to recruit highly qualified personnel as technical supervisers in order to make sure that the quality of prostheses/orthoses offered to patients is of an acceptable level, to introduce newly developed techniques, and to utilise newly designed and appropriate components.

More detailed comments were concerned with the desperate need for education not only for the clinicians who are the prescribers of the appliances, but also for the prosthetic/orthotic technicians who are fabricating such appliances. This education is considered vital both for the future development of the service and to upgrade the technical standards.

Although a few seminars have been held during the past years, organised by manufacturers in collaboration with different centres in Riyadh, these have been on lower limb prostheses which only represents 6.4% of the total appliances prescribed. Further they were rather considered as marketing seminars by these companies to introduce their newly produced components than occasions for providing the desperately needed short-term technical courses for upgrading and extending professional competence. Prosthetists/orthotists would also benefit from courses which reviewed the theoretical and practical aspects of prosthetics/orthotics for the most common conditions. As the majority of the referral prescriptions for prostheses/orthoses were inadequate, improper or redundant, it was also felt that the referring prescribers need more information and knowledge in the art and rationale of prosthetic/orthotic prescription. Therefore, short courses were strongly recommended for physicians, surgeons and therapists practising in this field to acquaint members of the clinic team with basic concepts and modern practices in prosthetics and orthotics.

Due to the large number of patients and the absence of an appointment system at the Ministry of Health Medical Rehabilitation Centres, practitioners have inadequate opportunity to assess or discuss existing conditions to ascertain the needs of the patients. It would, therefore, be justifiable to provide additional medical staff with special interest in rehabilitation and physical therapy, as well as to include occupational therapists in order to give more appropriate pre- and post-prosthetic/orthotic training.

The main bulk of the specialised prosthetic/orthotic services is concentrated in Riyadh (Table 1). Most of the facilities are well equipped, but the overwhelming majority of rehabilitation workers are still expatriates (92%). Despite the fact that the survey showed that 30% of the total number of patients who required prosthetic/orthotic services were female, there was no female practitioner/technician employed by any of the surveyed centres/units. To help rectify the current situation, recruitment and education of national professional manpower must be promoted at different levels, taking into account the desperate need for and important potential of female participation, by establishing a high calibre prosthetic/orthotic education programme within the context of a university.

It was also felt that the type of prostheses/orthoses prescribed for patients depends on the availability of materials and components rather than on the actual need of the patients. This is true for all the services, including those provided by the non-Ministry of Health sector. Linked to this, a number of respondents, in the centres in remote areas, stated that they had shortages of the most essential materials. It is also interesting to note that there was only one centre that offered a few externally powered upper limb prostheses of the myoelectric type. This is probably due to the non-availability of occupational therapy departments which usually provide the amputees with appropriate prosthetic training.

Considering the high percentage of amputations due to road traffic accidents

(29.94%) and the high number of amputees between 20 to 40 years of age, who are the ideal wearers of hydraulic knee joints and energy-storing feet, it is unfortunate, that such components have not been made available to the amputees at any of the centres in the Kingdom.

Ischial ramal containment sockets for the above-knee amputee have not even been introduced yet.

Although there are a great number of disabled children and adults in desperate need of specialised seating and mobility aids, it is disappointing to observe that such services are neither provided nor planned for. As far as car adaptation is considered, the only available services are provided by private automobile maintenance workshops (centres) which are not subjected to any kind of clinical/technical quality control.

Although there were no comments regarding the area of facility, the main centre in the Kingdom is located in a rented two-storey building, without an elevator, and has pre-fabricated buildings for prosthetic/orthotic workshops and prosthetic/orthotic technician training section with a very small area for patient fitting/training. The second biggest centre is located in a block of flats previously built to accommodate the nursing staff of the hospital.

Realising that the needs of the disabled in the Kingdom are not fully met by the existing facilities, the government has planned to establish six medical rehabilitation centres by the year 1993, these are at King Fahad Medical City and Eman Hospital in Riyadh, the Red Sea Hospital in Jeddah, the Gulf Hospital at Dammam, the Onaizah General Hospital in Qassim and the Beeshah General Hospital at Beeshah.

In the opinion of the authors, some deficiencies are regrettably due to the presence of a high percentage of expatriates who are employed despite certain limitations. These may be categorised as: intrinsic and extrinsic.

A. Intrinsic limitations:

1. Personal aims for taking employment in the Kingdom.
2. Cultural and educational incompatibility,
3. Language difficulty.
4. Lack of commitment.
5. Tendency to utilise their country's products out of loyalty and familiarity which consequently limits the availability of other materials.
6. Inadequate level of qualification.

B. Extrinsic limitations:

1. Lack of continuity leading to instability and inadequate planning for future development.
2. Less job competition giving rise to less incentives for upgrading professional capabilities.
3. Lack of authority granted to principal workers.
4. Lack of awareness amongst administrators as to the acceptable level of the services provided.
5. Fear of contract termination, if conflict arises with superiors.
6. Lack of professional interaction and planned management of the disabled.

Conclusion

This survey, indicates that the quality of prosthetic/orthotic services provided by the Ministry and non-Ministry of Health existing facilities falls short of the ideal and that there is a need for very considerable improvement in quality and delivery of service and research.

The profession of prosthetics/orthotics is beset by three major problems:

1. low number of national staff leading to less continuity;
2. too few clinical (properly qualified) prosthetists/orthotists;
3. lack of continuing education and training programmes.

The interaction of these problems has created a lack of confidence and dissatisfaction with the services. The appointment of junior or inexperienced staff, both medical and paramedical, to attend to the disabled, the attitude of key personnel and the poor response or lack of response from senior staff to the need to establish a supervisory body in the Ministry of Health, seriously aggravate the situation and are only a reflection of the lack of concern for this least fortunate group of people and this important field of medicine and rehabilitation.

Upgrading the current services could be achieved through the following:

1. recruitment of highly qualified staff;

2. provision of materials and equipment;
3. introduction of prosthetic/orthotic safety standards;
4. establishment of a supervisory body within the Ministry of Health;
5. establishment of a professional body of prosthetists/ orthotists in order to assist in organising short-term courses and to set standards in this field;
6. introduction of a B.Sc. prosthetic/orthotic training programme in the context of the university. A technician training programme should also be instituted.

Such recommendations would, if implemented, enable the recruitment and education of more national professional manpower, utilising the existing resources effectively and expanding services methodically, as well as ensuring proper co-ordination between institutions and professionals caring for the disabled.

Acknowledgements

The data presented in this survey have been provided by those who kindly gave precious time to completing the questionnaires. The authors are grateful to them, and sincerely hope that the data which have resulted from the survey will help promote prosthetics/orthotics in Saudi Arabia. The authors are also indebted to all members of the Ministry of Health Rehabilitation Committee and to their colleague, Mr. Hamayun Zafar, for their valuable assistance.

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The ORLAU VCG (variable centre of gravity) swivel walker for muscular dystrophy patients

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Abstract
Swivel walkers are being increasingly used for muscular dystrophy patients in order to prolong the period of their ambulation. Existing designs did not address the special problems of accommodating such patients comfortably and providing the easier and more assured ambulation which their weakened condition requires. The ORLAU VCG (variable centre of gravity) swivel walker has been developed so that the walking mechanics can be adjusted independently of patient posture. Additional patient support features permit the patient to be secured in their chosen position of comfort prior to setting the ambulation mechanics.

Patients using the device, which is now approved for supply by the Department of Health in England and Wales, have improved their walking performance and extended their period of walking.

Introduction

Swivel walkers have long been established as a means of ambulation for particular groups of paralysed patients (Motloch and Elliot, 1966; Edbrooke, 1970; Rose and Henshaw, 1972; Stallard *et al.*, 1978; Butler *et al.*, 1982; Farmer *et al.*, 1982). Their advantage is that they permit heavily handicapped individuals to walk with a high degree of stability, without the use of additional walking aids such as crutches.

More recently it has been recognised that swivel walkers have the potential to extend the period during which muscular dystrophy patients can continue to ambulate. They are used when the patient would otherwise be wheelchair bound. Initially standard designs of swivel walker were used, but it became apparent to those responsible for the ongoing care of these patients that they had unique problems which existing designs did not address. Assessment of these made it clear that a swivel walker with additional features was required in order to accommodate muscular dystrophy patients more comfortably and pemit them more effective ambulation.

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Orthotic problems of muscular dystrophy

Careful analysis of muscular dystrophy patients showed they had the following particular difficulties in comparison with other groups (primarily paraplegic) using swivel walkers:

1. sensitivity to posture greater than in other pathologies;
2. variability of hip and knee contractures;
3. sensation in the lower limbs which can result in discomfort at the patient/orthosis interface;
4. proprioception of hip position which can lead to a wish for greater abduction;
5. general weakness, which limits the input of propulsion forces;
6. apprehension of unsteady support and excessive step-length;
7. difficulty with the transfer of heavy, weak patients into the device for physiotherapists and parents.

Design features

Because of the problems experienced with existing swivel walkers, a new design called the ORLAU VCG (variable centre of gravity) swivel walker (Fig. 1) has been developed which has special features that fall into two main categories:

1. postural support;
2. ambulation mechanics.

Postural support in the VCG swivel walker is provided, as in conventional swivel walkers, by 4 point fixation (Stallard *et al.*, 1986) but incorporates additional adjustability for positioning of the feet through variable heel cups in order to permit careful alignment of posture.

The postural deformities which the VCG swivel walker is designed to accommodate are:

1. equinus;
2. knee flexion;
3. hip flexion;
4. hyperlordosis of the spine.

The contractures which cause these deformities tend to develop with the progress of the disease and can also be variable in nature over short time spans. To address these problems the VCG swivel walker has:

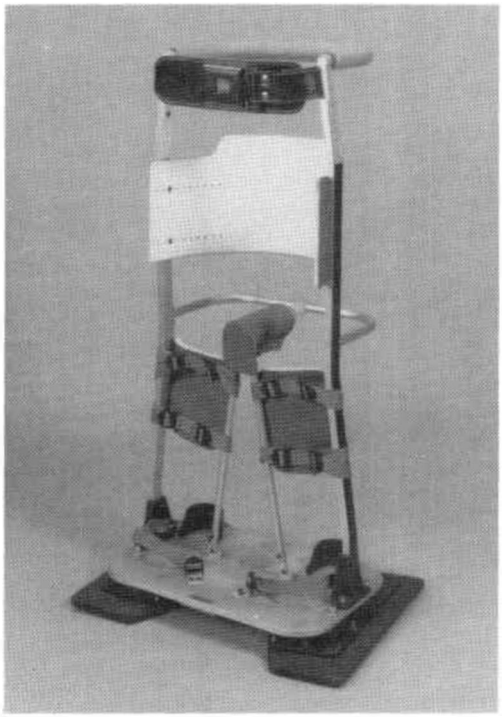

Fig. 1. The ORLAU VCG swivel walker.

1. the facility for insertion of compensatory wedges for equinus;
2. a range of easily adjusted knee pads with sheepskin interfaces;
3. an adjustable sacral band with an Evazote lining for patient comfort;
4. a range of thoracic bands which permits the attainment of the appropriate compromise between convenience and patient comfort.

The Swivel walker ambulatory mechanism (Stallard *et al.*, 1986; Rose and Henshaw, 1972) consists of two linked footplates, free to swivel about an essentially vertical axis, mounted beneath the baseplate, each one indexing forwards as the patient rocks from side to side.

Ambulation is more difficult for muscular dystrophy patients because of their general weakness. Consequently they require more careful adjustment of the swivel walker mechanism. This problem is aggravated by their postural sensitivity which prevents adjustment being achieved through changes in postural alignment, as is conventional in standard swivel walkers.

The two main factors affecting ease of ambulation in a swivel walker are:

1. position of centre of gravity relative to the footplate bearing centre in the sagittal plane. Ideally it should be 18—25 mm forward of this;
2. distance between footplate bearings in the coronal plane. The closer they are the easier it is to ambulate (although the step length and speed of walking are consequently reduced).

Both of these factors have been made more readily adjustable in the VCG swivel walker through a new design of baseplate. This consists of a double plate arrangement which allows the upper part of the swivel walker frame, including the patient support structure, to be moved forwards or backwards relative to the lower plate, to which are attached the footplates. This enables the orthotist to adjust the patient's centre of gravity relative to the footplate bearing centre to the optimum position without altering the patient's posture in the swivel walker frame. The footplate bearings are bolted to the lower plate and their relative spacing in the coronal plane is adjustable via a series of additional holes.

In order to cope with the apprehension which muscular dystrophy patients have of unsteady support, additional stability is provided by extended footplates.

Conclusion

Physiotherapists treating patients in a clinical trial of the new design reported that it had beneficial effects on patient confidence with commensurate improvements in ambulatory performance. The additional comfort and assurance which the patients attained further extended the periods of their walking.

It is very important that swivel walkers for patients with muscular dystrophy are used within a fully planned treatment regime for the individual. Appropriate control of clinical supply is vital if the best interests of patients are to be served. An important aspect of the orthotist training for the ORLAU VCG swivel walker is the philosophy of supply within a co-ordinated treatment regime. This is intended to eliminate inappropriate and harmful prescription and for this reason orthotist training is a mandatory condition for supply of the device to an orthotic contractor.

The ORLAU VCG swivel walker is now routinely available on Department of Health contract in England and Wales with the proviso that it **must** be fitted by an orthotist who has attended the Department of Health approved swivel walker course run by ORLAU.

Acknowledgements

The authors gratefully acknowledge the assistance and support of the physiotherapy staff and patients of the Hebden Green School, Winsford and Ysgol Erwir Delwy, Cardiff. Without their generous assistance the clinical trials which permitted the successful outcome of this orthotic development would not have been possible.

REFERENCES

Butler PB, Farmer IR, Poiner R, Patrick JH (1982). Use of the ORLAU swivel walker for the severely handicapped patient. *Physiotherapy* **68,** 324–326.

Edbrooke H (1970). The Royal Salop Infirmary "Clicking Splint". *Physiotherapy* **56,** 148–153.

Farmer IR, Poiner R, Rose GK, Patrick JH (1982). The adult ORLAU swivel walker — ambulation for paraplegic and tetraplegic patients. *Paraplegia* **20,** 248–254.

Motloch WM, Elliot J (1966). Fitting and training children with swivel walkers. *Artificial Limbs* **10 (2),** 27–38.

Rose GK, Henshaw JT (1972). A swivel walker for paraplegics: medical and technical considerations. *Biomed Eng* **7,** 420–425.

Stallard J, Farmer IR, Poiner R, Major RE, Rose GK (1986). Engineering design considerations of the ORLAU swivel walker. *Eng Med* **15,** 3–8.

Stallard J, Rose GK, Farmer IR (1978). The ORLAU swivel walker. *Prosthet Orthot Int* **2,** 35–42.

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Primary orthotic treatment of ruptured ankle ligaments: a recommended procedure

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Abstract

The results of a study after 1 and 2 years of a prospective randomised trial of operative versus conservative treatment of ankle ligament rupture, demonstrate that purely functional orthotic therapy is the method of choice. This relates both to patient need and economical considerations. The trial demonstrated that without an operation it was possible to achieve a high degree of mechanical stability, a reduction of work disability time down to 3 weeks and full sports capability within 3 months. Consequently, and as a result of the trial, the only remaining surgical indications would seem to be dislocations of the foot and ankle, ankle ligament rupture with additional intra-articular pathology, and second-stage injuries or re-ruptures.

The joint-stabilising function of the prototype splint developed in this study was improved on the basis of experimental investigations, using a Y-shaped leather band (designated CALIGAMED), which is available in 6 sizes for right and left ankle.

Introduction

Since the early eighties there has been considerable controversy in Europe regarding the treatment of severe ankle sprains with ruptured lateral ligaments. This is the most common injury of the recreational and school athlete and most patients are between 15 and 25 years old. There are important socio-economic implications to this injury in respect of medical expenses as well as time off work and also disability payments through the workman's compensation system. The 1985 rehabilitation study of the workman's compensation system in Germany (REHA, 1985) demonstrated, that during the twelve month period 13,554 patients were hospitalised for an average of 12.6 days for the treatment of ruptured ankle ligaments as an isolated injury. In the light of these important economic implications and with the instruments of modern medical research currently available, it was felt that treatment strategies should no longer be based on personal judgement, general trends or retrospective studies. Experimental studies and prospective randomised trials are necessary to make these important decisions.

A number of clinical trials which compared conservative immobilisation with operative treatment could not show any difference in the results between these two treatment modalities (Brooks *et al.*, 1981; Evans *et al.*, 1984; Klein *et al.*, 1988; Niedermann *et al.*, 1981; Zwipp, 1986). Other authors have published excellent results with early functional treatment with tapes, orthoses and special shoes (Brooks *et al.*, 1981; Hoogenband *et al.*, 1982; Hoogenband and Moppes, 1987; Jakob *et al.*, 1986; Klein *et al.*, 1988; Korkala *et al.*, 1987; Neumann, 1987; Stover, 1980; Stover 1986; Wetz *et al.*, 1987; Zwipp *et al.*, 1986).

Study design

A prospective randomised trial on the treatment of ruptured ankle ligaments was carried out at the Trauma Department of Hannover Medical School between 15th April,

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Group	Treatment	N	excellent	good	sat.
A		52	76%	21%	3%
B		50	89%	11%	0%
C		48	76%	21%	3%
D		50	79%	21%	0%

Fig. 1. Treatment groups and ankle performance score results averaged over all 3 follow-up periods.

1985 and 31st July, 1986. Two hundred patients were randomly assigned into 4 groups in respect of treatment method.

Fifty patients in each group were evaluated on a 100 point scoring system based on clinical, radiographic and dynamometric results. They were evaluated 3, 12 and 24 months after the injury (Fig. 1).

The groups were classified as follows:

Group A: Surgery and cast immobilisation.

Group B: Surgery and early functional orthotic treatment.

Group C: Conservative treatment and cast immobilisation.

Group D: Conservative and early functional orthotic treatment.

The study was evaluated and accepted by the Institution's ethical commission.

Group A

Within 60 hours after the injury surgery was performend under general, or regional, anaesthesia. A lateral malleolar approach to the ankle was used, as previously described (Zwipp *et al.*, 1986). The patients were hospitalised for one or two days and the involved extremity was immobilised in a short leg plaster cast and elevated on a foam splint. Oral non-steroidal anti-inflammatory agents and mini DHE dose Heparin were administered while the patients were in hospital. Active range of motion exercises to 10° of plantarflexion and 20° of dorsiflexion were initiated on the first postoperative day. Between the 5th and 8th postoperative days, a lower leg walking cast was applied in a neutral ankle position and used until 5 weeks after the operation.

Group B

Surgery and medical management were the same as for Group A. However, 8–10 days after the operation this group was fitted with a newly designed small splint which provides protection against supination and also incorporates a pronation wedge. This splint can be worn in a sports shoe and patients were instructed to wear the orthosis continuously day and night. Co-operative patients were allowed to remove the orthosis to shower. Patients were allowed to drive a motor vehicle as soon as pain free full weight-bearing was possible. Light work was allowed after 3 weeks.

Group C

These patients were initially immobilised in a short leg plaster cast. This was removed at about 3–5 days post-injury when the haematoma around the malleolar aspect had subsided with the aid of oral non-steroidal anti-inflammatory agents and elevation of the affected limb. A walking cast was then reapplied moulded in a position of pronation and eversion of the foot as recommended by Schatzker (1984). This was retained for a total of 5 weeks post-injury.

Group D

This group was initially immobilised in a short leg plaster cast for 3–5 days. They were then fitted with the functional orthosis used on Group B. The total period of functional treatment was again 5 weeks. The patients were allowed to drive some days after injury, when full weight-bearing was pain free. Most patients returned to work within 3 weeks after the injury.

For all patients, after removal of the cast or discontinuation of the use of the ankle splint, a series of 6 sessions of physical therapy was prescribed with special emphasis on proprioception and pronator muscle training.

The development of the functional orthosis and its use

The intention was to create a technical aid which would provide stabilising functions of conventional plaster cast therapy (short-leg cast) while utilising the pathological and kinetic knowledge behind modern therapeutic

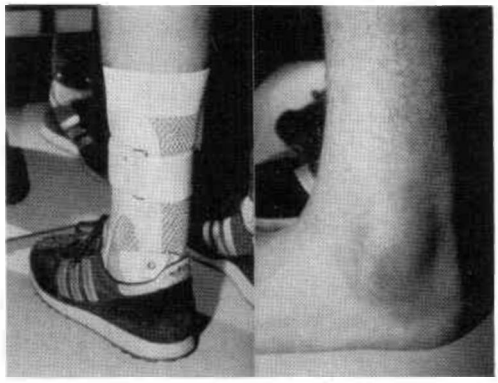

Fig. 2. The MHH splint.

concepts, especially in the case of injuries to the outer ligaments. The result of this work is the MHH (Medizinischen Hochschule Hannover) ankle splint (Fig. 2).

The functional principles of the therapeutic concept were to protect against supinatory sprain through stabilisation of the ankle and talocalcaneal joints while allowing the wound and swollen areas in the outer ankle area to remain accessible.

The design was developed through a number of successive prototypes before a practical solution was achieved. This involved application trials which benefitted, in particular, from trial fittings on the initial patients.

The large number of plaster impressions made to provide patient data gradually revealed certain consistent features, since the patients, for the most part, had no major anatomical abnormalities. There was also the urgent demand for a technical aid that could be applied immediately.

For these reasons, the authors' workshop constantly produced new, improved, versions prepared in the form of raw thermoplastic blanks requiring only customised fitting to the individual patient. Since the splints did not generally constitute a fixed element in therapeutic plans for longer than 4–6 weeks, costly and complicated cushioning and leather reinforcement were deemed unnecessary.

The best structural method of achieving relief of motion of the outer ankle ligaments is by the use of a moulded thermoplastic orthosis incorporating a pronatory wedge and in combination with strong elastic adhesive bands. This fulfilled the basic requirement of limiting undesirable tilt in the entire foot and especially in the outer ankle area.

The device described here can be compared to a shoe inlay. During the day it is worn inside a normal shoe of a size sufficient to provide room for non-attached insoles. The shoe should have a flat heel profile. Used additionally and simultaneously as an overnight and bedding aid, this special splint form can be worn practically "round-the-clock".

Three closure tapes wrapped around in opposite directions to provide reaction forces give the MHH ankle splint an adaptive but solid fit on the foot. A modification of this orthosis, described as the CALIGAMED ankle splint, was developed with anti-supinatory stabilising components to resist tilt and displacement of the talus under extreme varus stress load. An important element is the Y-shaped leather cuff, which encloses the ankle area tightly from the inner side, while leaving the wound surface accessible (Figs. 3 and 4).

Therapeutic success depends, amongst other things, on total patient co-operation. When technical aids are prescribed, careful instructions for their use must be provided. It is also important to make regular physical checks

Fig. 3. The MHH splint (right side) and the CALIGAMED splint (left side).

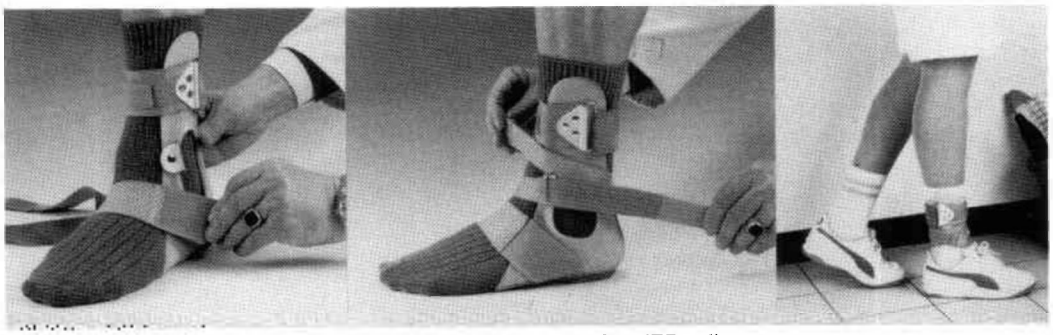

Fig. 4. Application of the CALIGAMED splint.

on the use of the device during the course of therapy. A special patient brochure was written in conjunction with the development of this ankle splint, which conveys important information on the condition, the orthosis and the likely progress of treatment.

The splint may be removed 5 weeks after the accident. This is followed by physical therapy. This consists of at least 6 repetitions of muscle-strengthening exercises (dorsiflexion of foot) and reflex training (balancing exercises, jump-rope, etc.). This follow-up treatment is important and extremely necessary in order to regain control over the foot for walking on uneven surfaces.

Results

In the operative Groups A and B there were no cases of wound infection or neuroma and only 6 cases of dysaesthesia in the vicinity of the wound. A small wound dehiscence developed in 2 patients. These wounds eventually healed with conservative management without any residual problems. No manifestation of thrombosis or pulmonary embolism were observed.

The extent of the injury – single ligament (34%), versus double ligament (66%) – was evenly distributed in the treatment groups and each treatment group had approximately 50 patients. All other characteristics, i.e. age, sex, laterality, were also evenly distributed in the study populations.

At 3 months follow-up it was possible to evaluate 185 of the 200 randomised patients (93%). Statistically significantly better active range of motion was found for the patients in the primary functional group. The subjective feeling of stability was, however, better in the operative groups with a tendency towards less favourable impressions in the cast group. There were no significant differences in all other subjective criteria, such as fear of the ankle giving way, gait stability, and limitations in sports or work. The results of radiographic stress testing did not confirm the subjective impression and showed no statistically significant differences when all patients in all groups were compared. It was, however, observed that there was a higher percentage of absolutely stable ankle joints as determined by stress views in the operative groups. This difference did not reach statistical significance. No patient developed radiographic signs of degenerative joint disease.

As summarised in Figure 5, the evaluation score in relation to excellent performance was somewhat lower in the group which was operated and cast. When compared with all other groups there was, however, no statistically significant difference when analysis of variance was used.

At 12 months follow-up 168 patients (84%) were available for review. At this interval no significant differences were found between the 4 treatment groups with regard to clinical joint stability, range of motion, incidence of re-injury, subjective limitation, or joint instability graded by stress X-rays. Additionally, the ankle performance score was not significantly different between the treatment groups. After 24 months 159 patients (80%) were available for follow-up. After the one and two years follow-up it was observed that there was a more even distribution of joint stability patterns in the different treatment groups. The type of after-treatment did not seem to have an influence on joint stability. Subjective results, however, did not reflect these findings. Neither the rate of recurrent "giving way", nor the

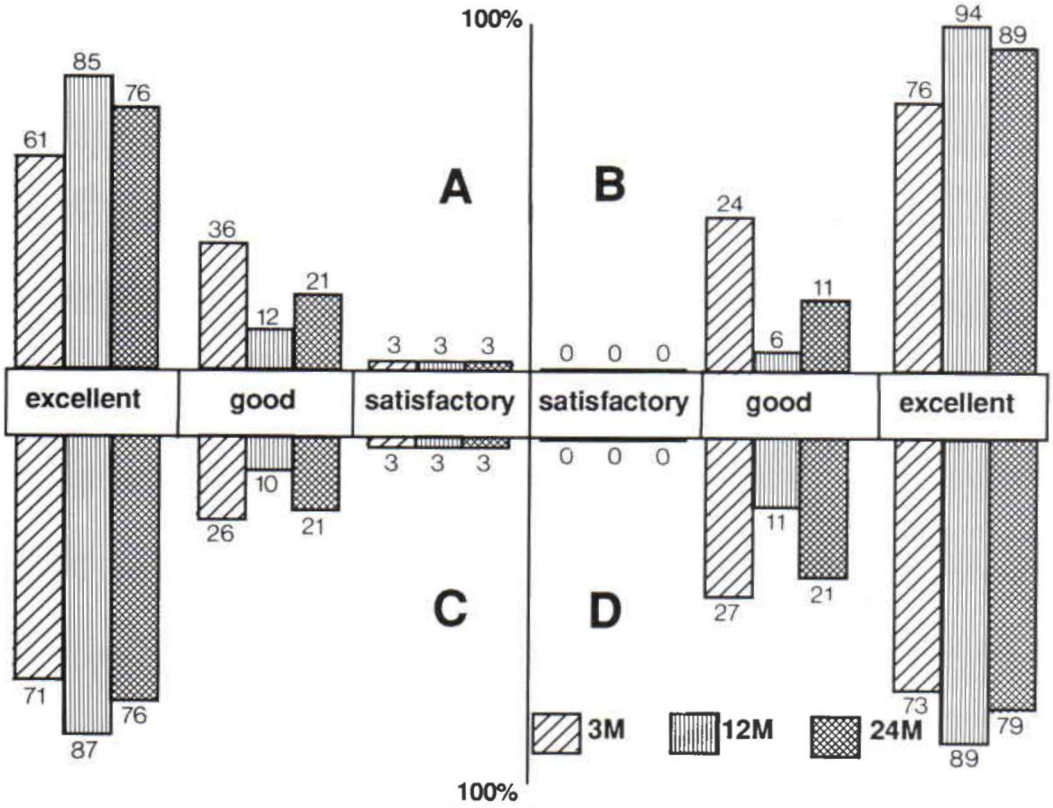

Fig. 5. Detailed description of ankle performance score results.

subjective grading of joint stability was significantly different within the treatment groups. No radiological signs of progressive degenerative joint disease were observed. The functional score again did not reveal a difference between the treatment groups as determined by analysis of variance.

When the results of the stress radiographs taken at follow-up were analysed separately for those patients with single versus double ligament tear, it was not possible to demonstrate different results for each period of follow-up with regard to clinical or radiological instability (Fig. 6).

Therapeutic results

Further experience with the use of the MHH ankle splints makes it clear that at least the following patient groups can be treated successfully:

1. Persons suffering from a lateral ligament rupture at the ankle joint:
   a) where the lateral ligaments require follow-up treatment after surgery (5 weeks);
   b) where, mostly without surgery, they are to be treated for purely functional reasons (5 weeks);
   c) where there is chronic instability and muscular decompensation and there is a requirement for external anti-supinatory foot support until the ankle joint is restabilised by specific pronator strengthening and proprioceptive training and/or surgery.
2. Persons who require a temporary foot support to treat a purely functional instability, post-traumatic sinus-tarsi syndrome, isolated instability of the calcaneal joint or other foot instability syndromes.
3. Persons requiring long-term external stabilisation of the ankle joint, in whom the ankle and/or calcaneal joints are unstable and for whom surgical

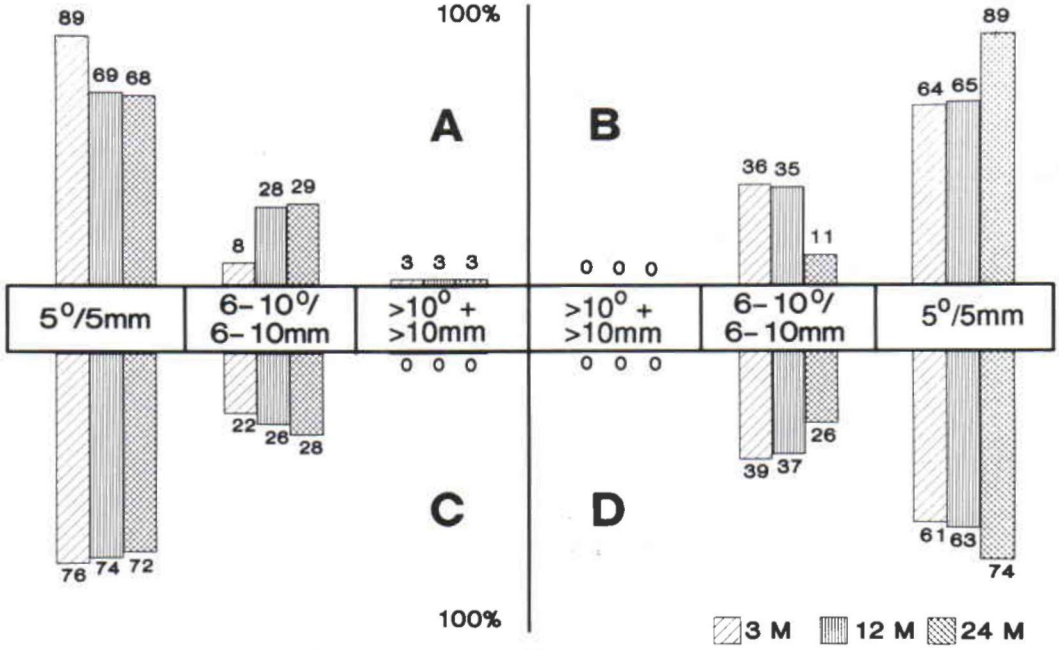

Fig. 6. Joint stability determined by stress X-rays.

stabilisation is contra-indicated when other conservative measures have failed.

Application of the splint concept has revealed several apparent advantages which must be subject to objective measurement and subjective evaluation:

1. Protection against rotation:
   a) external rotation through stabilisation of the ankle joint;
   b) supinatory: as a result of the force actions provided by the guidance bands.
2. Relief of the lateral malleolar ankle ligaments – through the heel effect of the pronation wedge.
3. Relief of pressure on the wound or haematoma in the lateral malleolar region – as a result of the large, pear-shaped window.
4. Protection of damaged ligamentous structures when putting on, or taking off, shoes through protection provided by orthotic device inside shoe.
5. Usable 24 hours a day, as night or positioning aide without shoes and as support splint in ordinary shoes.
6. Facilitation of patient co-operation: for hygiene splint may be removed for cleaning of foot and as a part of functional therapy during post-operative wound healing.

Discussion

A change in scientific knowledge and clinical practice formed the background of the development of this improved form of splint. The development, however, was only possible as a result of the closest collaboration between the medical specialist and the orthopaedic industry.

The indications for these ankle supports are, as a result of this trial, well defined and the clear finding that surgery is no longer a necessity for the majority of cases meant that a cost effective solution is available.

Orthopaedic manufacturers can, using modern practice, often make use of prefabricated construction elements to provide quick and dependable service to the clinic team. As a result of the development described above, there are now available:

- Splint blanks and prepared strapping in construction kit form (to be customised by the orthopaedic master craftsman and delivered to fill the clinical prescription).
- Completely prepared splint blanks (for rapid availability).

- Splint blanks (to be used by the physician in combination with tape bandages and self-adhesive wrapped bandages).

These product forms are available to satisfy individual requirements. Physician and manufacturer must both ensure that they are communicating properly in regards to the therapeutic requirements, as well as technical feasability. Medical practitioners and the manufacturing industry both share an interest in making a real contribution to lowering medical costs while ensuring and protecting improvement in patient therapy. The technical development of the MHH/CALIGAMED ankle splints described have, as supported by a controlled scientific study, contributed to these efforts.

Splints are now available in 6 sizes (each for left and right) for use ranging from child's size 27 to adult size 50 (measured in French "stitches"). This corresponds to English sizes from 9 for children up to 14 for adults.

REFERENCES

Brooks SC, Potter BT, Rainey JB (1981). Treatment of the partial tears of the lateral ligament of the ankle: a prospective trial. *Br Med J* **21 February,** 606–607.

Clayton ML, Miles JS, Abdulla M (1968). Experimental investigation of ligament healing. *Clin Orthop* **61,** 146–153.

Ent FWC vd (1984). Lateral ankle ligament injury. An experimental and clinical study. Proefschrift, Universiteit Rotterdam. Drukkery Elinkerijk BV, Utrecht.

Evans GA, Frenyo SD (1979). The stress-tenogram in the diagnosis of ruptures of the lateral ligament of the ankle. *J Bone Joint Surg* **61B,** 347–351.

Evans GA, Hardcastle P, Frenyo AD (1984). Acute rupture of the lateral ligament of the ankle: to suture or not to suture? *J Bone Joint Surg* **66B,** 209–212.

Gamble JG, Edwards CHC, Max SR (1984). Enzymatic adaption in ligaments during immobilisation. *Am J Sports Med* **12,** 221–228.

Hoogenband CR vd, Moppes FI v, Stapert JWIL, Coumans PF, Greep JM (1982). Konservative Behandlung der fibular-talaren und fibular-calcanearen Bandverletzung mit Coumans-Bandage, eine prospektive Vergleichsstudie. Abstraktband zur 130. Tagung der Vereinigung Nordwestdeutscher Chirurgen, 2–4. Dezember 1982, Hamburg.

Hoogenband CR vd, Moppes FI v (1987). Die Behandlung der lateralen Ligamentrupturen des oberen Sprunggelenkes mit der Coumans-Bandage und direkte Mobilisation (eine prospektive Vergleichsstudie), Teil 2. *Hefte Unfallheilkd* **189,** 1030.

Jnoue M, Gomez MA, Hollis JM, Roux RD, Lee EB, Burleson EM, Woo SL-Y (1986). Medial collateral ligament healing: repair vs non-repair. Vol 11: Transactions of the 32nd Annual ORS, New Orleans Louisiana, p 78.

Jakob RP, Raemy H, Steffen R, Wetz B (1986). Zur funktionellen Behandlung des frischen Außenbänderrisses mit der Aircast-Schiene. *Orthopäde* **15,** 434–440.

Klein J, Schreckenberger C, Roddecker K, Tiling TH (1988). Operative und konservative Behandlung der frischen Außenbandruptur am oberen Sprunggelenk: Randomisierte klinische Studie. *Unfallchirurg* **91,** 154–160.

Korkala O, Rusanen M, Jokipii P, Kytömaa J, Avikainen V (1987). A prospective study of the treatment of severe tears of the lateral ligament of the ankle. *Int Orthop* **11,** 13–17.

Neumann K (1987). Ist die konservativ-funktionelle Behandlung frischer Außenbandrupturen am OSG gerechtfertigt? Teil 2. *Hefte Unfallheilkd* **189,** 1018.

Niedermann B, Anderson A, Byrde-Anderson S, Funder V, Jörgensen JB, Lindholmer E, Fuust M (1981). Rupture of the lateral ligaments of the ankle: operation or plaster cast. *Acta Orthop Scand* **52,** 579–587.

REHA 85 Rehabilitation und Rehabilitationsstatistik der gesetzlichen Unfallversicherung (1985). Editor: Hauptverband der gewerblichen Berufsgenossenschaften e.V.

Schatzker J (1984). Personal communication.

Schievink B (1986). Verbesserte Stabilisierungsmaßnahmen im Bereich des oberen Sprunggelenks. Orthopädie-schuhtechnik.

Sommer HM, Arza D, Ahrendt J (1987). Behandlungsergebnisse von operativ und konservativ versorgten fibularen Kapselbandrupturen, Teil 1. *Hefte Unfallheilkd* **189,** 1012.

Stover CN (1980). Air stirrup management of ankle injuries in the athlete. *Am J Sports Med* **8,** 360–365.

Stover CN (1986). Functional sprain management of the ankle. *Ambulatory Care* **11,** 25.

Tipton CM, James SL, Mergner WT, Cheng T (1970). Influence of exercise in strength in medial collateral knee ligaments of dogs. *Am J Physiol* **218,** 894–902.

Weise K, Rupf G, Weinelt J (1988). Die laterale Bandverletzung des OSG beim Sport. *Aktuel Traumatol* **18 (Suppl 1),** 54–66.

Wetz B, Steffen R, Raemy H, Jakob RP (1987). Spätergebnisse nach konservativer Thearapie fibulotalerer Bandläsionen mit der Aircast-Schiene. *Schweizz Sportmed* **35,** 115–118.

Zwipp H (1986). Die antero-laterale Rotationssinstabilität des oberen Sprunggelenkes. *Hefte Unfallheilkd* 177.

Zwipp H, Tscherne H, Hoffmann R, Wippermann B (1986). Therapie der frischen fibularen Bandruptur. *Orthopäde* **15,** 446–453.

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Application of a shape memory alloy to hand splinting

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Abstract

This paper describes new passive splints which have been developed using a shape memory alloy. The peculiar feature of the splints is that the way in which they change shape in use conforms to the stretching motion which it would be desirable to apply in certain conditions of deformity.

The alloy consists of 55.66% by weight Nickel and 44.34% Titanium. The heat treatment of the alloy for memorising shape was implemented at 500°C for one hour. This alloy was easily bent when cool, but the original shape was recovered on heating.

It was used as the supporting structure of the reverse knuckle bender splint and the cock-up splint. The new splints could be easily attached to the deformed limb after cooling. The splints avoided the development of spasticity, because they gradually recovered their original shapes and corrected the deformities when the heat of the room or body heat warmed the splints.

Since the shape memory alloy has the dual function of thermal sensor and kinetic power source it was a simple device. The splint was, as a result, small and smart. It was apparent from clinical use that the splint was easy to wear and could be worn with comfort for an extended period.

The design of the splints and the fabrication process are described and their application is indicated.

Introduction

Shape memory alloy has attracted the authors' attention because it can be easily bent when below the transformation temperature, but the original shape is recovered on heating.

Since the shape memory alloy was discovered by Kurdjumov in 1949, many investigators have tried to find practical applications. The alloy has a dual function of thermal sensor and kinetic power source, hence many kinds of functional devices have been developed by its use. Nevertheless, the applications are mainly for industrial purposes and it would appear that no reports on the application to splints have been published.

If the alloy is used as a splint, what kinds of benefit are there, and how should it be applied? The purpose of this paper is to answer these questions. Furthermore it is considered whether the application makes the best use of the characteristics.

Splints can be classified roughly into two groups; dynamic splints and static splints. Although it may be expected that shape memory alloy is suitable for application to dynamic splinting because it automatically moves on heating, there are many problems to solve in the application of the principle. An electric control could be considered the most effective means of activation. The transformation of electricity to heat however is most inefficient. Furthermore, the responses of the alloy to temperature rise and fall are slow. Speed of activation, is however necessary in dynamic use.

The response of shape recovery on warming by using heat from body or room is very slow. The shape memory alloy may be applied as an assist in a static splint. Since the alloy remains the same shape for a time after its shape is changed on cooling, a splint made from it is

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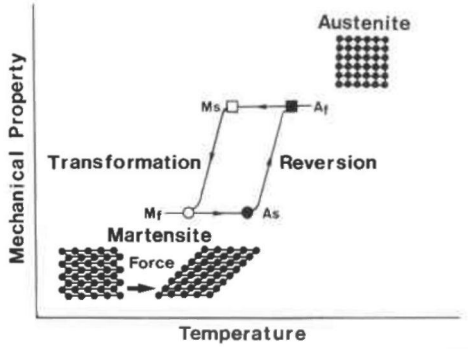


Fig. 1. Schematic representation of the mechanical property — temperature relationship for a thermoelastic martensitic transformation.
M_s: temperature starting martensitic transformation on cooling.
M_f: temperature finishing the transformation.
A_s: temperature starting the beginning of the reverse transformation.
A_f temperature finishing the reverse transformation.

easily attached to the deformed hand. The slow recovery speed on heating will not develop spasticity in the spastic hand. Accordingly, the application of the shape memory alloy to the static splint was tried. This is a preliminary report.

Methodology

Shape memory effect

In crystallography, it is understood that the shape memory effect is induced by thermoelastic martensitic transformation. The cause is explained by the fact that the crystal lattice is deformed without the diffusion of metal ions. Figure 1 shows the transformation-reversion mechanism of the shape memory effect. M_s is the temperature marking the beginning of martensitic transformation and M_f is the temperature marking its conclusion. A_s is the temperature starting the reverse transformation to the austenitic phase and A_f is the temperature ending it. If the alloy is cooled down, the martensitic transformation will start. The crystal lattice will zigzag and the alloy will be soft. In this state, the metal is easily deformed. Then, as the alloy is heated, a large stress generates and the reversion starts. The alloy "springs back" to its original shape and becomes hard.

Materials

It is widely known that Nickel-Titanium (Ni-Ti) alloy is available as a shape memory alloy. This alloy features a strong recovery force, high wear resistance, and resistance to corrosion. The safety of this alloy also has been proven in animal toxicity tests within a period as long as the splint is used. Hence, the Ni-Ti alloy has been selected as the assist of the static splint.

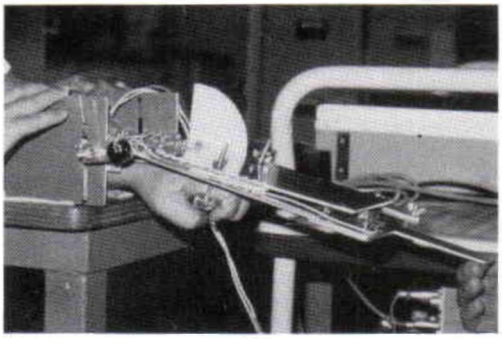
The transformation temperature is dependent upon the concentration of Ni and the temperature of the heat treatment needed to memorise the desired shape. The transformation temperature was measured with a differential scanning calorimeter. Since the transformation temperature of 55.66% by weight of Ni is close to room temperature, a Ni-Ti alloy* with this Ni concentration was chosen as the best material for the new splint. As the heat treatment temperature becomes higher, the transformation temperature drops, and the deflection range from zero load to the yield point is less. If the material is deformed past the yield point, the restoring force will not be able to correct the deformed hand. Thus, the heat treatment is implemented at 500°C for one hour. Consequently, the transformation temperature A_s is 13°C, A_f is 28°C, M_s is 24°C and M_f is 9°C.

The load versus angular movement characteristic

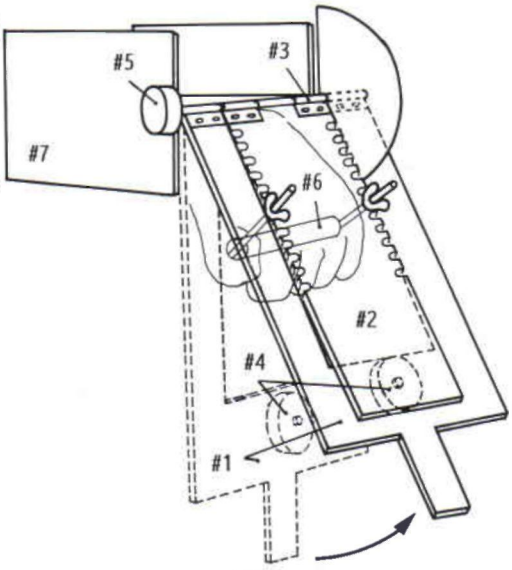
Before designing splints, it is necessary to know the bending force being exerted by the shape memory alloy to correct the deformed hand, as the size of the alloy will be determined by the force.

A device to measure the load against angular movement characteristic was constructed (Fig. 2). A duralumin frame (#1), which was cut to the shape shown in Figure 2b, and a rectangular duralumin plate (#2) was attached to a base. There was an overlapping area and they had the same axis of rotation (#3). The two plates were able to rotate smoothly about this axis. A disc load transducer (#4) was mounted in the overlapping area between the plates and a potentiometer (#5) was fixed to the shaft of the axis. The two opposite longer edges of the rectangular duralumin plate were U-notched at intervals of 1 cm. In these notches a hand rack (#6) was attached to transfer resisting force from a hand to the frame (#1). The load transducer and the potentiometer were connected by amplifier to an XY-recorder.

*Ni-Ti alloy is a product of HITACHI METALS Ltd., 1-2, 2-chome, Marunouchi, Chiyodaku, Tokyo, Japan. Tel: (03) 3284-4511.



(a)



(b)

Fig. 2. A device to measure load against angular movement characteristics using a potentiometer and a load transducer.

a) test in progress.

b) construction of the measuring device.

The deformed hand is inserted into the hand rack, and the palm or the fingers are put on its bar. The forearm is fixed to the base (#7) with the wrist joint close to the axis of the hinge. When the handle of the plate (#1) is pulled in the direction of dorsiflexion of the spastic hand by an operator, a resisting force is generated. This force is transmitted through the hand rack to the plate (#2). Pressure is applied to the load transducer which is between plates #1 and #2. The characteristic curves of the force detected from the load transducer and the angular movement detected from the potentiometer are drawn on an XY-recorder. The joint torque of the deformed hand is obtained by multiplying the force by the distance between the axis of the hinge and the hand rack.

An example of the curve of force against angular movement obtained is shown in Figure 3. The resistance shown by the joint to extension increases with increasing extension, that is, as the angle of flexion decreases. On the contrary, the restoring spring force of the shape memory alloy increases with the bending angle. Where these two curves intersect is the point at which the resistance to extension of the joint is balanced by the restoring force of the splint acting on the fingers.

Fabrication process

1) The function of the upper limbs of the patient are evaluated by an occupational therapist. Then the type of splint and corrective position are prescribed based on this evaluation.
2) The pattern of corrective force, which is the load against the angular movement, of the deformed hand is measured with the above device. The dimensions of the plate of the splint assist are chosen to give an equilibrium of forces at the angle at which correction is desired.
3) The splint assist plate is made by a machining process. Since Ni-Ti alloy is very hard, the plate is cut to the desired size from the alloy using a diamond cutter and holes are bored as required with a carbide drill. It is then formed into the required shape with forming tools. At this stage, it is necessary to keep below the limit of

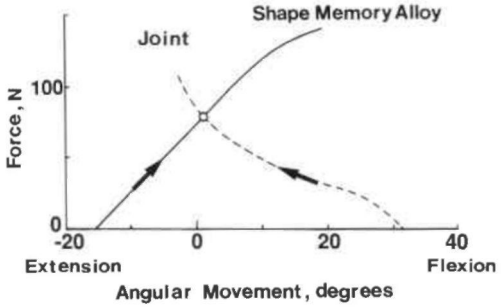
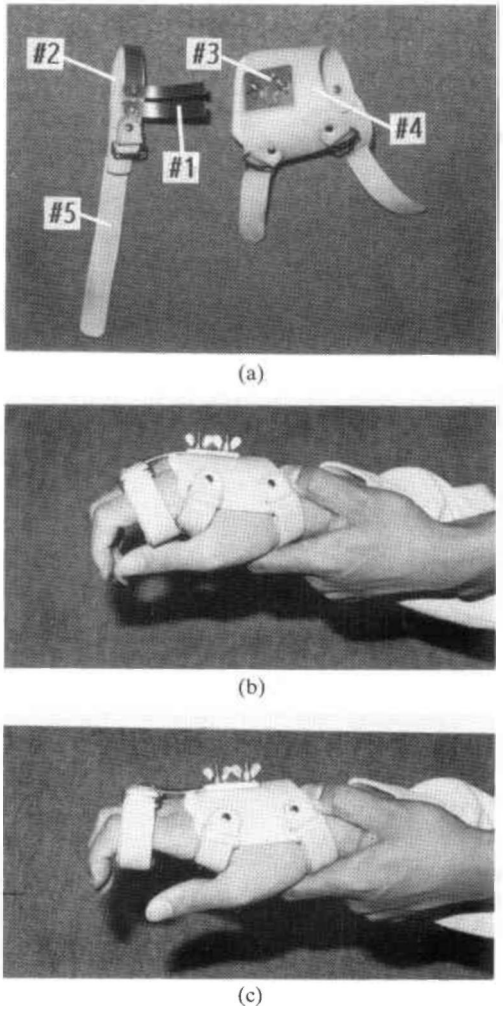


Fig. 3. An example of the curve of force against angular movement. The broken line shows the load characteristics for the metacarpophalangeal joint, and the solid line shows the characteristics for the alloy.

elasticity. Specifically speaking, the strain must be below about 6% for the bending of Ni-Ti alloy. In the case of a plate of 1.4 mm thickness, this limitation corresponds to bending with a radius of curvature of about 15 mm.

4) The shape memorising treatment is performed by fixing the shape, putting the splint assist into an electric hearth and heating for one hour at 500°C. It is then taken out and immediately plunged into cold water.

5) It is finished off into the shape required for the splint. The best connection of the plate to the substance of the splint is by a mechanical method, fastening with a bolt or rivetting for example. Connection by welding or brazing will be easily separated under a small change of temperature of the alloy and it should be avoided because the ratio of expansion or contraction is large. For example, the ratio is from 2% to 6% for Ni-Ti alloy and is approximately 0.5% for other metals in general.



(a)

(b)

(c)

Fig. 4. A new reverse knuckle bender splint.

a) the new splint using the shape memory alloy. Two plates of the alloy are used for the finger extension assists.

b) the splint before it was activated.

c) the splint after it was activated to correct the deformed fingers.

Usage

1) The splint assist is cooled in a refrigerator at 0°C for about ten minutes, and becomes soft. Rapid cooling may be achieved by using a spray type freezing gas, such as Three Bond PANDO-29C®.
2) The shape of the splint is then adjusted to fit the deformed wrist or hand snugly. The splint is then ready to be worn.
3) When the splint is worn, the heat of the room or body warms the splint and it recovers its original shape gradually; the recovery time is about 5 minutes. A heater or hot water may also be used as the means of warming the splint. The higher the temperature, the faster will be the recovery speed.
4) Care should be taken not to bend the splint assist until its temperature decreases below M_s, nor to warm it above A_f in use, since these cause loss of shape memory.
5) The splint is detached from the hand by loosening the straps which attach the splint to the hand.

Clinical applications

Reverse knuckle bender splint

The patient is a male, 27 years old, with quadriplegia caused by head trauma. His hand is in the fourth stage of the Brunnstrom's recovery stage. The MP joints are moderately deformed by spasticity of the flexors. The purposes of the splint prescription are to prevent the aggravation of spasticity and to improve the functions of the hand.

Figure 4a shows a new splint using the shape memory alloy. Figure 4b shows the splint before activation and Figure 4c shows the splint

after activation to correct the deformed fingers. Two pieces of plate of the shape memory alloy were used for the finger extension assists (#1). The dimensions of each plate were 10 mm wide, 40 mm long, and 1.4 mm thick. An MP bar (#2) was made of duralumin plate and was connected to the assists.

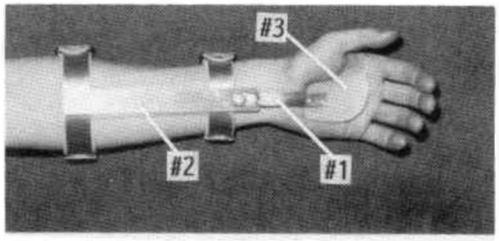
To obtain the optimum fitting position, a removable MP bar and a "slide fix" type mounting was adopted. The MP bar could slide along the attachment base (#3) and was bolted to it. Hence the MP bar could be attached rapidly to the base and the shape memory alloy could be easily cooled. The lateral sides of the splint were opened for easy insertion of the deformed hand. When the hand is dorsiflexed, there is a tendency for the transverse arch to flatten. To maintain the arch, the attachment #4, which was made of thermoplastic sheet was shaped to the transverse arch of the hand by casting. The MP bar is also shaped to the transverse arch. The wrist edge was trimmed to permit the range of movement of the wrist. VELCRO® straps (#5) were positioned so that they did not prevent the flexion of the MP joint. Urethane sponge sheets were pasted on the internal surface of the splint to distribute the pressure, and all rough edges were trimmed, and smoothed to prevent the tissue from being injured.

The new splint was easily attached to the deformed hand and was well fitted. Consequently, no pain was noted and the tissue did not turn red. The patient was able to wear the splint for a long period. After using the new splint, for 4 months the patient could extend his fingers fully. The spasticity decreased and the hand was improved from a non-functional hand to a functional assistive hand.

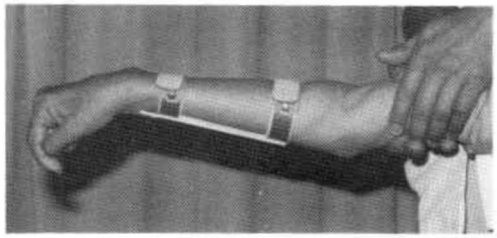
Cock-up splint

The patient, is male 66 years old, with left hemiplegia caused by subarachnoid haemorrhage. His hand is in Brunnstrom's third stage of recovery. The wrist joint is deformed severely by painful spasticity and contracture of the flexors. The purpose of the splint prescription is to correct the deformity.

Figure 5a shows a new splint using the shape memory alloy. Figure 5b shows the splint before activation and Figure 5c shows the splint after activation to correct the deformed wrist. A plate of the shape memory alloy was used for



(a)



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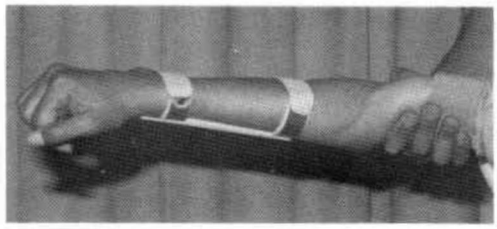

(c)

Fig. 5. A new cock-up splint.

a) the new splint using the alloy. One plate of the alloy is used for the wrist extension bar.

b) the splint before it was activated.

c) the splint after it was activated to correct the deformed wrist joint.

the wrist extension bar (#1). The size was 12 mm wide, 45 mm long, and 1.4 mm thick. The plate was fixed to the forearm piece (#2) by riveting. The forearm piece was made of duralumin plate with felt pasted on the inside surface and two VELCRO® straps added. A palm pad (#3) to distribute the pressure was fixed to the wrist extension bar by riveting.

The palm pad was made of thermoplastic sheet and was made in the shape of the arch of the palm by casting. The outline of the pad was trimmed to avoid the ball of the hand. The corners were rounded, and all edges smoothed off.

No pain was noted when the new splint was attached to the deformed hand, and the splint

was well fitted. The slow gradual stretching of the splint prevented the development of spasticity. The splint exerted sufficient force to correct the deformity, and expanded the range of motion (ROM) of the wrist after a few months.

Discussion

It is worth noting that the new splint prevented the development of spasticity.

The discharge in Ia fibres activates agonistic muscles and as a result, inhibits antagonistic muscles at the same time. It enhances agonistic muscle tone. On the other hand the activation of Ib fibres inhibits agonist and facilitates antagonist. Consequently they have a relaxation effect on the agonistic tone. The Ib fibres fire more slowly than the Ia, and the excitation continues for a time after stretching. When a muscle is stretched quickly, the Ib fibres hardly become sensitive to stretching and the Ia fibres fire, but when the stretching is performed slowly, the Ib fibres are excited and the agonistic tone is decreased. That is to say, when a spastic hand is moved suddenly, a spasticity occurs easily. If the deformed hand is corrected gradually, the occurrence of spasticity decreases.

The method of stretching by using this splint agrees well with the above-mentioned theoretical grounds. This is a peculiar feature of the new splint. The splint is attached to the deformed hand and stretches it slowly while being warmed by the heat from the environment.

Firstly, the Ib of flexor is fired for a while by using this splint, and subsequently, the extensor should be exercised voluntarily to strenghten it. Because it is necessary to increase the Ia fire of the extensor to decrease flexor spasticity more.

The recovery speed of the shape memory alloy is proportional to the rate of its increase in temperature. Since the temperature of a body or a room, which is the heat source to heat the alloy, is not high, the speed of temperature rise is slow and the recovery time is long. The speed can be slowed down by decreasing the conduction of heat to the alloy. Covering the surface of the alloy with a heat insulator is one of the methods for delaying recovery, and increasing the heat capacity by increasing the dimensions of the alloy is another.

For the purpose of decreasing spasticity, it is desirable that the wearing period is lengthy and the splint can be used during sleeping at night as described by Kaplan (1962) and Snook (1979). Although patients can understand the usefulness of a splint, in practice many of them hate it and are used to life without it. If a splint is small, smart and easy to attach, more patients might use it. These are important factors to continuous wear. Since the splint changes the shape gradually by gaining heat, it can correct the deformed hand while the patient is sleeping.

Furthermore, no assistant is needed to help the patient to extend the wrist or the finger joints before wearing the splint, since the new splint can be put on to the hand in the deformed state and the hand position corrected automatically. It is easy to wear independently. With other methods, it is necessary for a therapist to correct the deformed hand forcibly or to do ROM exercises sufficiently before wearing the conventional type splint.

As shown in both cases of reverse knuckle bender splint and cock-up splint, the shape memory alloy improved the wearing procedure of the splint and extended its wearing term. Since the alloy has a dual function of heat sensor and kinetic power source, it can eliminate components such as an electric motor and mechanical gears along with a sensor system, and the correcting mechanism of the splints is simple and small.

On the other hand, the high cost is one of the shortcomings of the shape memory alloy. Mass production will make the price lower, and the total cost of the splint can be decreased by using the minimum amount of the alloy required. Another shortcoming is that cutting and drilling of Ni-Ti alloy are difficult, because the alloy is harder than duralumin.

Particular attention is also needed in fabrication. Welding and brazing should be avoided as mentioned above. For the same reason, covering the alloy with a heat contractible tube is a better way of coating it.

The safety of Ni-Ti alloy has been proven in animal toxicity tests extended over some weeks. A thorough check over a long duration would also be necessary to confirm bio-compatibility.

Conclusion

The authors especially emphasize the point that using a shape memory alloy as the assist or the extension bar of a splint may be the most

desirable method of correcting a deformed hand with spasticity. The present experience is not sufficient to prove this fact statistically. More practical research is required.

REFERENCES

ANDERSON MH (1965). Upper extremities orthotics — Springfield: C C Thomas.

FESS EE, GETTLE KS, STRICKLAND JW (1981). Hand splinting. Principles and methods. — St Louis: CV Mosby Company.

FUNAKUBO H (1984). Keijou kioku goukin (Shape memory alloys). — Tokyo: Sangyou Tosho (Japanese).

KAPLAN U (1962). Effect of splinting on relfex inhibition and sensorimotor stimulation in treatment of spasticity. *Arch Phys Med Rehabil* **43,** 565–569.

LONG C, SCHUTT AH (1986). Upper limb orthotics. In: Orthotics etcetera. 3rd ed. Edited by Redford JB.— Baltimore: Williams and Wilkins. p 198–277.

PERKINS JS (1975). Shape memory effects in alloys. — Proceedings of the International Symposium on shape memory effects and applications held Toronto, Canada, May 19–22, 1975. Edited by Perkins JS. — New York: Plenum Press.

REDFORD JB (1986). Principles of orthotic devices. In: Orthotics etcetera 3rd ed edited by Redford JB — Baltimore: Williams and Wilkins. p 1–20.

SNOOK J (1979). Spasticity reduction splint. *Am J Occup Ther* **33,** 648–651.

SUZUKI Y (1987). Jituyou keijou kioku goukin (Practical use of shape memory alloy). — Tokyo: Kogyo Chosakai Publishing Co. Ltd. p 1–160 (Japanese).

TAKAMI M, SUGIYAMA I, FUKUI K (1988). Application of shape memory alloy to hand-splint. *Sogo Rihabiriteshon* **16,** 475–481 (Japanese).

TODERAN D, LUNSFORD T (1978). Tissue pressure tolerance as a guide to wrist-hand orthosis design. *Orthot Prosthet* **32,** 37–43.

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Case study

A remarkable transport device for a fibrodysplasia ossificans progressiva patient

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Abstract

This article contains a case report and a general description of a patient with fibrodysplasia ossificans progressiva.

This is followed by a description of the design of a standing and transport device.

Introduction

The Academic Medical Centre (AMC) in Amsterdam is one of the largest hospitals in the Netherlands. A great deal of scientific research is done at the AMC and it also accommodates a training centre (University of Amsterdam).

The rehabilitation department of this hospital encompasses the following disciplines:

- physiatry;
- physiotherapy;
- occupational therapy;
- orthopaedic instrument making;
- social work.

Because of the research facilities of the AMC, many rare diseases are seen in the hospital and are often sent for treatment to the rehabilitation department.

One of these patients was suffering from fibrodysplasia ossificans progressiva (FOP). His recurrent decubitus ulcers and problems in moving and standing were the reasons for his referral. After analysing the specific problems of this patient, a special standing and transport device was designed by an occupational therapist and a physiatrist.

General description of FOP

Fibrodysplasia ossificans progressiva (FOP, formerly known as myositis ossificans progressiva) is a rare disorder in which there is progressive ectopic ossification and characteristic skeletal malformation (Connor and Evans, 1982; Rogers and Blair Geho, 1979; Heutzer *et al.*, 1977). It is an autosomal dominant disease, but it usually occurs as a fresh mutation.

It is likely that there is a paternal age effect related to occurrence. Because of the early onset of symptoms there is a lack of reports concerning offspring. The skeletal malformation consists of:

- abnormal big toes. These are mostly short, monophalangeal with valgus deviation;
- short thumbs, due to short metacarpals;
- short broad femoral necks;
- abnormal cervical vertebrae with small bodies, large pedicles and large spinous processes.

Formation of ectopic bone usually starts in early childhood in the neck of the dorsal paraspinal muscles. The initial manifestation is the appearance of tender soft tissue swellings, which may progress in non-tender ossified nodes. Only four to six months after the appearance of these swellings there is radiological evidence of ossification. Certain areas of connective tissue are especially prone to ossification. These areas include the connective tissue of the paraspinal muscles, the muscles in the limb girdle and the mastication muscles. Joint capsules, ligaments and plantar fasciae are also often involved. Calcification, affecting the shoulders, elbows, hips and knees is commonly bilateral. The heart, tongue,

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larynx, diaphragm, perineum, eye and sphincters remain unaffected. Involvement of the abdominal wall is uncommon. There is no involvement of the skin by ossification, but ulceration of the skin over a projecting spur of bone or due to pressure sores is an occasional feature. Other features can be precipitated by a number of factors such as trauma to the muscle, biopsy of the swellings, operations to excise ectopic bone, intramuscular injections, careless venipuncture and dental therapy. Episodes of inactivity can last for many years. The ectopic ossification is most marked before puberty but new swellings can still occur in the sixth and seventh decades of life.

The progression of disability is not correlated with the sex of the patient, the age or site of onset of ossification, the type or extent of skeletal malformation, nor is it affected by medical treatment.

Because of the severe reduction of mobility, most patients either require assistance in activities of daily living or are totally unable to undertake any activities at all. Management of the patient should concentrate on the avoidance of exacerbating factors, prevention and prompt treatment of pneumonia, the provision of adaptations in order to cope with the activities of daily living, and the prevention of decubitus ulcers.

Case report

The patient described in this case report is a 41 year old man suffering from severe FOP. He has had ossifications since the age of seven. He was born when his mother and father were 32 and 37 years old respectively. There was no consanguinity of the parents. No members of the family had malformations or diseases of connective tissues (including two older brothers and one older sister). The pregnancy and delivery were without complications. At birth he showed a bilateral hallux valgus, a microdactyly of the great toes (monophalangic) and of the thumbs (due to short first metacarpals). The first swellings and ossifications occurred when he was seven years old, at the thoracic and cervical spine. Since then the disease has progressed steadily.

At the time of writing the largest part of his body has become fixed in one rigid position. The following specific characteristics can be observed with this patient:

- the cervical spine is completely stiff, which causes difficulties with eye-hand co-ordination;
- the thoracic and lumbar spine are also in complete ankylosis with a slide anteflexion and left latroflexion. This position produces a tendency to fall sidewards. He uses a stick to prevent this;
- the legs are ankylosed in all joints. There is a slight flexion position of the left knee and hip, and less of the right knee and hip;
- the right arm is the most functional. There is a limited function, about 20 degrees in all directions of the shoulder, and a good function of the elbow, wrist and hand. He is able to eat, hold a book and handle the television by remote control with the use of his right hand.

In the left arm there is a good function of the wrist and hand. However, there is a severe limitation of the elbow and an ankylosis of the shoulder. Therefore it is impossible to use his left hand for anything else but holding his stick while standing.

There is a severe restriction of the jaws, resulting in eating problems and problems with dental care.

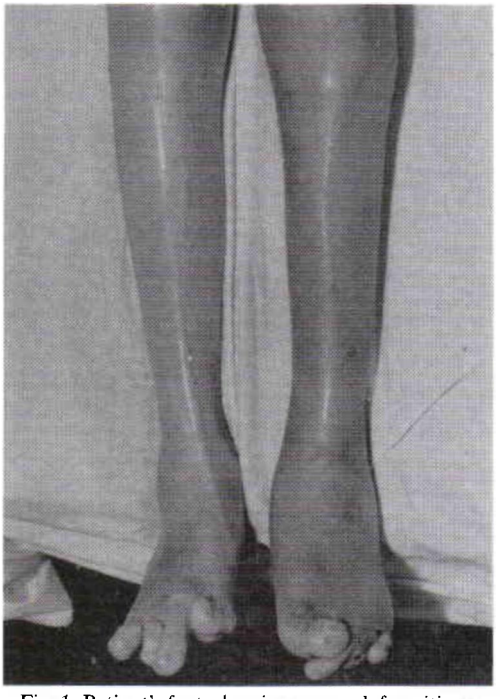


Fig. 1. Patient's feet, showing severe deformities.

The thoracic expansion is totally inhibited. Sufficient respiration is still possible, because his abdominal wall is not involved. He has not had pneumonia until now.

When he first presented, his major problems were recurrring pressure sores of both feet over a 14 year period, and the left thigh for a period of one year. After having provided him with a special anti-decubitus mattress (a ROHO mattress), the problems of his thigh were solved. Because of the severe deformity of his feet (Figs. 1 and 2) rehabilitation shoes were provided to achieve an optimal distribution of bodyweight over the whole of both foot soles and to give better balance while standing. With optimal fitting, however, the feet remained a problem. It appeared that his feet could not bear his bodyweight any more during the whole day. Because his only other alternative was lying in bed and because of his wish to move without help, a standing device was designed. In this standing device he should be able to drive his wheelchair and change position in order to relieve his feet.

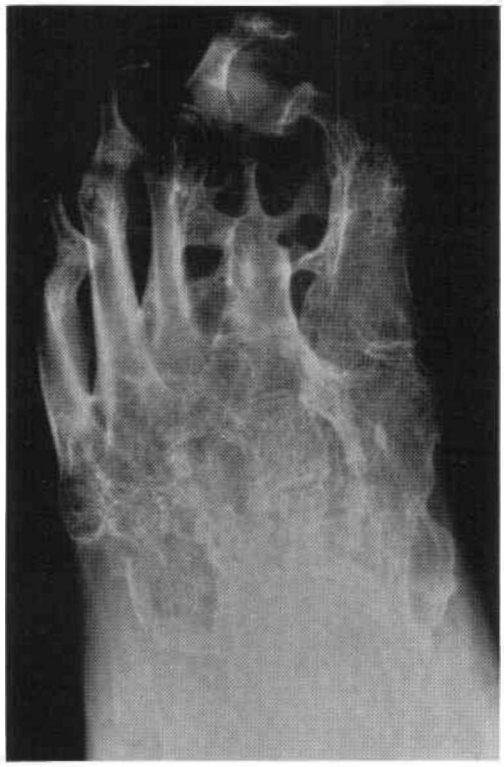


Fig. 2. X-ray left foot at the age of 40, showing severe ossification and osteolysis based on recurring ulcerations.

Design

Before starting to design the device a problem analysis was carried out, containing the following components:

the patient's physical level of function, as described above;
the medical goal of the device;
the user's wishes concerning the device;
specific safety comments.

Medical goals

To relieve the pressure on the feet, it is necessary to make it possible to change position from lying to standing in an easy way. To prevent decubitus ulcers elsewhere on his body, the body contours should be accurately followed. An optimal spread of pressure and a good air circulation (to prevent extra perspiration in a totally fitted device) were required.

User's wishes

The patient had the following wishes concerning the device:

— to rise from and sit down in his chair independently;

Fig. 3. Patient standing in his device. Later the position of the armrest and footboard were changed.

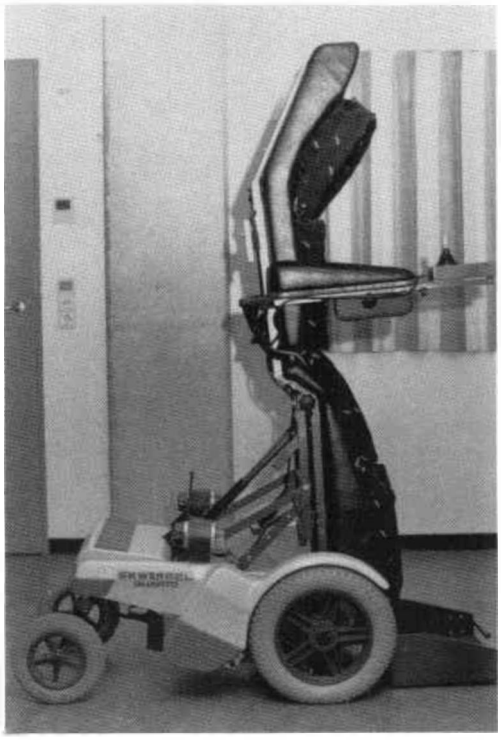


Fig. 4. Upright standing position in which he is able to leave and enter the device. Note the curves necessary to follow the patient's ankylosed posture.

— to move in and outdoors without help;
— to drive his wheelchair on slightly rough terrain, such as a garden or a camp site; this really was a great desire;
— to change from standing to lying and back in a very easy and independent way.

Specific safety comments

The patient has to drive in an almost upright standing position so that he can see where he is going. This means great stability is necessary for the undercarriage. It must be possible to use safety devices (such as belts) independently. The driver's control has to be visible as well as easy to operate with his right hand.

Practical development of the device

After completing the problem analysis, the manufacture of the device was commenced (Figs. 3—5). A children's wheelchair undercarriage (HUKA-Squirrel) was used as a base. This undercarriage has a standard feature high-low control, which was transformed into a rotating function. The undercarriage is designed to be very steady, because of the high-low function. It has a front wheel drive, which makes it suitable for use on rougher terrain. A square metal frame was fixed to the base. The frame was fitted approximately to the patient's body. Since the neck and the hips of the patient had a flexion position, two bends were made in the frame. After this, the frame was covered with foam. The top was made of a ROHO low-profile anti-decubitus mattress. As a footboard, a very thin metal carrier was used. The footboard came flat on the ground in the upright standing position. This made it possible for the patient to enter the chair without help.

After completing the building of the chair, the driver's control was placed in a good position. Safety aspects were as follows:

— the elbow-rests were extended from the trunk and were made tip-up;
— for outdoor-driving a safety belt was added.

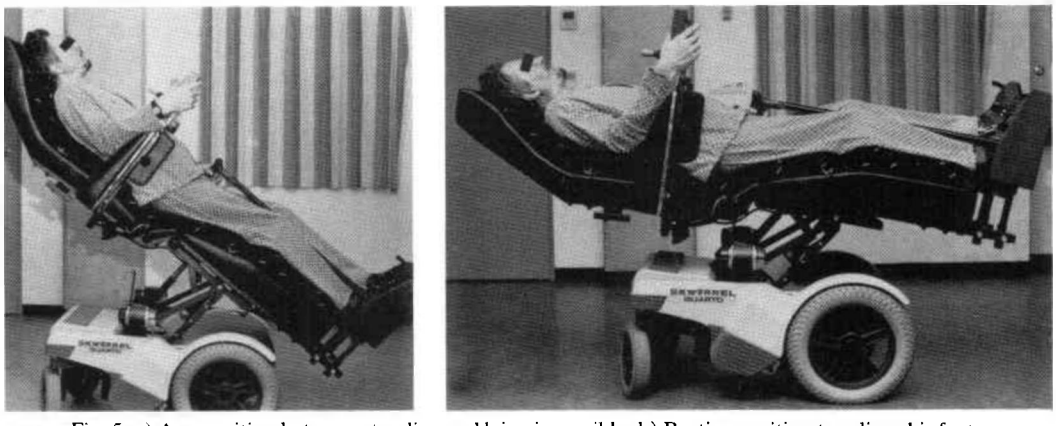


Fig. 5. a) Any position between standing and lying is possible. b) Resting position to relieve his feet.

When standing the left leg is standing more in anteflexion than the right leg. Therefore, the left foot was extended somewhat over the ridge of the footboard. To avoid stubbing his toe, a tip-up frame for the left foot was made. This frame could be handled independently by the patient with the help of his stick.

Evaluation of the user

The chair is regarded by the patient as very comfortable. His action radius has improved enormously, and he has the opportunity to go out on his own. Since he has had his chair no new decubitus ulcers have occurred, and the ones he had have healed.

REFERENCES

Connor JM, Evans DAP (1982). Fibrodysplasia ossificans progressiva: the clinical features and natural history of 34 patients. *J Bone Joint Surg* **64A**, 76–83.

Heutzer B, Jacobsen HH, Asboe-Hansen G (1977). Fibrodysplasia ossificans progressiva. *Scand J Rheumatol* **6**, 161–171.

Rogers JG, Blair Geho W (1979). Fibrodysplasia ossificans progressiva: a summary of 42 cases. *J Bone Joint Surg* **61A**, 909–914.

Prosthetics and Orthotics International, 1992, 16, 69–72

VII World Congress, Update

28 June — 3 July 1992, Chicago, Illinois, USA

Dear Colleague,

The main objective of any scientific program is the exchange of information, The Seventh World Congress of ISPO should reach that objective. A comprehensive scientific program has been organized with many aspects of prosthetics/orthotics, rehabilitation, and rehabilitation engineering to be presented by experts from around the world. The Congress organizers have received over 450 abstracts from 33 different countries. Topics range from creative technology using local materials in developing countries to high-tech imaging and automated manufacturing technology. The state-of-the-art in rehabilitation and prosthetic/orthotic practice will be presented, discussed and demonstrated. All disciplines involved in prosthetics and orthotics rehabilitation, including prosthesis and orthosis users, will participate and present at this congress.

Thirty-seven traditional instructional programs are scheduled, and for the first time, nineteen manufacturer-sponsored instructional programs are also being offered. Your new registration form has those manufacturers and course titles listed (see later in this update).

Twenty overview (plenary) sessions have been organized. These sessions are designed to bring out discussion and to clarify opinions concerning current practice. Thirty-minute papers will be presented, followed by a ten-minute response by a selected discussant. This discussant may provide additional information or may present an opposing point of view. Audience participation is encouraged, and this will prove to be an enlightening forum for discussion.

Afternoons are dedicated to symposia and other scientific sessions. Approximately thirty symposia have been pre-organized around research and development themes. In addition, approximately thirty more scientific/clinical/engineering sessions (regular and poster) will be offered based upon contributed papers (free papers). The symposia and contributed-paper sessions will be 90 minutes in length and include six presentations, of fifteen minutes each. They will cover a wide variety of topics related to prosthetics/orthotics, rehabilitation, and rehabilitation engineering. A video/film program is also scheduled.

Further information with regard the program can be found in the December 1991 issue of Prosthetics and Orthotics International.

The Exhibition will be an educational event. There are over 90 exhibitors and nearly 200 booths that will be filled with the latest technology, components, and scientific information.

The planned social events and the City of Chicago will permit you to experience the hospitality and warmth of America. We are all looking forward to seeing you in Chicago.

NB Deadline for early registration has been extended to 27 April 1992.

Mel Stills, C.O.
Chairman
International Congress Committee

HOTEL RESERVATION FORM

Seventh World Congress of ISPO, June 28 to July 3, 1992
Hyatt Regency Chicago Hotel, Chicago, Illinois, USA
Deadline for *Receipt* of Early Registrations: June 3, 1992
After June 3: Reservations accepted as space available.

Personal Information

Name (Family Name, Given Name, Initial)

Title or Position

Company or Organisation

Mailing Address

Telephone/Fax

Accompanying Person(s) or other Attendees in room with you *(limit 2 persons per room):*

Registration Category *tick one*

☐ *Member* ☐ *Non-Member* ☐ *Student* ☐ *Exhibitor*

Would you like a room for persons with disabilities? Tick this box ☐ *(Special rooms are limited. Please request early.)*
Do you need other assistance because of a disability? Tick this box ☐ and describe needed assistance below:

Room Stay Information

Arrival Date: _____ Estimated Time of Arrival: _____ am pm *(circle)*

Departure Date: _____ *(Checkout is at 11.00 hours)*

Tick type of room desired:
☐ SINGLE $102 US plus tax per night
☐ DOUBLE $102 US plus tax per night
☐ TWIN $102 US plus tax per night
(All room rates are subject to state and local taxes, currently equivalent to 12.4%.)

Reservation Fees

One night's deposit is required to guarantee your hotel reservation. You may charge the fee ($102.00 US per room) to a major credit card, or you may issue a check, to be paid in US dollars, drawn on a US bank, payable to the "Hyatt Regency Chicago Hotel". Please note that bank charges will be incurred and you will be invoiced for those charges unless your check is drawn on a US bank. If you must cancel your reservation, contact the hotel directly, before 16.00 hours on the day of arrival (312–565–1234).

Please charge my credit card for $102.00 US per room to reserve my accommodations:
Credit Card Name *(VISA, Mastercard, American Express, Diners Club, Discovery Card):* _____
Expiration Date: _____
Card Number _____ Name as it appears on Card _____
Cardholder Signature _____
Send all forms and fees to: Moorevents, 676 N. St. Clair St., Suite 1765, Chicago IL 60611 USA

For Office Use Only:
Date Received ____ **Date Acknowledged** ____ **Amount Received** ____ **Amount Due** ____ **Special Needs Y_ N_**

CONGRESS REGISTRATION FORM

Seventh World Congress of ISPO, June 28 to July 3, 1992
Hyatt Regency Chicago Hotel, Chicago, Illinois, USA
Deadline for *Receipt* of Early Registrations: April 27, 1992 (Late after April 27)

Personal Information

Name (Family Name, Given Name, Initial)

Title or Position

Company or Organization

Mailing Address

Telephone/Fax

Accompanying Person(s) *name as it should appear on badge*

Registration Category *tick one*

☐ *Member* ☐ *Non-Member* ☐ *Student* ☐ *Exhibitor*

Do you need assistance because of disability? Tick this box ☐ and describe needed assistance below:

Fee Worksheet	**TOTALS**
1. Congress Registration Fee *tick all applicable and write total in column at right.*	
☐ 1a Member Early $400 ☐ 1a Member Late $500 ☐ 1a Member One Day $180	1a
☐ 1b Non-Member Early $520 ☐ 1b Non-Member Late $620 ☐ 1b Non-Member One Day $220	1b
☐ 1c Student Early $200 ☐ 1c Student Late $250 ☐ 1c Student One Day $100	1c
☐ 1d Accompanying Person Early $150 How many? __ ☐ 1d Accompanying Person Late $200 How many? __	1d
2. Instructional Courses *ticked on reverse side of this page* How many? __ × $50 each course. Write total in column at right.	2
3. Activities *tick all applicable and write totals in column at right.*	
3a Meals: ☐ Lunch-Monday $10 How many? __ ☐ Lunch-Tuesday $10 How many? __ ☐ Lunch-Thursday $10 How many? __	3a
3b Tours: ☐ Tour A $42 How many? __ ☐ Tour B $45 How many? __ ☐ Tour C $50 How many? __ ☐ Tour D $34 How many? __ ☐ Tour E $31 How many? __ ☐ Tour F $30 How many? __ ☐ Tour G $60 How many? __ ☐ Tour H $45 How many? __	3b
3c Banquet: ☐ For All Attendees and Accompanying Persons $75 each. How many? __	3c
3d Accompanying Persons Tours ☐ Tour A $45 How many? __ ☐ Tour B $30 How many? __ ☐ Tour C $42 How many? __ ☐ Tour D $30 How many? __ ☐ Tour E $42 How many? __	3d
TOTAL FEES DUE *add lines 1a, 1b, 1c, 1d, 2, 3a, 3b, 3c, 3d and write total in space at right.*	

Full payment is due at time of registration. Checks and money orders must be payable to "VII World Congress of ISPO" in US dollars drawn on a US bank. Failure to do so will result in bank charges that will be invoiced to the attendee.

Or charge to VISA or Mastercard: *tick one* _____ VISA _____ Mastercard Expiration Date: _____

Card Number _____ Name as it appears on Card _____

Cardholder Signature _____

Send all forms and fees to: Moorevents, 676 N. St. Clair St., Suite 1765, Chicago IL 60611 USA.

For Office Use Only:

Date Received ____ **Date Acknowledged** ____ **Amount Received** ____ **Amount Due** ____ **Special Needs Y**__ **N**__

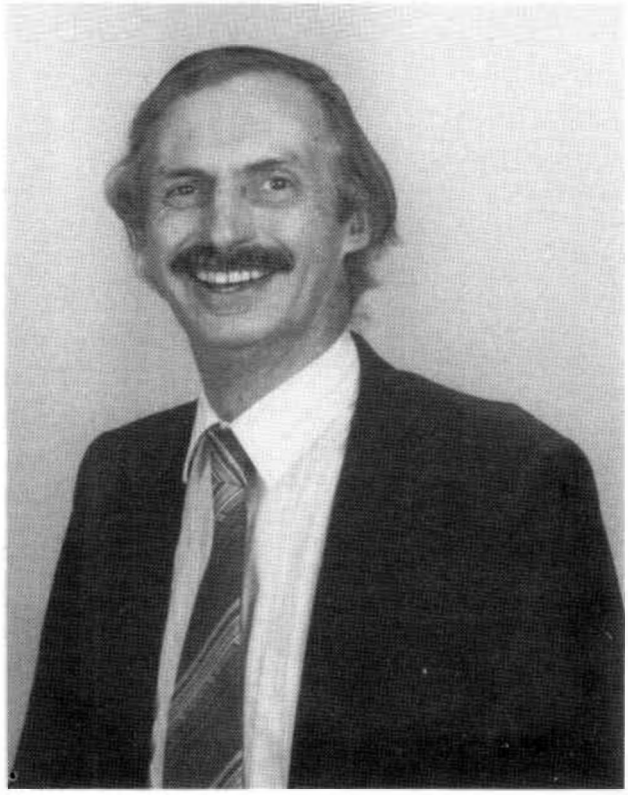
Instructional Courses and Manufacturers Workshops

(please tick boxes to enrol in course)

Sunday, June 28 (10.00–12.00 hrs)	Monday, June 29 (08.00–09.30 hrs)	Tuesday, June 30 (08.00–09.30 hrs)	Wednesday, July 1 (08.00–09.30 hrs)	Thursday, July 2 (08.00–09.30 hrs)	Friday, July 3 (08.00–09.30 hrs)
IC 101 Amputation surgery ☐	*IC 102* Amputation surgery (cont.) ☐	*IC 103* Management of the diabetic foot ☐	*IC 104* Management of the arthritic and deformed foot ☐	*IC 105* Postsurgical and early management of BK and AK amputations ☐	*IC 106* Orthotic management of bone fractures ☐
	IC 202 Early post trauma management of persons with spinal cord injury ☐	*IC 203* Surgical management of spinal fractures ☐	*IC 204* Management of the upper limb in spinal cord injury ☐	*IC 205* Mangement of brachial plexus injuries ☐	*IC 206* Orthotic and therapeutic management of stroke ☐
IC 301 Seating and positioning ☐	*IC 302* Introductory biomechanics ☐	*IC 303* Introductory biomechanics (cont.) ☐	*IC 304* Advanced biomechanics ☐	*IC 305* Biomechanics of human locomotion ☐	*IC 306* Wheelchair biomechanics and prescription ☐
IC 401 Advanced below-knee and Syme fitting practice ☐	*IC 402* Advanced below-knee and Syme fitting practice (cont.) ☐	*IC 403* Above-knee and through knee fitting practice ☐	*IC 404* Above-knee and through knee fitting (cont) ☐	*IC 405* CAD/CAM in a small clinical practice ☐	*IC 406* Hip disarticulation and hemipelvectomy fitting ☐
	IC 502 Survey of modern gait analysis equipment ☐	*IC 503* Clinical applications of gait analysis ☐	*IC 504* Surgical management of cerebral palsy ☐	*IC 505* Gait training for the leg amputee ☐	*IC 506* Gait training for the leg amputee (cont.) ☐
IC 601 Upper limb prosthetics ☐	*IC 602* Sockets for upper limb prostheses ☐	*IC 603* Myoelectric training of children ☐	*IC 604* Hand surgery and prosthetic fitting ☐	*IC 605* High level upper limb fittings ☐	*IC 606* Upper limb amputee training ☐
IC 701 Upper limb orthotics ☐	*IC 702* Lower limb orthotics ☐	*IC 703* Lower limb orthotics (cont.) ☐	*IC 704* Management of the geriatric amputee ☐		
	MC 102 Flex-Foot Inc., Dynamic lower limb prosthetic system ☐	*MC 103* College Park Industries, Practical applications of the College Park foot/ankle system ☐	*MC 104* US Manufacturing Co., New lower limb extremity prosthetic components and their applications ☐	*MC 105* Orthomedics, Semi-custom prosthetic/orthotic opportunities ☐	
	MC 202 American Plastics Inc., Thermoplastics and thermoforming techniques and standards applied to the orthotic and prosthetic field ☐	*MC 203* Durr-Fillauer Medical, Recent technological developments: the endoskeletal structural system and horizontal cable RGO ☐	*MC 204* Motion Control, Utah procontrol system and new alternatives to myoelectric control ☐	*MC 205* Becker Orthopedic, Synergetic approach to lower extremity orthotic fabrication using new technologies ☐	
	MC 302 Ohio Willow Wood Co., Carbon Copy System III BK endoskeletal system ☐	*MC 303* Pin Dot Products, Development of CAD/CAM for pressure relief cushions ☐	*MC 304* Physical Support Systems, Boston soft body jacket ☐	*MC 305* Apex Foot Health Industries, Conservative orthotic management of diabetic foot disorders ☐	
	MC 402 Southern Prosthetic Supply, CAD/CAM by Southern Prosthetics ☐	*MC 403* Otto Bock USA, Technical update to prosthetic and orthotic systems ☐	*MC 404* P.W. Minor and Son, Advances in footwear/footcare: the role of extra depth footwear in prosthetics and orthotics ☐	*MC 405* Chas. A. Blatchford & Son Ltd., Endolite phase II ☐	
	MC 502 Maramed, Miami fracture orthosis seminar ☐	*MC 503* AFI/Finnieston Clinic, The new "Care-CAM" orthotic and prosthetic design and modeling system ☐	*MC 504* Friddle's Orthopaedic Appl., Application procedure of 3D orthopaedic halo system ☐		

(Continental breakfast will be provided 07.00 – 07.30 Monday – Friday).

Editor's Retirement



Happy Retiral, Ron!

Ron Donovan was, from its inception until the end of last year, one of the Editors of this Journal. He was also co-ordinator of Teaching Services in the National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde where the Journal is produced. Ron took early retiral at the end of last year and so, sadly his experience and his energy are no longer available to us.

Ron, of course, is a well-recognised figure to many of our members because his contribution to ISPO was wide-ranging and goes back over many years. He provided audio-visual management for all but one of our World Congresses starting with Montreux in 1974 and did the same for most of the United Kingdom National Member Society's annual meetings. Ron started his career in Dundee and many members will also remember him from the famous Dundee Conferences organised by George Murdoch, which predate even ISPO.

Ron always brought a quiet professionalism and unfailing humour to his activities. His contribution to this Society has been enormous. His fellow Editors felt sure that members would wish to join in wishing Ron a long and happy retirement along with his wife Isobel and his family.

Calendar of Events

6–8 May, 1992
13th Annual International Conference of the Young Adult Institute, New York, USA.
Information: Mr. Ben Niven, Conference Director, Young Adult Institute, 460 West 34th St., New York, NY 10001-2382, USA.

7–10 May, 1992
Annual Meeting of the American Board of Physical Medicine and Rehabilitation, Rochester, MN, USA.
Information: American Board of Physical Medicine and Rehabilitation, Suite 674, Northwest Center, 21 First Street SW, Rochester, MN 55902, USA.

15–17 May, 1992
Annual Meeting of the American Spinal Injury Association, Toronto, Canada.
Information: American Spinal Injury Association, 2020 Peachtree Road NW, Atlanta, GA 30309, USA.

21–24 May, 1992
Meeting of the International Society for the Study of the Lumbar Spine, Chicago, USA.
Information: Prof. Alf Nachemson, Dept. of Orthopaedics, Sahlgren Hospital, S-413 45, Goteborg, Sweden.

25–30 May, 1992
2nd International Congress of the International Federation of Societies of Hand Therapists, Paris, France.
Information: Covergences, IFSHT '92, 120 Gambetta Ave., F-75020, Paris, France.

3–5 June, 1992
46th Congress of the Scandinavian Orthopedic Association, Malmo, Sweden.
Information: Congress Bureau, ICM AB, Geijersgaten 50, S-216 19 Malmo, Sweden.

6–11 June, 1992
15th Annual Conference of RESNA Rehabilitation Technology, Toronto, Canada.
Information: RESNA, Suite 700, 1101 Connecticut Ave. NW, Washington, DC 20036, USA.

14–18 June, 1992
Annual Conference of the American Physical Therapy Association, Denver, USA.
Information: APTA, 1111 N. Fairfax St., Alexandria, VA, USA.

21-22 June, 1992
European Spinal Deformities Societies Meeting, Lyon, France.
Information: Eric Bancilhon, 29 Rue President Ed. Herriot, 69002 Lyon, France.

21–24 June, 1992
8th Meeting of the European Society of Biomechanics, Rome, Italy.
Information: ESB '92, Istituto di Fisiologia Umana, Universita "La Sapienza" Piazzale Aldo Moro, 5, 00815 Rome, Italy.

21–26 June, 1992
9th Combined Meeting of the Orthopaedic Associations of the English Speaking World, Toronto, Canada.
Information: Dr. J. Kostuik, Organising Chairman, 9th Combined Meeting of the Orthopaedic Associations of the English Speaking World, Suite 350, 55 York St., Toronto, Canada, M5J 1R7.

24–26 June, 1992
2nd International Congress of Movement Disorders, Munich, Germany.
Information: Secretariat ISMD, PO Box CH-4005, Basel, Switzerland.

25–28 June, 1992
3rd Common Meeting of the Cervical Spine Research Society, Athens, Greece.
Information: Scientific Committee, 3rd Common Meeting of CSRS, c/o Prof. D.S. Korres, 10 Heyden St., 104 34 Athens, Greece.

28 June–2 July, 1992
9th International Congress of International Society of Electrophysiological Kinesiology, Florence, Italy.
Information: CESPRI, Fondazione Pro Juventute, Don Carlo Gnocchi, Via Imprunetana 124-50020 Monte Oriolo, Florence, Italy.

28 June–3 July, 1992
7th World Congress of ISPO, Chicago, USA.
Information: 7th World Congress of ISPO, Moorevents Inc., 400 North Michigan Avenue, Suite 2300, Chicago, IL 60611, USA.

29–30 June, 1992
Annual Meeting of the National Association of Rehabilitation Facilities, Chicago, USA.
Information: National Association of Rehabilitation Facilities, PO Drawer 17675, Washington, DC 20041, USA.

5–10 July, 1992
6th Mediterranean Conference on Medical and Biological Engineering, Italy.
Information: Prof. M. Bracale, Sec. General Medicon '92, Departmento di Ingegneria Elettronica, Via Claudio, 21, 80125 Napoli, Italy.

6–9 July, 1992
8th International Conference on Mechanics in Medicine and Biology, London, England.
Information: Dr. R.I. Kitney, Dept. of Electrical Engineering, Imperial College of Science and Technology, Exhibition Rd., London SW7 2BT, England.

16–19 July, 1992
Summer Meeting of the American Orthopaedic Foot and Ankle Society, California, USA.
Information: Arlene Napolilli, AOFAS, 222 S. Prospect Ave., Park Ridge, IL 60068, USA.

24–28 August, 1992
2nd North American Congress on Biomechanics, Chicago, USA.
Information: Dr. L. Draganich, Dept. of Surgery, University of Chicago, 5841 South Maryland Ave., Box 421, Chicago, IL 60637, USA.

26–29 August, 1992
10th Congress of the Hungarian Orthopaedic Association, Budapest, Hungary.
Information: Prof. T. Vizkelety, Ortopediai Klinika, H-1113, Budapest, Karolina ut 27, Hungary.

4–5 September, 1992
International Conference on Experimental Mechanics, Technology Transfer between High Tech Engineering and Biomechanics, Limerick, Ireland.
Information: Conference Secretariat (BSSM 92), Dept. of Mechanical and Production Engineering, University of Limerick, Plassey Technological Park, Limerick, Ireland.

7–11 September, 1992
17th World Congress of Rehabilitation International, Nairobi, Kenya.
Information: The Association for the Physically Disabled of Kenya Headquarters, Lagos Rd., PO Box 46747, Nairobi, Kenya.

8–10 September, 1992
Paraplegia '92, International Forum on Paraplegia, Barcelona, Spain.
Information: BRP, Barcelona Relaciones Publicas, C/Pau Claris, nº 138, 7º -4ª, (Edifici Laietana), E-08009 Barcelona, Spain.

8–11 September, 1992
Autumn Scientific Meeting of the British Orthopaedic Association, London, England.
Information: BOA, 35-43 Lincoln's Inn Fields, London WC2A 3PN, England.

14–18 September, 1992
Update Course in Amputation Surgery and Related Prosthetics, Moshi, Tanzania.
Information: N.A. Jacobs, National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, 131 St. James Road, Glasgow G4 0LS, Scotland.

14–18 September, 1992
XI World Congress of The International Federation of Physical Medicine and Rehabilitation, Dresden, Germany.
Information: Prof. Jurgen Kleditsch, Dept. of Physical Therapy and Research, Clinic of Orthopaedics, Medical Academy "Carl Gustav Carus" Dresden, Germany.

17–18 September, 1992
The Science and Technology of Orthopaedic Implants, Glasgow, Scotland.
Information: Prof. J.C. Barbenel, Bioengineering Unit, University of Strathclyde, 106 Rottenrow, Glasgow G4 0NW, Scotland.

27–30 September, 1992
2nd Conference of the European Orthopaedic Research Association, Varese, Italy.
Information: Prof. Ugo E. Pazzaglia, Clinica Ortopedica, 2 Facolta di Medicina e Churugia dell'Universita di Pavia, Ospedale F. Del Ponte, Via F. Del Ponte, 19, I-21100 Varese, Italy.

26–31 October, 1992
American Orthotic and Prosthetic Association Annual National Assembly, Orlando, USA.
Information: AOPA, 717 Pendleton St., Alexandria, VA 22314, USA.

29 October–1st November, 1992
14th Annual Conference of the IEEE EMBS, Paris, France.
Information: J.L. Coatrieux, Lab Traitement du Signal, Universite de Rennes 1, Campus de Beaulieu, Rennes Cedex, France.

8–13 November, 1992
Annual Meeting of the American Academy of Physical Medicine and Rehabilitation, San Francisco, USA.
Information: AAPMR, 122 South Michigan Ave., Suite 1300, Chicago, IL 60603, USA.

21–22 November, 1992
7th National Meeting of the Pakistan Orthopaedic Association, Lahore, Pakistan.
Information: Prof. Naseer Mahmood Akhtar, ORTH-CON-92, Conference Secretariat, PO Box 760, Lahore, Pakistan.

2–4 December, 1992
7th International Conference on Biomedical Engineering, Singapore.
Information: The Secretary, 7th ICBME, 1992, Dept. of Orthopaedic Surgery, National University Hospital, 5 Lower Kent Ridge Rd., Singapore 0511, Republic of Singapore.

8–13 December, 1992
16th Annual Convention of the American Academy of Neurological and Orthopaedic Surgery, Las Vegas, USA.
Information: Dr. Michael M. Rask, 2320 Rancho Drive, Suite 108, Las Vegas, Nevada 89102-3592, USA.

1993

18-23 February, 1993
Annual Meeting of the American Academy of Orthopaedic Surgeons, San Francisco, USA.
Information: AAOS, 222 South Prospect, Park Ridge, IL 60068, USA.

28 March–1 April, 1993
Biomedical Engineering Society, New Orleans, USA.
Information: BMES, PO Box 2399, Culver City, CA 90231, USA.

30 March-4 April, 1993
American Academy of Orthotists and Prosthetists Annual Meeting and Scientific Symposium, Las Vegas, USA.
Information: AAOP, 717 Pendleton St., Alexandria, VA 22314, USA.

9-11 April, 1993
66th Annual Meeting of the Japanese Orthopaedic Association, Kobe City, Japan.
Information: Dr. K. Ono, President, Dept. of Orthopaedic Surgery, Osaka University Medical School, Fukishimu 1-1-50, Fukishimu-ku, Osaka 553, Japan.

21–23 April, 1993
1st European Congress of Orthopaedics, Paris, France.
Information: Convergences-CECO 93, 120 Gambetta Ave., F 75020 Paris, France.

26–28 May, 1993
2nd European Conference on the Advancement of Rehabilitation Technology, Stockholm, Sweden.
Information: Ms. Catarina Brun, ECART 2, Swedish Handicap Institute, Box 510, S-162 15 Vallingby, Sweden.

10–12 June, 1993
7th Congress of the European Society for Shoulder and Elbow Surgery, Aarhus, Denmark.
Information: Orthopaedic Hospital, Randersvej 1, DK-8200 Aarhus N. Denmark.

24–26 August, 1993
SICOT Pre-Congress and 15th Annual Meeting of the Thai Orthopaedic Association, Bangkok, Thailand.
Information: Chusakdi Suwansirikul, M.D., Thai Orthopaedic Association, 94/2 Supavdee Tower, 100, Soi Mitanant, Nakornchaisri Rd., Dusit, Bangkok 10300, Thailand.

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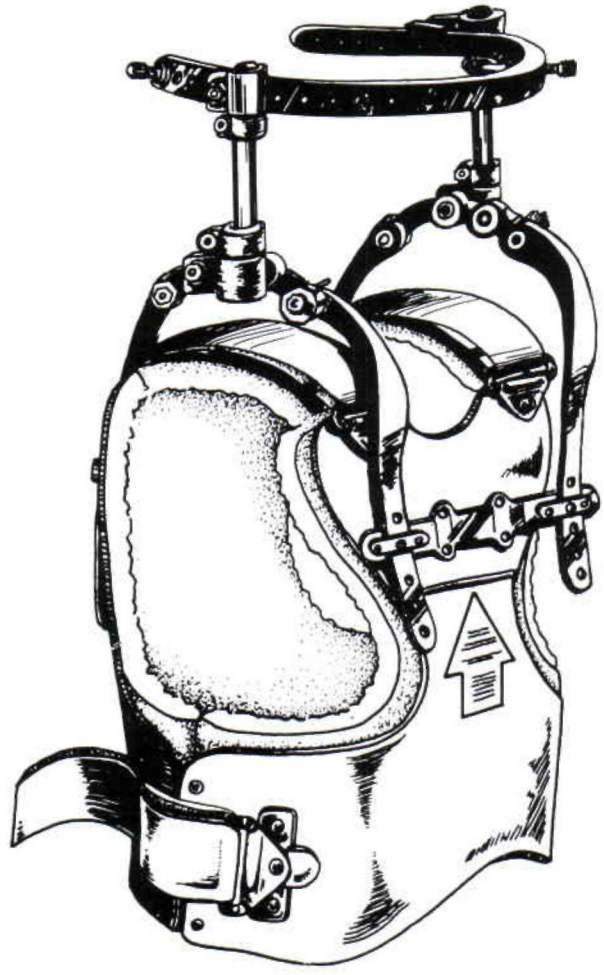

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