The Journal of the International Society for Prosthetics and Orthotics

# Prosthetics and Orthotics International

August 1995, Vol. 19, No. 2

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# Prosthetics and Orthotics International



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# The Journal of the International Society for Prosthetics and Orthotics

## August 1995, Vol. 19, No. 2

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# ISPO

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It is a great honour for me to have the opportunity to serve as President of the International Society for Prosthetics and Orthotics. On behalf of the newly elected ISPO Executive Board I accept this responsibility and its attendant tasks which we will carry towards the purposes and goals of the Society in the coming three years. It is my great pleasure to announce that our membership is continuing to grow, extending to 75 countries and 28 National Member Societies.

With the superb leadership of President Melvin Stills over the past three years and the cooperation of the Board members, Task Officers and National Member Societies we have organized many consensus conferences and courses. We look forward to carrying on the same kind of activities in the next triennium. Happily the newly elected Board members are a group of experts from all branches of prosthetics, orthotics and rehabilitation engineering and I am sure they will form an excellent multidisciplinary team well able to meet everybody's needs.

The Society's goals and functions over the coming triennium will be determined by the newly elected Board and the International Committee and Task Officers will be appointed for each task to apportion responsibility. In my personal opinion the following three points are of utmost importance to our Society.

The first is education. Our goal in this area is the improvement in the quality of personnel from all the professions regarding prosthetics, orthotics and rehabilitation engineering, under the excellent leadership of John Hughes, chairman of the Education Committee. In the industrial world the certification and accreditation of prosthetists and orthotists is a major task, particularly in the clarification of criteria for Category I and II personnel, specifying job description, learning objectives and content and format of examinations.

Another great task before us is the improvement of the level of education of prosthetic and orthotic technologists in developing countries and for this the Society needs a positive plan. In two months the Society will hold a consensus conference on "Appropriate Prosthetic Technology for Developing Countries" in Cambodia with the support of USAID under leadership of Melvin Stills, and it is hoped that much will be achieved from this event.

The second important task for the Society is for National Member Societies and the Executive Board to mutually strive for closer collaboration. At the same time I think it is important for us to establish key people in countries which do not have a National Member Society and strengthen networks between these people. Fortunately the goals and functions of international consultants are now clear, so with the cooperation of the National Member Societies and the key persons they should hold regional courses and conferences to provide information to as many people as possible so that people from all over the world can participate in the Society.

The third challenge is to advance cooperation with the many international organizations which are concerned with prosthetics and orthotics, such as INTERBOR, WHO, UN, WOC, ICRC, RI and others.

Now we are coming to the end of a very successful congress in Melbourne, Australia under the leadership of Secretary-General Valma Angliss. A tremendous amount of information has been shared. We have gained the latest knowledge, renewed old friendships and started new ones. I would especially like to express my thanks to Valma Angliss and her colleagues for their tremendous efforts and hard work in organizing this Congress.

Steady progress is being made in the arrangements for the next World Congress, which is to be held in Amsterdam, Netherlands in 1998. The preparations are in the capable hands of Secretary-General

Hans Arendzen who is our most reliable friend and our Past President Willem Eisma with his skillful chairmanship, so I am doubly sure the Congress will be a great success. Let us support the Congress in Amsterdam in 1998.

I would also like to thank our president Melvin Stills for his hard work and enormous contribution to the Society. In the last three years we have worked very closely together. I feel his enthusiasm and sincerity for the Society and his superb executive skills have made the Society what it is today.

The new board meets shortly to begin planning the activities of the Society during this triennium. I hope you will all pass to the society the benefits of your experience, your keen opinions and the fruits of your research. I would like to see the Board and the International Committee's activities to work towards the goal of normalization of people with disabilities.

I look forward to meeting you again in Amsterdam in 1998.

Seishi Sawammura
*President*

# Official Opening Address Eighth World Congress Melbourne, Australia

It is indeed a great honour to be a Patron of this very important organisation for the work of ISPO, which began some 25 years ago, provides outstanding care and opportunities for injured peoples throughout the world. I, myself, was injured in war in Korea 43 years ago and as a young man had no knowledge at all at that time of what lay ahead of me. The very word prosthetic was foreign to me as, of course, was the word orthotic, but I was soon to become familiar with the concept of getting up again having been knocked down.

Your founder, Dr. Knud Jansen, who will be recognised later today in a lecture named after him, had the foresight to see the need for the application of science and technology to assist humans who were injured. Of course, the processes of providing prostheses and orthoses has been in existence since the early times and we are all aware of the peg-leg worn by the pirate chiefs, the various hooks and devices used to simulate a hand, but it is only in recent times that refinements due to science and dedicated people has allowed us to provide items that allow for near perfect function in many cases.

With this, ISPO has developed unequalled expertise in its specialised areas with particular emphasis on education, training in the many specialities that had developed in the two fundamental disciplines and perhaps, most important of all, the wherewithal for quality research, development and evaluation in this ever changing field of medicine. ISPO is a non-governmental organisation, and can justifiably be proud of the good will and true advice it provides to countries and peoples of the world.

I recall after being injured in Korea being evacuated through the American MASH – yes the very MASH that was made famous in the TV series – and I can assure you that the TV series was quite accurate in the description of the equipment and procedures that were used. I owe my life to its skills, but I suppose I have one regret, and that is that I didn't see Hot Lips – she must have been away for the day. It was then that I learned about orthotics in the form of plaster casts and splints and so began an evacuation home through a variety of medical installations comforted by quality care and my broken limbs being properly immobilised. The life of a wounded soldier is different from civilian casualties in that he is evacuated through a series of medical units in which each provides care including surgery, if necessary, to reach home in a good condition for the next phase of care. It was then that I learned of rehabilitation – another new term to me – and I would like to point out how very important this word is to society, and, in particular, its precise meaning: I quote from the Collins dictionary, which says – "rehabilitation: specialised treatment of wounded to restore the patient to full normality".

The meaning of this word has changed over the years realising that return to full normality is not always possible, but I would like to make the point that to return to full normality is still the aim, in my view. I think this is sometimes forgotten in our world of compensation, support programmes, counselling and the like. The best advice I received after being injured and returning to Australia was that in the case of an amputation, in the words of Shakespeare (Macbeth), "What's done cannot be undone", therefore, recognise what you are. This leads on to the philosophy I always recommend to injured people, which is: "It is not what you can't do that matters, it is what you can do that matters". In simple terms, maximise what you are good at, exploit it and the rewards will be there for you in life.

Ladies and Gentlemen, I would like to make reference to a famous Australian specialist in rehabilitative medicine, Dr. Robert Klein, MBE, who founded ISPO in Australia. Dr. Bob Klein is here today and I wish to acknowledge his outstanding work in Australia for thousands of veterans of our armed forces who have passed through the Repatriation Department Clinics over the years receiving world class orthotic and prosthetic care for injuries and ailments sustained in war. It was Dr.

Bob Klein who recognised that our services to our veterans in regard to these specialities were sadly behind world standards and set about putting us equal to the best in the world. I must say I was a beneficiary of his work and sincerely thank him personally, but also, for all others who have had the good fortune to receive the excellent care and skills from the people he has trained throughout Australia. I am not going to give an historical account of the equipment developments over the years except to say that after returning to Australia from Korea, my limbless soldier colleagues of other wars came to help me by promising to find me a job. From their experience amputees could only be employed in limited careers. After some weeks they returned to tell me they had found a job for me as a lift driver. They saw that as the ideal career path for the lower limb amputees. Such was the view in medicine at the time and also the quality of the limbs provided. That has all changed, and indeed, in my later years in the Army as Director General of Medical Services I was able to change the rules to allow amputees, both upper and lower limb, to continue to serve in the forces should they so wish.

Many continued to serve with distinction and I particularly would like to mention a double lower limb amputee from the Vietnam conflict who went on in the Army to study law and only recently retired as the Senior Legal Officer of the Australian Defence Force.

As National President of the RSL I look at many aspects of ISPO that do, and will impinge on the veteran, and indeed, the whole general community in the future. Without doubt two aspects of the future will dominate the requirements of medical care and the work of ISPO. These are:

Firstly, that we are all living longer and that the number of older people in the community requiring special medical care will increase annually for many years to come. They will need the special skills and care of the prosthetist and the orthotist. This demand for such services will grow, I believe, at a consistent rate for the next 20 years before plateauing. We must be ready for this work.

Secondly, the continuing casualty product of war. It is a fallacy to believe that we will not have continuing victims of war following the cessation of the Cold War. One has only to realise the many campaigns (often called Peace Keeping Operations) that have been going on for years and continue at this moment. The number of land mines laid in recent years in places like Cambodia, Afghanistan and Angola exceed many millions that are currently in the ground and active. Little action is being taken to remove the mines so that casualties are constantly occurring today and predictably will continue. It is a tragedy, but it is true and it means our work will continue to be needed for years to come.

I believe this week will prove to be excellent in every way and that all participants who travelled so far will enjoy their stay in Australia and, in particular, enjoy a visit to our southern capital, Melbourne. I do welcome you all most sincerely on behalf of all the organisers of this event, and would particularly like to recognise the oustanding work of Valma Angliss whom you would all know well. She is a remarkable person who has worked tirelessly for some three years to ensure the success of this great occasion. I wish to thank her on behalf of all participants for the outstanding work she has done and recognise that the benefits of this Congress to disabled peoples of the future will be a lasting tribute to her.

Ladies and Gentlemen, I have tried to outline the importance of ISPO, not only now, but into the future, and I wish to commend all involved in this great organisation. I wish you well for the week that follows and that the programme fulfils your aims and objectives. You are needed and I wish all of you and the organisation every success in your future endeavours and have great pleasure in officially opening your 8th World Congress in Melbourne on this the 2nd April 1995.

MAJOR GENERAL W. B. (DIGGER) JAMES

*Prosthetics and Orthotics International, 1995, 19, 67–69*

# Eighth Triennial Assembly 1995

# Melbourne, Australia

The Eighth Triennial Assembly of the International Society for Prosthetics and Orthotics was held on Thursday, 6th April, 1995 in Melbourne, Australia at the time of the World Congress.

The President, Melvin L. Stills, formally opened the Eighth Triennial Assembly and presented his report on ISPO activities during the years 1992-95 *(Editor's Note: The President's Report was published in the April 1995 Issue of Prosthetics and Orthotics International).* In particular the President thanked the Executive Board Members and Task Officers who had given him much support during his term as President.

The Honorary Secretary summarised the changes to the Constitution. The proposed amendments had been published in *Prosthetics and Orthotics International* and had been duly discussed and voted on by the International Committee at its meeting held prior to the Congress. The changes to the Constitution are as follows:

| Original Clause | New Clause |
|---|---|
| 2.3.2 FELLOW: the highest level of individual member who by professional achievement, integrity, reputation, and by his activities as a Member in the opinion of the Executive Board of the International Committee has contributed to a high degree to the objectives of ISPO. | 2.3.2 FELLOW: the recognition of the individual member who by professional achievement, integrity, reputation, and by activities as a Member in the opinion of the Executive Board of the International Committee has contributed to a high degree to the objectives of ISPO. |
| 2.3.3 HONORARY FELLOW: those Fellows who have been selected by the International Committee for special recognition. | 2.3.3 HONORARY FELLOW: those Members or Fellows who have been selected by the International Committee for special recognition. |
| 2.5.1 An individual may resign at any time. Members who fail to pay the specified fees will be automatically severed. | 2.5.1 An individual may resign at any time. Members or Fellows who fail to pay the specified fees will lose membership status. |
| 4.2.1 The International Committee of the ISPO will consist of Fellows selected by National Committees of National Member Societies of ISPO. | 4.2.1 The International Committee of the ISPO will consist of Members or Fellows selected by National Committees of National Member Societies of ISPO. |
| 4.2.4 The International Committee shall elect by majority vote from the Fellows at large the Officers of ISPO and others to serve on the Executive Board, all to take office at the conclusion of the Assembly. | 4.2.4 The International Committee shall elect by majority vote from the Members or Fellows at large the Officers of ISPO and others to serve on the Executive Board, all to take office at the conclusion of the Assembly. |
| 4.3.1 The Executive Board will consist of the President, the President-Elect, two Vice-Presidents, and four other Fellows of ISPO. The Honorary Secretary, the Honorary Treasurer, the Immediate Past-President and all Standing Committee Chairmen join the Executive Board as non-voting members. The President, with majority approval of the Board, may appoint non-voting consultants to the Board. | 4.3.1 The Executive Board will consist of the President, the President-Elect, two Vice-Presidents, and four other Members or Fellows of ISPO. The Honorary Secretary, the Honorary Treasurer, the Immediate Past-President and all Standing Committee Chairmen join the Executive Board as non-voting members. The President, with majority approval of the Board, may appoint non-voting consultants to the Board. |
| 4.3.5 In the event of a vacancy arising in the Executive Board during the Triennium through illness or other reason, the Executive Board may co-opt from the Fellowship at large to fill that vacancy. A Fellow co-opted in this way, where applicable, enjoys full voting rights and has the same status as those members of the Executive Board elected in the normal way. | 4.3.5 In the event of a vacancy arising in the Executive Board during the Triennium through illness or other reason, the Executive Board may co-opt from the membership at large to fill that vacancy. A Member or Fellow co-opted in this way, where applicable, enjoys full voting rights and has the same status as those members of the Executive Board elected in the normal way. |

| | |
|---|---|
| 4.5 Standing Committees, Ad Hoc Committees and Task Officers. | 4.5 Standing Committees, Ad Hoc Committees, Task Officers and International Consultants. |
| 4.5.1 There shall be a Standing Finance Committee and a Standing Protocol and Nominations Committee. | 4.5.1 There shall be a Standing Finance Committee, a Standing Protocol and Nominations Committee and a Standing Education Committee. |
| 4.5.5 The Protocol and Nominations Committee shall comprise the President, the immediate Past-President, the President-Elect, two Fellows from the membership at large, the Honorary Secretary (ex officio) and up to two Past-Presidents nominated by the President. | 4.5.5 The Protocol and Nominations Committee shall comprise the President, the immediate Past-President, the President-Elect, two Members or Fellows from the membership at large, the Honorary Secretary (ex officio) and up to two Past-Presidents nominated by the President. |
| 4.5.7 The two Fellows on the Protocol and Nominations Committee shall be appointed by the Executive Board for a three year term and may be re-appointed. | 4.5.7 The two Members or Fellows on the Protocol and Nominations Committee shall be appointed by the Executive Board for a three year term and may be re-appointed. |
| | 4.5.9 The Education Committee shall comprise a Chairman, two members, the President, the President-Elect, the Honorary Secretary (ex officio). Further members may be co-opted to carry out specific tasks. |
| | 4.5.10 The Chairman and the two members shall be appointed by the Executive Board for a three year term and may be re-appointed. |
| | 4.5.11 The Education Committee shall:<br><br>4.5.11.1 have a continuing responsibility to review the educational policy of the Society and make recommendations to the Executive Board.<br><br>4.5.11.2 oversee the educational activities of the Society as directed by the Executive Board. |
| | 4.5.13 The President, with the approval of the Executive Board, may appoint International Consultants with regard to identified tasks in relation to specific countries or geographical regions. |
| 4.6.3 A National Member Society consisting of five Members and Fellows is entitled to apply to the Executive Board for representation on the International Committee. The representative(s) must have Fellow rank or meet the requirements of Fellowship. Any membership in excess of 15 Members and Fellows will entitle the National Member Society to appoint two representatives. In no case will any member society be entitled to more than two representatives. To ensure proper representation at any given meeting, the National Committee may appoint alternatives acceptable to the International Committee. Where two representatives are involved they shall be different professional disciplines. | 4.6.3 A National Member Society consisting of five Members and Fellows is entitled to apply to the Executive Board for representation on the International Committee. Any membership in excess of 15 Members and Fellows will entitle the National Member Society to appoint two representatives. In no case will any member be entitled to more than two representatives. To ensure proper representation at any given meeting, the National Committee may appoint alternatives acceptable to the International Committee. Where two representatives are involved they shall be different professional disciplines. |

The President thanked the members of the Executive Board who were rotating off the Board: Willem H. Eisma, Per Christiansen, Jean Vaucher, Hans Arendzen and Thamrongrat Keokarn. He thanked Valma Angliss for all the efforts that she and her committee had made in making the Eighth World Congress such an outstanding success. The President made a presentation to the Honorary Secretary who had served the Society for 12 years in that office. The President also made a presentation, together with some friendly words of advice to the incoming President, Seishi Sawamura.

The Honorary Secretary announced the new Executive Board which had been elected by the International Committee prior to the Assembly. It is as follows:

| | | |
|---|---|---|
| President: | Seishi Sawamura | Japan |
| President-Elect: | Norman A. Jacobs | UK |
| Vice-Presidents: | David N. Condie | UK |
| | Harold G. Shangali | Tanzania |
| Members: | Gerhard Fitzlaff | Germany |
| | Jean Halcrow | Australia |
| | Björn M. Persson | Sweden |
| | C. Michael Schuch | USA |
| Immediate Past-President: | Melvin L. Stills | USA |
| Honorary Treasurer: | J. Steen Jensen | Denmark |
| Honorary Secretary: | Brendan McHugh | UK |

The chair was then handed over to the Incoming President, Seishi Sawamura. The Incoming President addressed the Assembly (*Editor's Note: The Incoming President's Address is printed as the Editorial to this issue of* Prosthetics and Orthotics International).

Following this, the President invited general comments and subsequently formally closed the Eighth Triennial Assembly.

Norman A. Jacobs
*President-Elect*

# ISPO – ARGENTINA

The Society is pleased to announce the formation of a new National Member Society in Argentina:

Following is a list of its officers:

*President*
Daniel Enrique Suarez

*Vice-President*
Vincente Saquez

*Secretary*
Luis A. Rivero

*Treasurer:*
Roberto Perez

The above may be contacted at the following address:

ISPO – Argentina
Ramsey 2250
1428 Capital Federal
ARGENTINA

*Prosthetics and Orthotics International, 1995, 19, 70–73*

# International Committee and Executive Board Meetings

**Executive Board: 30 March 1995**
**International Committee: 31 March-1 April 1995**
**Executive Board: 8 April 1995**

## Introduction

The Executive Board met twice in Melbourne. The first meeting preceded the World Congress and was accompanied by the pre-Board meetings of the Standing Committees (Education Committee, Finance Committee, Protocol and Nominations Committee) and Ad Hoc Committees (International Congress Committee, Publicity and Publications Committee). This was the final meeting of the outgoing Executive Board. It was followed directly by the International Committee Meeting. The second meeting of the Executive Board took place on the Saturday immediately following the World Congress. This was the first meeting of the incoming Executive Board which had taken office at the World Assembly, during the World Congress.

## Finance

The accounts for the past year were presented. There was a larger than normal deficit for the year. There were two reasons for this. Firstly there was a reduction in the market value of the Danish bonds in which much of the capital of the Society is invested. The short term loss in paper value would only be of real concern if bonds were sold at this time. It was explained, by Bent Ebskov, the Chairman of the Finance Committee, that these bonds represent a very secure investment with guaranteed maturity value but are subject to fluctuation from year to year.

Secondly, Steen Jensen, the Honorary Treasurer, highlighted the fact that the Society has been particularly active in recent years in organising courses and conferences. Since these have been largely focused on the needs of the developing world they have required a considerable input of funds.

It was agreed that the Society should continue to recognise the needs of the developing world and organise such activities but it was also recognised that there would need to be increased efforts in the area of fundraising (without increasing Members fees!). This should be possible by organising some profitable events in the industrial world and by seeking collaboration with external funding agencies to help finance projects in the developing world. This is a challenge for the incoming Executive Board!

## Education

The Chairman of the Education Committee, John Hughes, reported on a number of varied activities. Courses in Amputation Surgery and related Prosthetics had been held in Thailand (March, 1994), Slovenia (September, 1994) and Panama (November, 1994). A Consensus Conference on Orthotic Management of Cerebral Palsy had been organised by David Condie, with local management by Mike Schuch, in the USA (November, 1994).

A Consensus Conference on Appropriate Prosthetic Technology was at an advanced stage of planning. The intended venue was Phnom Penh in Cambodia; the time June 1995. A substantial 80% of the funding was being provided by the United States Agency for International Development (USAID). Binks Day was Project Officer while Mel Stills handled liaison with USAID and looked after the incredibly complex travel arrangements. The organising group was chaired by John Hughes. I can now report, with the benefit of hindsight, that the Conference was an outstanding success and, in my opinion, a great credit to ISPO. Full reports on both consensus conferences will be produced during 1995.

The Education Committee had also been concerned with professional standards in prosthetics and orthotics worldwide. This had led to the concept of different categories of practitioners according to the education and training facilities available. In the industrial world, the end point of recognised courses is the Prosthetist/Orthotist (Category I). In the developing world, where it is unrealistic to expect the same facilities, the achievable grade is that of Orthopaedic Technologist (Category II). The

Education Committee had made significant strides in developing universally accepted standards for the training of Category II practitioners in terms of job description, learning objectives and examinations. It was agreed that efforts should be made, in collaboration with INTERBOR, to develop agreed standards for Category I practitioners.

Another aspect of education which had been of great interest to ISPO, and was relevant to the discussion about professional standards, was certification. In conjunction with the American Board for Certification (ABC), the ISPO/Interbor Joint Education Committee had been conducting a series of trials of the ABC examination. These had been completed in the UK and Australia. Further trials were planned for Tanzania and Germany. The aim was to establish internationally accepted standards for certification procedures.

**Membership**

Norman Jacobs had been in communication with several newly forming National Member Societies. Applications from Argentina and Slovenia were endorsed by the Executive Board. Recently, Hungary and France had also formed National Member Societies and Mexico and Venezuela had made enquiries. There are currently 28 National Member Societies.

Membership had continued to increase steadily with some perturbations around the time of the World Congress. ISPO Membership currently stood at around 2450.

**Publicity and Publications**

The Publicity and Publications Committee, chaired by Hans Arendzen, had been active in seeking new ways to promote the Society. To this end they had produced a new colourful brochure. In addition, they had designed and distributed a new promotional leaflet aimed at attracting subscribers to Prosthetics and Orthotics International.

They had also been concerned with the editing of a new video tape on trans-tibial amputation produced by Amar Jain and John Guy. It was agreed that the video would be made available, as widely as possible, at a cost of USD 35 + postage.

**Standards**

The Task Officer for Standards, David Condie explained that the working groups dealing with ISO TC168 (Prosthetics and Orthotics), ISO TC173 (Technical Systems and Aids for Disabled Persons) and CEN TC293 (Technical Systems for Disabled Persons) are making good progress and new documents are at an advanced stage of development. More detailed information can be obtained from David Condie.

**International Organisations**

ISPO continued to enjoy a working relationship with a number of international organisations. These included the International Association of Orthotists and Prosthetists (INTERBOR), the World Health Organisation (WHO), Rehabilitation International (RI), the International Commission on Technology and Access (ICTA), the African Rehabilitation Institute (ARI), the International Verband der Orthopädie Schuteckniker (IVO), World Orthopaedic Concern (WOC), the United Nations (UN), the International Committee of the Red Cross (ICRC), the World Rehabilitation Fund (WRF) and the US Agency for International Development (USAID). Some examples of collaboration with these organisations have been given above when discussing the forthcoming Consensus Conference on Appropriate Prosthetic Technology (USAID) and the establishment of standards for Category I practitioners (INTERBOR).

**Congresses**

At the first Executive Board meeting, Valma Angliss, Secretary General for the imminent World Congress in Melbourne, gave an update on the events which were about to unfold over the coming week. By the second meeting, she and her organising group were being warmly congratulated on a highly successful World Congress.

Hans Arendzen advised the Board that preparation was going well for the 1998 World Congress to be held in Amsterdam. He anticipated that the programme structure would be similar to that of previous World Congresses.

The Executive Board proposed that bids from National Member Societies, wishing to hold the 2001 World Congress, should be received by December 1995. They would then be scrutinised in January 1996 by the Executive Board which may ask for clarification of some details and a final decision will be made in July 1996. This was approved by the International Committee.

**Reports from National Member Societies**

At the International Committee Meeting, reports were presented verbally and in the form of papers by the following National Member Societies: Caribbean, Colombia, Finland, Hong Kong, Netherlands, New Zealand, Canada, USA, UK, Japan, Australia, Panama, Belgium, Hungary. Verbal reports were also given by the following: Sweden, Argentina, Slovenia, Denmark, Germany, Switzerland and Belgium.

**Executive Board Elections**

The Protocol and Nominations Committee had been directed by the International Committee to examine the election protocol with a view to increasing the input of National Member Societies in the early stages of the election process. According to the existing protocol, a "slate" of nominations was drawn up by the Executive Board and sent to all National Member Societies for additional nominations. The Executive Board was then elected from this extended slate by the International Committee. Under the new protocol, approved by the International Committee, National Member Societies will be invited to suggest candidates for election before the slate is drawn up. It was hoped that National Member Societies would seek to nominate individuals from other countries as well as from their own National Member Society.

**International Consultants**

The President-Elect, Seishi Sawamura, presented a paper on the role of International Consultants and provided a list of present International Consultants. Verbal and written reports were provided by Črt Marinček (Central and Eastern Europe) and Eiji Tazawa (South East Asia) while written reports had been supplied by Khalil Abadi (Middle East), John Craig (Central and South America), M.A.A. El-Banna (Middle East) and Jose Miguel Gomez (South America).

**Twinning**

The Netherlands National Member Society had been very active in establishing supportive links with other nations, particularly in Eastern Europe. Jan Geertzen described some of these activities. He referred, in particular, to the efforts of Ed van Laar, who whilst providing training courses in Hungary, had been instrumental in the formation of the new Hungarian National Member Society. The Netherlands was also involved in twinning activities with Poland, Croatia and Czechia.

David Condie described the UK twinning with Malawi, in partnership with the Chartered Society of Physiotherapists, which had provided for six ISPO membership subscriptions, course attendance for three individuals and shipment of equipment to Malawi.

John Michael described USA twinning activities in Latin America. It has involved the transfer of information in the form of reciprocal lectures, collaborative conferences and exchange of papers. It has stimulated the development of friendship between the different countries and has enabled the USA to provide financial assistance, such as money for scholarships.

Rosie de Saez reported that the Panama National Member Society is currently helping their neighbours, Guatemala and Costa Rica to form National Member Societies. It was clear that the twinning activities in Central and South America were having a beneficial secondary twinning effect where the countries which had received support and help from the USA were then offering similar help to other countries in the area.

**Amendments to the Constitution**

The amendments to the Constitution, which had been detailed previously in Prosthetics and

Orthotics International (December, 1993 and August, 1994), were approved by the International Committee and are reported on in this issue of *Prosthetics and Orthotics International* in the report of the Eighth World Assembly.

**Incoming Executive Board**

The Executive Board, elected for the triennium 1995-98, is outlined in the report of the Eighth World Assembly.

**In Conclusion**

This is an exciting time as the new triennium begins. Important ongoing projects and exciting new challenges await us. We look forward, under the leadership of our President Seishi Sawamura, to maintaining a high level of effort and achievement on behalf of all members of our Society and for the benefit of people, throughout the world, who have physical disabilities.

Brendan McHugh
*Honorary Secretary*

## ISPO – SLOVENIA

The Society is pleased to announce the formation of a new National Member Society in Slovenia:

Following is a list of its officers:

*President*
Dušan Kaloper, MD, MSc

*Secretary*
Tomaž Maver, Eng.

The above may be contacted at the following address:

ISPO – Slovenia
Inštitut Republike Slovenije za Rehabilitacijo
Linhartova 51
61 112 Ljubljana
p.p. 21, SLOVENIA

*Prosthetics and Orthotics International, 1995, 19, 74–91*

# Carbon fibre and fibre lamination in prosthetics and orthotics: some basic theory and practical advice for the practitioner

B. L. KLASSON

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## Introduction

The first experiments with carbon fibre (CF) in prosthetics and orthotics were probably made by Mr. Nigel Ring, Chailey Craft School and Heritage, Sussex, England, around 1966.

This was in the very early days of CF technology, just a couple of years after the introduction of the very expensive stretched high modulus fibre. It was a very promising new technology, but it turned out, that there were some very expensive lessons to be learned by the high tech industry, before the new material could be safely used in product development. The most famous of these lessons is probably the one when the first use of carbon fibre in jet engine turbine blades failed after the production had started, with disastrous economical consequences for the company.

Mr. Ring tried to make light, stiff torso sockets for upper limb amelics, and soon after Dr. David Simpson, Edinburgh, the author and maybe some others followed. The results were interesting but the costs were prohibitive and it must be confessed, that we did not use the fibres very intelligently at the time.

In 1972 Mr. Bengt Östberg at our Een & Holmgren Uppsala branch tried to reinforce aluminium braces with carbon fibre prepreg after final adjustment of the brace. The aluminium was then used as a core in the final product. The results were excellent, but the manufacturing technique, including the use of a large and heavy autoclave, was too impractical for use in prosthetic and orthotic service.

In the late 1970s Mr. Össur Kristinsson, Iceland, invited us to join him in the development of his new trans-femoral socket concept, the flexible socket. A key component in this concept is a very stiff upright, and for this he suggested the use of CF. Since then the author has maintained a very productive contact with Mr. Kristinsson, although many others have provided very important inputs to our development and to our education programmes.

We can now look back at more than a quarter of a century playing around with CF and more than a decade routinely using it in prostheses, orthoses, corsets and orthopaedic footwear, and we are far from the only ones. CF is now widely used in prosthetics and orthotics and many allied industries, pioneered by Blatchford, UK, (prosthetic components) and Proteor, France (orthotic components) have introduced CF products and applications, one of the most recent ones the very interesting Icelandic Masterstep foot. Many of these products are beautiful examples of good professionalism in

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development and high quality production, while some of them bear evidence, that the developers did not know what they were doing, or that they hoped, that the customers would not know what they are buying. "If there is CF, it must be high tech and good!" The author has seen products where a black matrix has been used to give the impression that there is carbon fibre in it.

We have given many courses to our own Een & Holmgren/LIC staff, and we have been invited to give courses all over the world on fibre mechanics, CF and lamination since 1981. This activity has been very stimulating, although it has been a surprise, that there was such a need for improved understanding of basic fibre mechanics and lamination procedures There is no doubt, that the introduction of the expensive CF has stimulated us to improve our act and do much better also with GF (glass fibre) and other less expensive fibres.

This paper briefly summarises the classroom content of these courses and emphasises mechanical aspects. It includes basic engineering analyses, experience gained by the author and others, and recommendations and (printed) information from suppliers and manufacturers such as Union Carbide, Thoray and Exel. Complicated chemical and mathematical analyses have been purposely avoided because that is just not the way to communicate with the intended target group: the advanced, interested and demanding practitioners amongst prosthetists, orthotists and orthopaedic technicians.

The designer is strongly recommended to study the subject more closely in textbooks on fibre mechanics. Most designers are used to working with isotropic materials like metals, and this is a completely different game. Fibre composites are anisotropic (different properties in different directions), and strength and stiffness are much more dependent upon the manufacturing process.

In high tech applications in space and aeronautical engineering, the performance to weight ratio is very important, and thus high costs for calculating and testing the design and refining the manufacturing procedures are accepted even if the gain may appear to be small. In mass production and in less critical applications, such as fishing rods, ski poles, sailing masts, guitar necks etc., a marginal improvement of the performance is usually not very important. Thus the costs spent for optimising the product are modest, while efforts are spent on rationalising the manufacturing. Some years ago the author approached different manufacturers asking for cost estimations for manufacturing components for prosthetic systems. It turned out, that the high tech industries, specialised in space and defence technologies, were 6-10 times more expensive at maybe 10-15% better performance compared to what the other category of companies could offer. The other companies gave us, satisfying more limited demands, 5-8 times more for the dollar, if you prefer to put it that way.

In prosthetics and orthotics we see a lot of manufacturing of individual objects, using hand lay-up, vacuum membrane moulding and other techniques where the tooling costs are low. It is the author's view that in this category of design and manufacturing extremely good results and high quality products are within reach with only a basic understanding of fibre mechanics, let us call it "guided common sense", and this is what this presentation is about.

There are several problems when introducing CF, and also glass fibre (GF) although it is much less expensive, in prosthetics and orthotics. One is that the basic rules of materials distribution are not fully appreciated. The difference between stiffness and strength is sometimes not understood (CF is not significantly stronger than GF, 0 to some 40% only, but it is about three times stiffer). Many find it difficult to understand, that the fibres cannot be permanently deformed, and consequently the shape of a fibre dominated composite cannot, with exceptions to be discussed later, be adjusted after curing if full strength is expected.

Maybe one reason for misunderstandings is the confusing concept of reinforced plastic (RP). We do not reinforce the plastic. We use the plastic as a matrix to hold the fibres in such a position, so that they can do as much as possible of the job. It would be more correct, maybe, to say "matrixed fibres", but "fibre composite" is an excellent name.

A composite material is a material consisting of more than one component. In this context it consists of fibres to provide strength and stiffness, a matrix to bond the fibres in order to utilise their properties and, sometimes, a core to distribute the material in an optional way.

Due to limited understanding of moment of inertia many professionals in prosthetics and orthotics believe, that CF can just be added to the lamination, and if the results were not sufficient, the quality of the available CF is blamed. This misunderstanding is very common, and also in medicine there is a tendency to be trapped by "materials voodoo". It is believed, that new materials will, in a miraculous way, solve old problems. The use of materials and the utilisation of materials properties are based on engineering science with a long tradition, and maybe this science should more frequently be applied properly before miracles are asked for.

One early mistake was that layers of CF were sometimes mixed in between layers of GF in the lamination. We studied the result of such a lamination during the early development of the flexible socket (now called the ISNY socket or the Scandinavian Flexible Socket) (Fig. 1). The weight bearing structure is a medial beam, subject to a bending moment. Strength is no problem, but if the beam bends (i.e. is not stiff enough), there will be problems when the lateral wall of the flexible socket collapses, and also other stresses may occur and cause problems.

A bending moment was applied (a force acting with a lever arm of 60mm) and the deflection of the beam was measured (Fig. 2). It turned out, as expected, that a properly laminated CF beam with straight, continuous fibres was 3 times stiffer than a correctly made GF structure. But this GF structure was some 5 times stiffer than the CF-GF-mix, where no efforts had been spent on straightening the fibres or orienting them in the correct direction (Fig. 3). The conclusion is that excellent results are possible if GF is correctly used while very

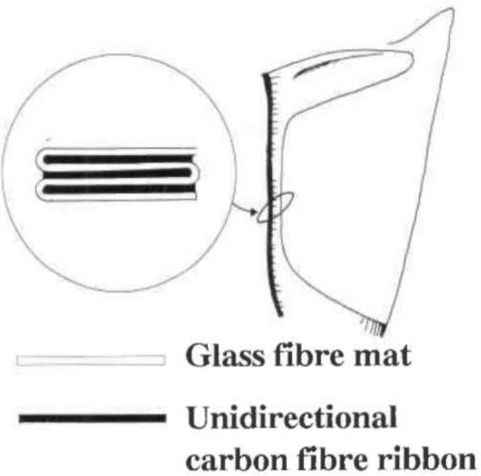


Fig. 1. Mixed layers in early flexible socket lamination.

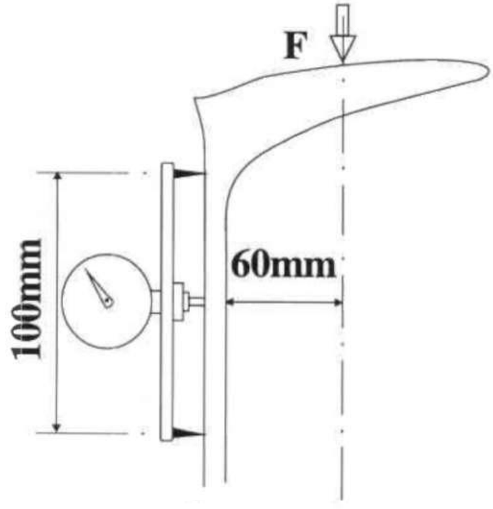


Fig. 2. Arrangement for measuring deflection.

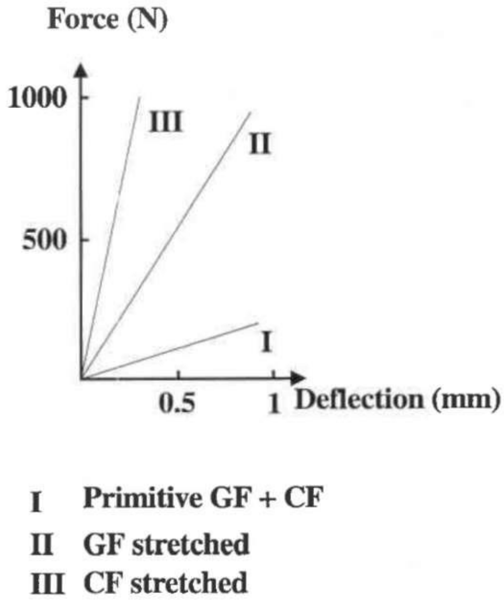


Fig. 3. Deflection of medial upright.

bad results at very high costs are likely to appear when using CF improperly. A CF fibre composite will not be much stronger, but it will be three times stiffer and some 10-20% lighter than a GF one.

At the time of this experiment, 1980-81, our annual CF consumption in Stockholm and Uppsala was about USD 200,000, most of it wasted, obviously. Our fibre mechanics and lamination education programme was one of our best investments ever!

**Stress, strain and Hooke's law**

Although mathematical calculations will be avoided as much as possible in this paper, it is necessary to agree upon some definitions and principles and share frames of reference. This is why I take the liberty of hooking up on the wall some elementary aspects of strength of materials.

The *tensile stress ($\sigma$)* is the pulling or pushing force divided by the cross-section area, or the force per unit of cross-section area (assuming that the stress is uniform across the section).

The *strain ($\varepsilon$)* is the relative deformation, i.e. the elongation or shortening divided by the length where the cross-section area is not changed.

When there is stress there is strain, or, when there is strain, there is stress. If the material returns to its original dimensions when the load is removed, there is elasticity, and if the stress is proportional to the strain, there is linear elasticity, and the material follows *Hooke's law*. Proportionality means that the stress can be calculated by multiplying the strain by a constant. This constant or stress to strain ratio is called *modulus of elasticity* or *Young's modulus (E)*. The higher the modulus of elasticity, the stiffer the material, and the less strain at a given stress (or more stress at a given strain).

It may be confusing, but the lower the modulus, the more elastic the material.

Hooke's law states that: $\boldsymbol{\sigma} = \mathrm{E} \times \varepsilon$

The material is elastic if no deformation remains when the stress is relieved. If there is remaining deformation, the stress must have exceeded the yield point into the plastic range. This is what happens when a metal rod is permanently bent (Fig. 4).

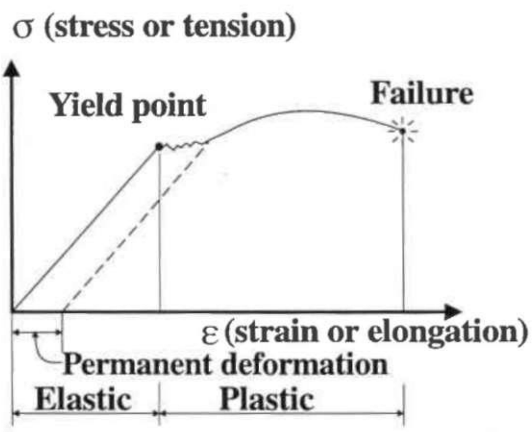


Fig. 4. Stress versus strain in steel.

The principles are basically the same for shear. There are shear stresses and shear elasticity modules, related to angular deformations. For further information see any elementary textbook on strength of materials.

**Basic principles of handling bending and torsion**

Fibre composites are frequently used in structures subject to bending. Bending is a balance or equilibrium between an external bending moment on the structure and an internal resisting moment in the structure. (A bending moment, of course, is a force combined with a lever arm).

If a beam is subject to bending (Fig. 5), it is stretched along the top face and compressed along the bottom face. Navier's principle tells us that the strains are proportional to the distance from the neutral line (NL), and so are the stresses if Young's modulus is not changed

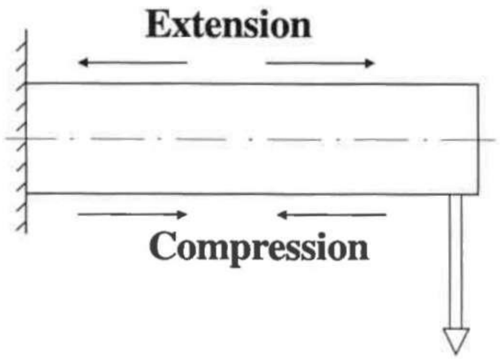


Fig. 5. Extension and compression when bending a beam.

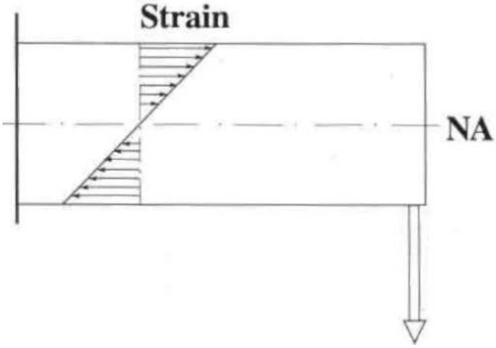


Fig. 6. Navier's principle: the strain is proportional to the distance from the neutral axis.

(Fig. 6). Hence the internal balancing contribution to the strength from each increment of the cross-section is proportional to its distance from the neutral line. The contribution to the stiffness is proportional to the square of this distance. In the neutral line, there is no compression or extension, and hence there is no point in putting strong and expensive material there, a common habit in prosthetics and orthotics. On the contrary, from the view point of costs and weight, the material should be moved as far away as possible from the neutral axis. And consequently, the further the fibres are away from the neutral axis the more effectively they are used.

The I-beam, made from steel, is a classical example of this (Fig. 7). It is available in many standardised sizes, and it has been used in bridges, buildings, cranes etc. for many, many years. The distance between the flanges is secured by a thin wall, called the web. The web also handles the shear flow (see later).

By moving away the material or the fibres from the neutral axis. the *moment of inertia* is increased without increasing the amount of material, and the moment of inertia expresses the ability to resist bending.

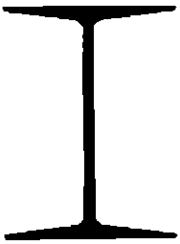

Fig. 7. The I-beam.


Fig. 8. Sandwich.

An alternative possibility to increase the moment of inertia is to build up a sandwich structure, where the thin wall is replaced by a light and/or cheap *core* (Fig 8).

The principle for moment of inertia also applies for torsion. A tube of slightly larger diameter does the same job with less material than a solid rod (Fig. 9).

Torsion is, however, not "pulling and pushing", but shear or shear flow in the cross-section perpendicular to the radius about the neutral axis. The shear flow must be *closed*, and if it has to find its way close to the neutral axis, most of the effect is lost. Thus the I-beam does not serve well against torsion. For this purpose the winner is a cylindrical tube. Also a sandwich used for torsion has to be modified, so that the face of the sandwich (with continuous shear resisting fibres. see later) closes the cross-section (Fig. 10).

**Fibre mechanics – the art of using strings**

The fibres are extremely thin, and

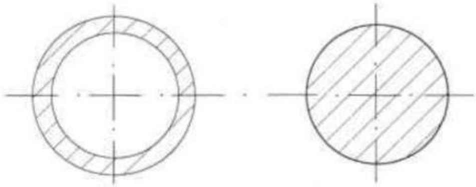

Fig. 9. Rod and tube.

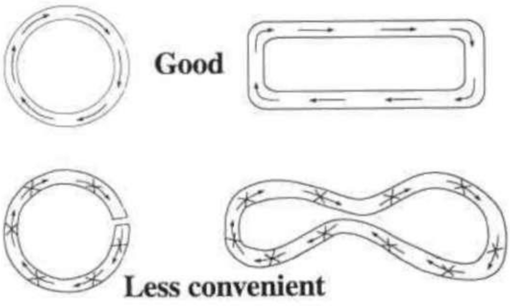


Fig. 10. Closed and open shear flow.

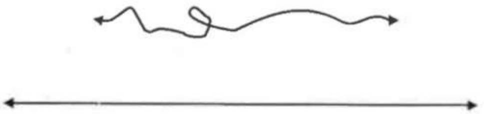

Fig. 11. Relaxed and stretched rope (or fibre).

consequently do not resist bending or twisting. However, they do resist pulling tension. They work like strings or ropes. If rope is laid out as in (Fig. 11) and pulled at both ends, it must be straightened out before it resists or takes the load. *All fibre data is related to continuous, straight fibres, parallel to the stress.*

If the rope is surrounded by some other material, like plastic, it is more difficult to straighten out. But under these circumstances the resistance is provided by the plastic material and not by the rope until of course, the rope is straight. In composite terms, the composite of rope and plastic, or fibre and matrix, is *matrix dominated* until the fibres are straight. When matrix dominated, the strength and the stiffness of the composite is determined by the matrix, and the fibres are more or less wasted.

Metals are usually *isotropic* i.e. they exhibit the same strength properties in all directions, and this is what most design engineers are trained to deal with. Fibre composites are *anisotropic*, i.e. stronger and/or stiffer in one direction than in another. This is a cultural shock for many designers, who are more or less lost when the strength properties are different in different directions. In some directions they are *fibre dominated* and strong, because the fibres take the load. In others they are *matrix dominated* and weak, because the matrix takes the load. But most of the time they are *partially fibre dominated* (Fig. 12).

The art of optimising fibre composites in a structure is the art of orienting and securing the fibres. In high tech applications this art is complicated, and there are still questions to be answered. It is, however, possible to obtain very good results, if some simple rules are obeyed.

There are two basic kinds of stresses to be considered:

- pulling/pushing
  (or stretching/compressing);
- shear.

### Pulling/pushing

Pulling/pushing is the most important kind of stress when bending, at least in isotropic materials. Also fibre composite strength to bending is built up by material as far as possible from the neutral axis, to be pulled above it and to be compressed below it.

Pulling is no problem, but how about longitudinally compressing a string? We have all seen that when bending a tube with thin walls, it has a tendency to break in a buckling mode, far before tensile failure is reached (Fig. 13). Buckling is a stability problem, and is handled by structional stiffness or by support. The situation is improved if the wall is made thicker, or if the tube is filled with some material, that supports the wall and prevents it from buckling. If this is the case, most materials are as strong in compression as they are in tension. This is also true for our fibres (except for Kevlar). But if the fibres are not straight when the longitudinal compression starts, buckling has already been initiated. But if everything is right, fibres can handle longitudinal compression stress.

### Shear

Shear not only occurs when there are external

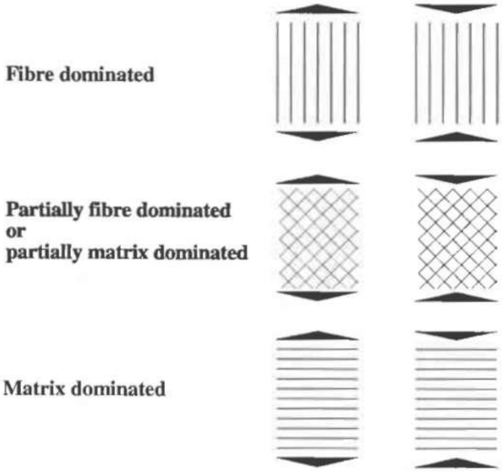


Fig. 12. Fibre domination and matrix domination.

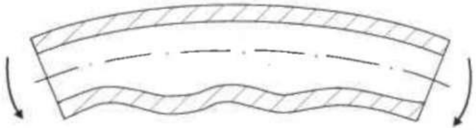

Fig. 13. Buckling.

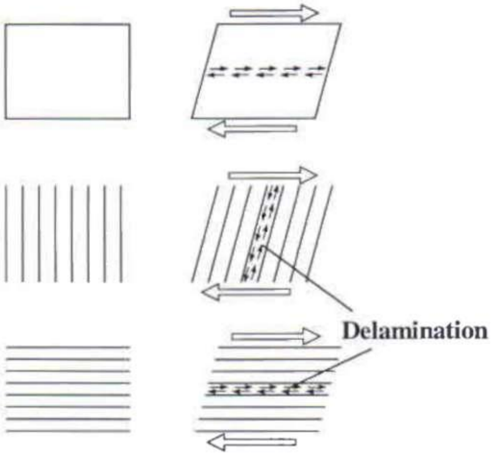


Fig. 14. Shear in matrix domination.

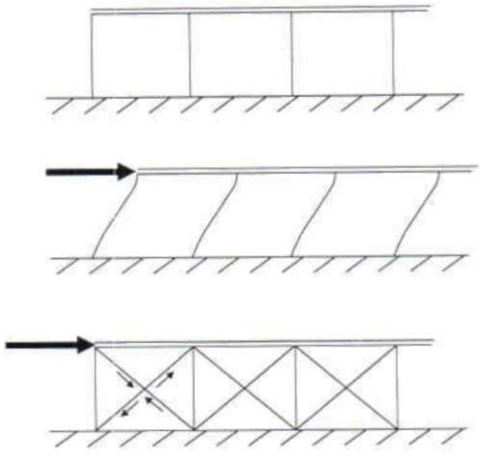

Fig. 15. The fence method.

shearing loads such as in torsion. There is usually (not in "pure" bending) internal shear when there is bending (see later). There are actually, if you penetrate the subject much deeper, stresses and strains due to pushing or pulling and due to shear in all the three dimensions, when there is a load. If the material will not resist shear, the other tensions will be the same everywhere and in all directions in the material, and the material behaves as a fluid obeying the laws of hydromechanics. Fibre composite mechanics has a little of hydromechanics too in its more advanced aspects, because plasticity of the matrix includes a tendency to relax shear stresses and redistribute (and "equalise") the load distribution. So there is partially hydromechanics, a lot of solid body mechanics and the complicating element of anisotropy. No surprise then, that it is sometimes confusing and that it may be difficult to interpret experiences.

Back to our more practical approach! If the above fibre configurations are used, the result will be delamination due to the matrix domination mode (Fig. 14). The shear forces cause the matrix between the layers of fibre to fail or delaminate. When such delaminations occur, careless manufacturing is often blamed, although it would probably be more reasonable to draw the conclusion that the shear strength of the matrix was not sufficient.

But this is only if matrix domination has to be accepted. The fibre dominated solution to this problem is the same as when reinforcing a fence (Fig. 15). We orient the fibres in such a way, that shear is transferred to extension and compression of fibres and not borne by the matrix. It is the same as in framework construction where bending moments in the beams are not accepted.

**Laminates**

In fibre composite language, fibre orientation is defined by angles (Fig. 16). A lay-up may consist of 0° (zero degrees, i.e. parallel to the "main" force) fibres to handle basic stresses from bending, 90° fibres to handle perpendicular, delaminating stresses and 45° fibres to handle shear. These different angles are usually (for obvious reasons) not woven together. They are added on top of each other, layer after layer, and this is actually the process of lamination. A work instruction, telling

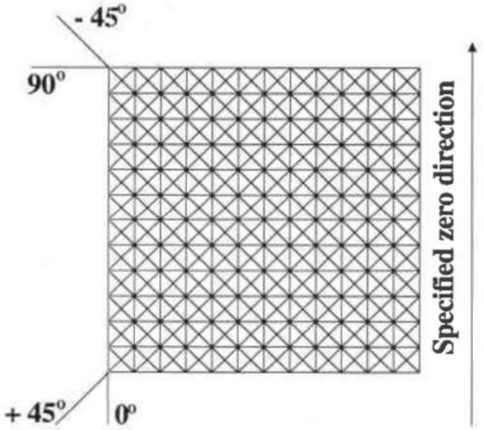


Fig. 16. Fibre orientation in laminates.

the orientation of each layer, may look like this:

1x +45°
1x -45°
1x +45°
1x -45°
5x 0°
2x 90°
core
2x 90°
5x 0°
1x -45°
1x +45°
1x -45°
1x +45°

This is a symmetrical sandwich, and the above instruction tells the operator how to orient the fibres.

Each layer is called a lamell, and the total result of several lamells is called a laminate or, in a sandwich structure, a face. Weaves, sheets, rowings, prepegs etc. used for this purpose will not be reviewed or discussed here except for two comments:

1. In prosthetics and orthotics "angulated weave" tubing has become very popular to handle shear. When the tubing is stretched out over a small cross-section the angle of the fibres is small, close to 0°. As the cross-section increases, the length of the tubing decreases and the fibre angle increases. If the extremes are avoided, the angle kept to say between 30° and 60°, this is a very elegant solution, also providing a closed loop for the shear flow at torsion.
2. There are weaves available where the carbon fibres are looped. The author has so far not heard a satisfying explanation of the reason for using expensive CF instead of GF when the composite is going to become matrix dominated anyhow!

**Holes**

Holes are very dangerous in laminates and should, if possible, be avoided. Unfortunately this is not always possible. Holes interrupt the continuity of the fibres, not only where the hole is, but also at a long distance from the hole (Fig. 17). The interrupted fibres can take care of load only where the load has been transferred to them by matrix shear from other adjacent fibres. As a result, there is a risk of delamination due to shear stress in the matrix not only at the hole, but also very far from it. How about letting the fibres pass outside the holes? (Fig. 18). Sorry, but when the load is applied, there is a tendency to straighten out the fibres resulting in perpendicular stresses in the matrix, and again there is a risk for delamination. Winding fibres around the holes does not help either. Firstly it just moves the problem to outside the winding, and secondly the radius would probably be smaller than the fibres would tolerate.

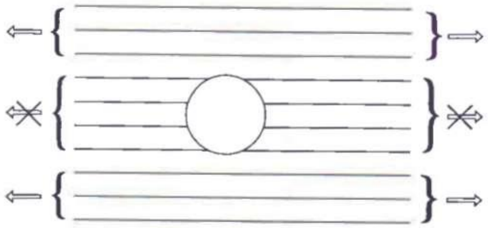
Fig. 17. Drilled holes.

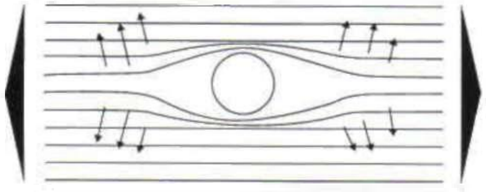
Fig. 18. Fibres passing outside holes

*Holes in a continuous fibre composite always result in high shear stress, partial matrix domination and risk of delamination.*

**Permanent, plastic deformation**

The reason why a metal rod can be permanently bent is that the material becomes plastic when the stress exceeds the yield stress. This property of

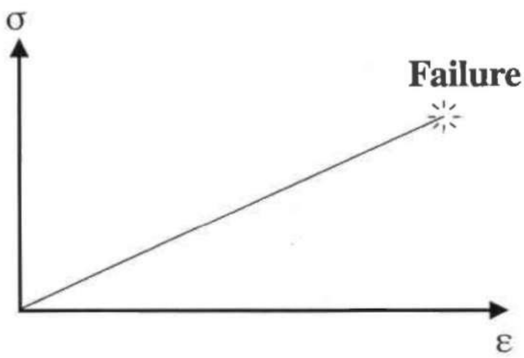

Fig. 19. Stress versus strain in GF and CF (no plastic range).

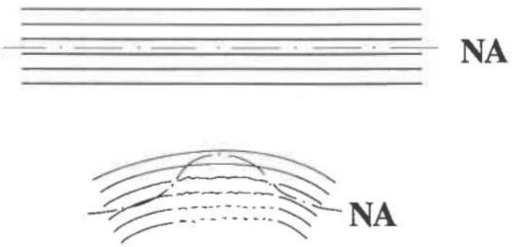


Fig. 20. Fibre collapse at buckling.

steel and other metals allows us to adjust for instance an orthotic upright. Neither glass fibres nor carbon fibres have any yielding properties. They do not deform permanently due to load (except for GF when heated to high temperatures). They deform elastically until they fail (Fig. 19).

Consequently a GF or a CF structure *cannot* be corrected or modified plastically after curing, at least not if the fibres are continuous. Also when heat is used and the matrix is thermoplastic, the compressed fibres lose their support, buckle and collapse (Fig 20). When they collapse they no longer contribute to the fibrous cross-section and the neutral axis moves towards the opposite face. We have nicely converted our fibre dominated composite to a matrix dominated one, and wasted our expensive fibres. If our composite still serves the expected purpose, there was no need for the fibres in the first place. A 100% fibre dominated GF or CF composite is, however, virtually not sensitive to fatigue. The matrix material may be much more sensitive to fatigue, and if there are no fibres to limit the extension due to loads, the structure may break after too short a time in use. Plastic deformation of a rod, with maintained fibre domination is possible only if the fibres can "slide" relative to each other and remain stretched in the correct orientation. This is possible if the total length of the rod is heated, and if the angle between the ends of the rod is kept unchanged (Fig. 21).

The above is an endless source for discussion and misunderstanding. The author has seen several demonstrations, where manufacturers have proudly introduced thermoformable carbon fibre sheets. But looking through a strong magnifying glass at the side where the fibres have been compressed has so far always given evidence, that many of the fibres have terminated their useful careers.

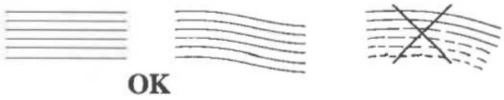


Fig. 21. Plastic deformation with thermoplastic matrix.

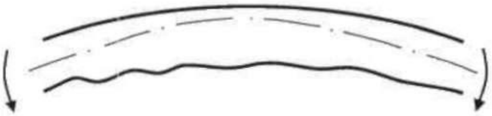

Fig. 22. Buckling of compressed face.

**In sandwich design**

When subject to bending the compressed face has a tendency to buckle (Fig. 22). Buckling is, mechanically, a *stability* problem, with very small forces required to initiate it. The main task of the core, besides maintaining the moment of inertia, is to support the face so that buckling is never initiated. For this purpose, the core does not have to be very strong, but it has to be *stiff* and well bonded to the face. Consequently, and for more than one reason, the use of a spongy core may not be very clever.

If a composite beam is curved, and subject to bending as in Figure 23, there will be a tendency to straighten out the convex face and further bend the concave face (b). If the core is not able to resist the stress thus developed between the faces, there will be a compression and a structural collapse and a loss of moment of inertia, thus, the structure will become weaker.

If the beam is being bent the opposite way, the opposite will happen, and we are likely to see a delamination due to the perpendicular stresses pulling the faces apart. In this case also the demands upon the bonding between the core and the faces increase.

In both the above cases the core needs to be strong enough to handle compression or extension. Perpendicular (90°) lamination may help, also if there is no sandwich structure, but a "homogeneous" lamination (Fig. 24).

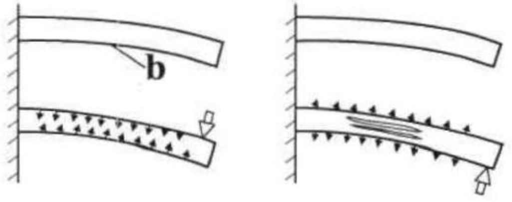


Fig. 23. Bending of curved beam.

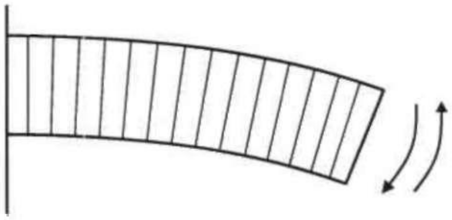

Fig. 24. 90° (perpendicular) lamination.

The "rule of thumb" is *that composite beams should be made straight if possible.* A change of the geometry may give a local bending moment, which stresses the matrix and may cause delamination in a homogeneous laminate as well as in a sandwich structure.

Let us look at a beam, consisting of two planks (Fig. 25). One is on top of the other, and they are not bonded to each other. When the beam or "beam-pack" is bent, the compressed side of each of the two parts will become shorter, and the stretched sides will become longer. As a result, it will look like the lower one is sticking out past the upper one. This does not happen in a solid beam because shear stresses prevent it. The shear stresses due to bending are zero at the upper and the lower end points of the cross-section, and they have their highest value on the neutral axis (Fig. 26). (Shear stresses due to bending do not appear when there is "pure" bending with constant bending moment along the beam).

In isotropic materials, like steel, the shear stresses due to bending can usually be neglected (exception: very short beams where, as a rule of thumb, the height of the beam is more than 1/5 of the length). In composite structures and sandwich structures, these shear stresses have a tendency to cause delamination or fracture of the bonding between the faces and the core or of the matrix between the fibres. They stress the matrix and, if the matrix is sensitive to fatigue, delamination and

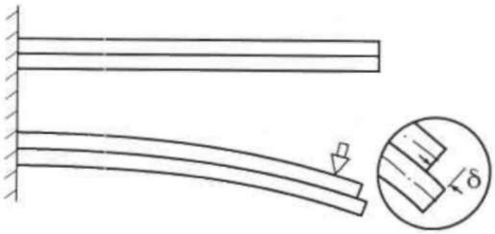

Fig. 25. "Beam pack".

Fig. 26. Shear due to bending.

failure may occur with limited use. The simplest solution is to use a high quality matrix and bonding and of course a core material that can handle shear flows.

The structure is converted to a fibre dominated one, at least at the edges, by adding a 45° (30° – 60°) lamination between the faces (Fig. 27). It may also, in a partially fibre dominated way, substitute for 90° lamination as shown in Figure 24.

This lamination will take over the shear load in its vicinity, but further away its influence fades out. If it is repeated at certain intervals by putting in webs, you have a "multi-box" less dependent upon matrix shear strength, but also much more expensive.

## Bending of fibres

Due to the high modulus (stiffness) of CF high stresses are developed when bending the fibres. This is why special attention is required in forming and handling. To avoid fracturing CF the bend radius should be as large as practical. If the fibres are forced to conform over a sharp edge, breakage is likely to occur.

*Rule of thumb: avoid bend radii of less than 40mm (1.5") for CF, the more critical the higher the modulus.*

## Composites

Fibre composite design calls for knowledge of the production methods. The properties of the composite depend on how it is made. *Pultrusion* stretches the fibres longitudinally, while *filament winding* stretches them tangentially. Hand lay-up often leaves the fibres loose.

Data that is available on the properties of long fibre composites is, as said before, related to well stretched, parallel, continuous fibres, where the force is parallel to the fibres. Thus, the fibres

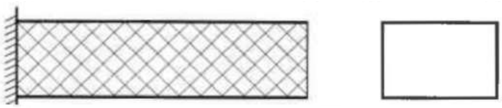

Fig. 27. Fibre orientation for handling shear due to bending.

determine the strength as well as the stiffness of the composite. If the force is perpendicular to the fibres, the matrix determines the strength, and the stiffness is determined by the fibres and the matrix. Short fibre composites approach the properties of perpendicularly loaded long fibres, i.e. significantly less fibre domination for strength than for stiffness.

How is it that a fibre composite can be matrix dominated regarding strength and partially fibre dominated concerning stiffness in the very same direction?

Look at the matrix dominated orientation in Figure 12. The fibre to matrix ratio does not matter. Also if there is 90% fibre, the fibres never bridge the force flow. The composite is 100% matrix dominated concerning strength. But if the fibres are much stiffer than the matrix, which they are, they will act as a stiffening filler. Then the composite is partially fibre dominated as far as stiffness is concerned, be it that we are no longer talking about loads along the fibres.

In reinforced concrete the reinforcing steel bars may be pre-stressed. There is a tension in the bars before external load is applied. It would be wrong to say that the fibres in plastics are pre-stressed the same way. The thermal shrinking of the matrix during setting may actually *compress* the fibres and the yielding of the matrix after setting relaxes the stresses. As a matter of fact, CFs may have a negative thermal expansion coefficient, which means that they shrink when heated during setting, and expand again when cooling to normal temperature.

It is important that the matrix tolerates a longer elongation to failure than the fibres with a considerable margin. If this is not the case, the properties paid for of the fibres cannot be fully utilised as matrix failure ruins the laminate before the fibres take the full load.

As the purpose of the matrix is first of all to create working conditions for the fibres, it is obvious that the fibre content should be high. When using autoclave, pultrusion or filament winding processes, 65% fibre by volume is the highest value for long fibre composites except for extreme requirements, when 70% may be reached, It is however, possible to exceed 50% by more primitive methods. It deserves attention, that the quality of laminations performed in limb fitting shops, using a combination of vacuum membrane moulding combined with manual removal of excess matrix, is sometimes so high that the composite industries are unable to offer competitive solutions!

For short fibres 15-30% fibres by volume is normal.

If cylindrical fibres are ideally distributed and pressed together, the space between them represents about 9% of the volume, and the fibres about 91%. There is, however, no reason to try to exceed 65-70%, because then there is a risk that the matrix may not be able to cover the surface of the fibres completely, resulting in inferior bonding.

There is no point in copying conventional metal structures or designs and making them from fibre composites. It is often said that the key to composite design economy and success is *integration of functions* or *integrated design*. It is not the purpose to emphasise on design philosophies here, so let us avoid it by suggesting that as much as possible is integrated in large modules and units instead of assembling a lot of very specialised small components. Specialised components are bolts, hinges, bearings etc.. It is not only the anisotropy that causes problems. Bolting, riveting, press fits etc. do not behave as in metals, very much because they rely upon pre-stresses possible, because the metals are used in their elastic range. The plastic matrix materials may not offer such assistance for a longer period of time.

One further reason that makes integrated design attractive is that fibre composites can be formed to shape without waste of material due to machining.

It may be reasonable at this point to draw the conclusion, that investing in "materials substitution" projects, where fibre composite components or modules are supposed to replace conventional metal ones may not necessarily be a very sensible approach.

## Prepregs

A prepreg is a tape or a fabric with unidirectional fibres, impregnated by a thermosetting resin, usually an epoxy resin, to serve as matrix.

The resin is partially cured to a "tacky" state (B-state). The resin is finally cured by heating under pressure when the intended structure has been formed.

Narrow tapes are used when the shape is more complicated, while large sheets are used for bending in one plane (aircraft wings etc.).

**Fibres**

The basic requirements for fibres in the composite are:

- higher tensile strengths than the matrix, as the load is supposed to be carried by the fibres;
- higher modulus of elasticity than the matrix, which is a condition (Hooke's law), that the fibres take the load;
- less tension to failure than the matrix, to avoid the matrix failing before the fibre strength is fully utilised;
- good adhesion to the matrix;
- good chemical resistance to the matrix.

The fibres are usually divided into three classes:

- low modulus fibres. E<5000 MPa (synthetic and cellulose);
- medium modulus fibres. E=5,000-150,000 MPa (glass and synthetic);
- high modulus fibres. E>150,000 MPa (synthetic, whiskers, carbon, boron).

Types of fibre are:

glass - good strength, low cost;

Kevlar - toughness, poor compression properties and thus less convenient for structural designs;

carbon - stiffness, strength, good shock absorbing and friction properties, low weight.

CF fibre is brittle. CF composites have no yielding behaviour, and tolerance to impact is low. This tolerance can be improved by mixing in some 30% by volume of Kevlar, glass fibre or synthetic fibre for protection.

CF has good electrical and thermal conductivity. Electrostatic painting is possible. The heat dissipation is low.

It should be noted, that some high modulus CFs exhibit zero or even a negative thermal expansion coefficient.

When comparing the relative merits of CFs and GFs in practical applications, the following may serve as a "rule of thumb" summary:

- in a fibre dominated composite, the strength will be about the same, but a CF composite will be about three times stiffer than a GF composite.

If there are 50% fibres, the CF composite will be about 10-15% lighter. The strains and consequently the stresses in the matrix will be about 1/3 when using CF.

The fibres can be long or short. Short fibres cost less and fabrication costs are lower in mass production.

Graphite is by definition a three dimensional crystallographic structure, which is not present in commercially available fibres. The term "carbon fibre" is technically correct. "Graphite fibre" is not. There is no such thing, and it is unfortunate that the term has ever been introduced.

The early development of CF may be summarised as follows:

- carbonised cotton 1879 (Edison)
- carbonised Rayon (terribly expensive) late 1950's (Courtauld and Union Carbide)
- stretched high modulus fibre 1965 PAN (polyacrylonitrile) based fibre (expensive) 1969;
- high modulus fibre from petroleum pitch (great promise for cost-effective industrial applications. High modulus per dollar);
- continuous pitch fibre 1976.

CF is mainly made by carbonising polyacrylonitrile fibres at high temperatures. The tensile strength and the modulus of elasticity can be adjusted by varying the end temperature in the final stage of the process.

Kevlar is a marvellous fibre when energy has to be absorbed. This is why it is used in bullet-proof vests, in sails and in other similar, tough applications. The author has never had any reasons to work with it, but would be delighted to pass on an advice given by several very experienced US experts many years ago: "The best way to use Kevlar in structural designs is to avoid it! It is so full of surprises for the developer that the development costs become disastrous!"

The main reason for the problems is probably the poor compression tolerance of the Kevlar fibre.

The fibres are very thin, typically 6-9 microns diameter for CF and 12 microns for Kevlar.

**Coupling agent**

After pulling, the fibre is usually treated with a surface coating, sometimes called sizing, to make it anti-static and to protect against wear and fracture etc. during the continued handling (spinning, weaving etc.). Sometimes the sizing has to be removed (by burning) before the fibre is laminated, but often it serves as a coupling agent to improve the bonding to the matrix. The quality of this bonding determines the quality of the strength properties and the resistance to damp environment of the composite. Different coatings are used for different fibres and sometimes for different matrices. The most common coupling agents for glass fibres are silanes. Vinyl silanes are used for polyester. Metacrylic based silane with polyester results in a transparent composite.

Amino silanes are used with Epoxy and phenols.

The sizing on carbon fibres is usually low molecular epoxy, not completely hardened.

*Warning: the above sizing on carbon fibre is usually aggressive to your skin, and should be handled like working with epoxy resins. Use long gloves to avoid skin contact.*

**Mixing fibres**

Sometimes CF is used in prosthetic and orthotic applications with the intention to locally increase the strength or the stiffness in a hand lay-up GF composite. As it is difficult to stretch the fibres, the gain is often minimal.

There is more to it, however. This may be visualised by looking at an example where the idea is that a laminate is supposed to be stronger if some GF is replaced by CF. This is a common idea. But it is not necessarily a good one, and some simple calculations may explain why, at the same time as they demonstrate how the available data can be used. Hopefully the departure from explaining without mathematical analysis will be excused.

The manufacturers information sheets tell us:

| Fibre | $\sigma$(failure) (MPa) | E (GPa) |
|---|---|---|
| Glass | 3,400 (or 3,400 N/mm$^2$) | 70 (or 70,000 N/mm$^2$) |
| Carbon | 2,600 (or 2,600 N/mm$^2$) | 210 (or 210,000 N/mm$^2$) |

Hooke's law says: $\sigma = E \times \varepsilon$, or, the higher the modulus the faster the stress develops as the material is stretched.

Hence for the GF the extension before the fibres break is 3,400/70,000 = 0.049, which is about 5%.

For the CF we get 2,600/210,000 = 0.0124, which is about, 1.2% extension before all the fibres break.

If we have a rod with a GF cross-section of 100mm$^2$, the fibres will break at a total load of 3,400 x 100 = 340,000 N. The equal amount of CF will carry only 260,000 N, which is less, but not very much less. But what happens if we replace 20% (by volume) of the GF with CF?

We know that all the CFs break at 1.2% extension, and as there are 20mm$^2$ of them the load they carry when they break is 2,600 x 20 = 52,000 N. The extension of the GFs is the same, 1.2%, at this point is 0.012, and hence the stress is 70,000 x 0.012 = 840 N/mm$^2$ in a cross-section of 80 mm$^2$. Hence the GFs carry 67,200 N when the CFs give up. The total load is at that moment 67,200 + 52,000 = 119,200 N. When the CFs have given up, the 80mm$^2$ of GFs are left to make the best of the situation, and they give up of course at a total load of 80 x 3,400 = 272,000 N.

The above is illustrated in Figure 28, but the bottom line is, that by replacing 20% of the GF with CF, we actually *sacrificed* 20% of the strength. The situation would have been even worse, if 50% of the GFs were replaced by CF.

There are hybrids, though, where GFs and CFs are said to work together in such a way, that they break simultaneously. One explanation to this is that the GFs and the CFs are stretched differently in the manufacturing process.

*Rule of thumb: mixing of different fibres is pointless if the interaction between the fibres is not carefully taken into account.*

**Matrices**

The main roles of the matrix are:

- stabilise the fibres and keep them in place;
- distribute and transfer loads between the fibres;
- transfer loads at partial or total matrix domination, notably shear loads;
- protect the fibres against the surrounding environment.

The matrix may be thermosetting or thermoplastic.

Thermosetting matrices include epoxy, vinyl ester, polyester, polyurethane, acrylic etc.

Thermoplastic matrices include nylon, polyethylene, petroleum pitch etc., but also metals like titanium and magnesium are used.

If the fibres are supposed to transmit the loads, the fibres must fracture before the matrix. If not, the fibres cannot be utilised to the limit of their strength. Hence the matrix must have a longer elongation (strain) to fracture.

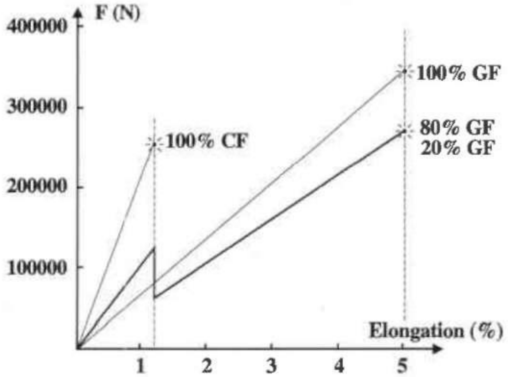


Fig. 28. Mixed fibres.

It is not correct to say, that the bond between the fibres and the matrix should always be as strong as possible. Too strong a bond may give a fibre cleavage failure mode and to weak a bond a delamination failure mode. There is an optimal bond for maximum tensile strength in a composite.

The different coefficients of thermal expansion of fibre and matrix will develop internal tensions when heating and cooling, also when there is no external load. When tensions due to external loads are added to these internal pre-tensions, we may too soon reach the tolerance level and get failure at less external load than we had expected. We may actually get cracks in the matrix before any external load is applied.

The first critical moment is actually during the exothermic setting process when laminating with a thermosetting matrix. If the internal stresses are eliminated at peak temperature during the setting, stresses will be developed during cooling, the greater, the more difference between the expansion coefficients. The thermal expansion of the fibres can usually be neglected. Hence it boils down to a requirement, that the thermal expansion coefficient of the matrix should be as small as possible. Different sources report different figures, but it is hopefully possible to agree that polyester expands about 10 times more than epoxy, and acrylic expands 10-30 times more than epoxy per unit temperature change.

This is one of several reasons why epoxy is the technically superior matrix material. Other reasons are that it is best for wetting, protection and fatigue. For the most advanced high tech CF applications epoxy is more or less the one and only serious alternative. But it is also the most expensive, and it is very aggressive to the skin before hardening.

The chemical resistance of epoxy depends on the hardening system. It is generally good, but amine cured epoxy has poor resistance to acids. Anhydride cured ones neither resist strong alkalis nor organic solvents. There is a wide range of epoxy resins and hardeners available, and there is a lot to gain by selecting them carefully to satisfy the requirements.

In mass production, at pultrusion and filament winding, epoxy is sometimes substituted by vinyl ester (CF sailing masts, ski poles etc.) with very good results due to its excellent resistance to weather and chemical stresses combined with good mechanical properties.

But GF with polyester is excellent for many applications. In optimised (integrated) design it made possible a weight reduction from 8 to 2 pounds of a rear wheel suspension element of a Volvo truck. The springs of the Corvette sports car are GF with polyester matrix.

It should be noted, that shrinking during manufacture using polyesters as well as vinyl esters, can be significantly reduced by using additives.

Acrylic is an interesting compromise. The reason why it is used in prosthetics and orthotics, also for CF, is probably that it has a successful tradition there for use with the other fibres. It was early discovered, that it had to be modified by thinning to be able to wet sufficiently for use with CF, and we witnessed the birth of "carbon acrylic". It may be an ideal matrix to work with in prosthetic and orthotic shop conditions, but it is certainly not the best way to utilise CF. Acrylics are actually very seldom, if ever, mentioned in the composite literature.

During the past decade a lot of interesting development has been going on in the area of fibre composites with thermoplastic matrices, but so far no simple method to add the matrix to the continuous, long fibres has been introduced. Such laminates are commercially available, and if the user knows what he is doing, they can be very useful. Sometimes the matrix is "semi-thermoplastic". If epoxy is used, it may be possible to soften it by heating and deform it permanently once or twice. This property expires, however, when the cross-linking starts (see below).

Unfortunately, however, it is frequently believed, that if the matrix is thermoplastic, the composite is also thermoplastic. This is, as discussed above, not true, at least not if the composite is supposed to maintain its strength properties after deformation.

It is very different with short fibre composites. Nylon with short GF or CF has become very successful, also in prosthetic and orthotic applications. These composites also exhibit a true, but limited thermoplastic behaviour.

**Matrix strength**

As indicated above, the most sought after strength in the matrix is shear strength. This is very much because it is sometimes more or less impossible to avoid matrix domination for shear stresses due to bending.

It is frequently suggested, that the matrix should be "post-cured" after hardening by heating to about 80° Celsius for a couple of hours. Many thermosetting materials give a better final result if heated to a much higher temperature. Some materials should actually be heated up to softening (for carbon acrylic 180°C) to enable them to develop full intramolecular strength by *cross-linking*. Tests reported by Hexel of France show that considerable permanent improvement of the shear strength in their epoxy bond for car body panels is possible by cross-linking by 150°C postcuring.

### Core

The main roles of the core in a sandwich structure are:
- to secure the geometry of the cross-section, i.e. the moment of inertia;
- to support the compressed fibres to prevent them from buckling;
- to resist shear stresses due to bending.

The demands upon the core material vary, according to the specific conditions at hand. Some requirements are general, though:
- good bonding to the matrix in the faces;
- sufficient stiffness to keep the faces in place and prevent them from buckling.

Specific requirements, related to the conditions include:
- strength, especially shear and compression;
- low cost;
- light weight,
- thermal resistance;
- chemical resistance;
- water resistance;
- UV resistance;
- flexibility.

Foams and honeycombs are frequently used for lightweight cores. Also felt can be used, provided that it is saturated by the matrix resin during manufacture.

In the ISNY version of the Scandinavian Flexible Socket GF is used for core, with the same matrix as for the faces. This is, of course, an excellent solution, if weight is not very critical.

In our early development of CF orthoses an Omega-shaped profile was used to maintain the cross-section. Unfortunately, the profile had a tendency to collapse at bending. It worked like a corrugation and, as the shear flow was not closed, it could not handle torsion either (Fig. 29).

To facilitate easy shaping of the core now used for our CF orthoses, we initially cut slits in the

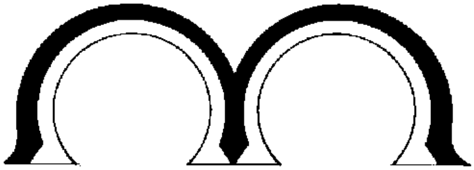

Fig. 29. Omega profile.

edges of the core (Fig. 30). It turned out, that during the lamination, the fibres were bent over the incisions by the external pressure (or the internal vacuum). We actually initiated buckling and slight matrix domination. This was enough to reduce strength as well as fatigue resistance.

We have come across a procedure, recommended by a supplier, where no core is used. Furthermore, a woven CF mat is used, where the fibres are not straight (commented on above). It is shown as "X" in Figure 31.

It is probably not unfair to say, that this is an example of waste of expensive material.

Figure 31 shows results of comparing some different core concepts in the apparatus shown in Figure 2. The PI kit for manufacturing medial upright and brim for an ISNY socket (distributed by PI Medical Co. in Sweden) was developed by two of our skilled technicians after having received fibre mechanics education equivalent to the content of this article. The core is made from ABS.

### Fatigue and lifetime

CFs and GFs resist fatigue loads very well. Some sources indicate that the fibres themselves

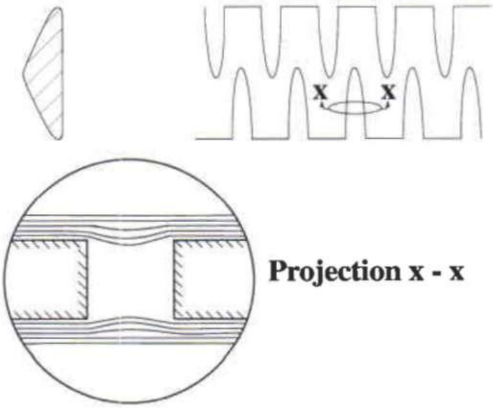


Fig. 30. Slits in core.

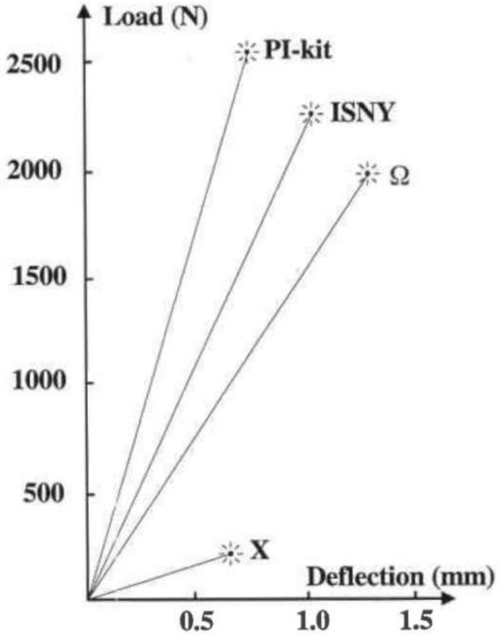


Fig. 31. Deflection with different core designs.

are not at all sensitive to fatigue. The problem is not the fibres but the matrix. Although there may be 100% fibre domination, there is always strain, and consequently also stress in the matrix. Thus the composite is of course sensitive to fatigue also if only the matrix is sensitive to fatigue, and some matrices are. As CF is three times stiffer than GF, CF composites offer better fatigue properties than GF composites because the matrix is about three times less strained.

A rule of thumb is, that for CF epoxy composites fatigue can be neglected if all fibres are parallel to the load (100% fibre domination). In less "pure" fibre orientation, with a mix of angles, at least 40% of the calculated static load can be tolerated as fatigue load (tolerated for 10,000,000 cycles). This is much better than steel, which is much better than aluminium alloys.

If designed properly, however, GF polyester makes for excellent springs with a long lifetime.

It is absolutely reasonable to state, that one of the distinct advantages of well made fibre composites is that they offer excellent fatigue resistance.

But there is a cloud in the sky. The strength of some matrixes, including the highly praised epoxy, decreases by ageing, and hence there may all of a sudden be a failure. This is not fatigue in the classical sense, but it definitely restricts the lifetime.

The fatigue limit is, by the way, in the classical sense defined as the highest stress accepted by the material for an eternal amount of times. In engineering it has been agreed, that for testing 10 million times represent sufficiently well an eternal amount of times. Repeated loads under the fatigue limit do not consume lifetime, whilst loads above it reduce the lifetime. This is not generally true, however. It is all right for steel, but aluminium and aluminium alloys have no fatigue limit. Every time they are stressed, a fraction of the lifetime is consumed. This is a reason why the author has never trusted aluminium components in prosthetics and orthotics, and it is a problem that aircraft designers are very well aware of. Hence, with reservation for ageing, fibre composites may be much more attractive than aluminium alloys as far as fatigue is concerned.

Fatigue in fibre composites usually presents itself as delamination, and tends to start at holes and edges (where the shear stresses in the matrix are highest). Dynamic loading before use to about 80% of maximum load is a common treatment to improve the static and the dynamic strength.

Kevlar is sensitive to UV light and some glass fibres are sensitive to water. Hence a good seal is necessary, especially if the composite is cut across the fibres, to avoid capillary penetration of substances and rapid deterioration. Better wetting and adhesion improve long time strength and fatigue resistance, which is an argument for the use of epoxy.

Electrocorrosion may reduce the lifetime of screws, rivets and metal embedments in CF composites. Stainless steel or titanium are strongly recommended.

**Lifetime tests**

Industries as well as authorities frequently perform lifetime tests, applying a certain load to a certain amount of cycles. Most of these tests are accelerated i.e. they are run at such high frequencies, that the "lifetime" behaviour can be tested in days or weeks. Accelerated tests of fibre composites do no usually provide true information. When testing a prosthetic foot, the author got much better lifetime, if the frequency was reduced and the tested object was allowed to rest overnight.

This is a subject that should be closer looked into, to avoid obsolete or inadequate test procedures preventing a very promising development.

## Safety

CF irritates in much the same way as GF. Furthermore, the risk of skin contact with not completely hardened epoxy sizing before lamination must always be avoided (see above). Skin irritation from such materials through pockets in trousers have been reported. Use work gloves, protective clothing, eye protection and fume extractor. Wash clothes used at CF or GF work separately.

Horror stories have been told about risks. One is that fibres, that find their way to the lungs, will stay there and occupy more and more space. Doctors have ensured, that there is no way that they will get there. The author prefers to trust nobody and be careful and use protection.

It has been suspected, that certain similarities between the CF molecules and the asbestos molecules give reasons to believe, that there may be similar risks. Experts have stated, that this must be a misunderstanding and that there is no documentation of such risks. Evaluating this conflicting information is beyond the author's competence.

There is, however, a lot of documentation of the inertness of carbon, and the author has had pieces of carbon in his body for 45 years without the slightest irritation or tendency to rejection.

Initially it was recommended that electrical equipment operating in the vicinity of CF composite fabrication should be protected against possible intrusion and consequent short circuit. Broken pieces of CF and CF dust produced when grinding, during lay-up etc. are light and it was said that they were easily airborne and could fly long distances. It has been suggested recently, that this risk has been overestimated. It may be overestimated, but it is probably very expensive if it happens.

Use of fume extraction is strongly recommended whenever handling and working with carbon fibres. Some industries use water to catch particles during grinding, cutting etc..

The risk and the protection measures when using epoxy, polyester, acrylic etc. are well known in prosthetic and orthotic practice and do not need to be discussed here.

## How do GF and CF compare?

There are many ways to compare, depending on priorities of course, but one way or another comparisons unfortunately usually boil down to expense comparisons instead of final economy considerations.

As said above, it is easy to waste a lot of money by using CF more or less as a filler. GF is a much cheaper filler. In more advanced applications, professionally used, GF may also be as good as CF at a much lower cost.

When comparing fibre composites to metals like steel and aluminium, the costs for tooling and for manufacturing must be taken into account, and a high raw materials cost may very well occasionally be balanced out by a low manufacturing cost.

Technically, the most interesting comparison is very often the performance to weight ratio. Using

Table 1. Stiffness, strength and density (data from Exel: Designer's Handbook).

| Fibres | Density (Kg/m$^3$) | E (N/mm$^2$) | E/dens. | $\sigma$ (N/mm$^2$) | $\sigma$/dens. |
|---|---|---|---|---|---|
| E-glass | 2,540 | 70,000 | 27.6 | 1,500 | 0.59 |
| S-glass | 2,500 | 85,000 | 34.0 | 2,100 | 0.84 |
| CF-high strength | 1,800 | 230,000 | 128 | 3,500 – 8,000 | 1.95 – 4.45 |
| CF-high modulus | 1,800 | 400,000 | 222 | 2,100 – 6,000 | 1.17 – 3.33 |
| Boron | 2,630 | 400,000 | 152 | 2,100 – 4,100 | 0.80 – 1.56 |
| Kevlar | 1,450 | 150,000 | 103 | 3,700 | 2.55 |
| $SIC_4$ (Whiskers) | 2,200 | 500,000 | 227 | 7,000 | 3.18 |
| Polyurethane | 1,100 | 70,000 | 63.6 | 1,500 | 1.36 |
| Metals | | | | | |
| Aluminium | 2,700 | 70,000 | 25.9 | 230 – 700 | 0.085 – 0.26 |
| Steel | 7,900 | 210,000 | 26.6 | 500 – 2,200 | 0.063 – 0.28 |

typical values with no reference to any specific products, Table 1 compares some fibres and aluminium and steel with reference to strength and stiffness, visualising the excellent performance available from fibre composites. The reader is advised to update himself concerning currently available products.

If fatigue strength is taken into account, the fibre composites come out even better.

But again, and this may serve as a short summary, *the product has to be designed for the material, and the manufacturing process is part of the design.*

**Literature**

This is not supposed to be a reference list, but rather a suggestion to literature for further penetration of the subject.

- Langley, M. (1973) Carbon fibres in engineering.- London: McGraw Hill.

This book, with contributions by several most distinguished authors, is for many engineers the basic textbook on the subject. It is unfortunately out of print, but it is available at engineering libraries. Other excellent books, frequently referred to, are:

- American Society for Testing of Materials. Composite materials:
  - Testing and design, 1969
  - Fatigue of composite materials, 1975
  - Fracture mechanics of composites, 1975.
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Some manufacturers and suppliers supply good and sometimes excellent information material. Practitioners, like orthopaedic technicians, may find such material adequate in the development of a sufficient understanding of applied fibre mechanics and lamination technology. But it is important to make sure that such material is updated, as the supply of products referred to (specific fibres, weaves, prepregs etc.) may have been discontinued.

Good examples of such manufacturers' information are:

- Designer's Handbook
  EXEL Oy, INDUSTRIAL COMPOSITES, Helsinki, Finland.

- THORNEL Product Information, Amoco Performance Products Inc., USA.

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# Functional outcomes in a lower limb amputee population

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**Abstract**

This paper reports an evaluation of 132 patients seen at the Nova Scotia Rehabilitation Centre amputee programme during a 24-month period, carried out to evaluate the programme's effectiveness. In addition to a reveiw of charts, a questionnaire was used (85% return rate) to help determine functional outcomes. The patient profile revealed a 3.4:1 male-to-female ratio and an average age of 64.8 ± 13.0 years. The average overall training time was 44.0 ± 26.5 days. Of the respondents, 65.5% wore their prosthesis at least 9 hours/day, 11.5% wore it at least 4 hours/day, and only 16.1% were no longer using their prosthesis. The programme's effectiveness appears to compare well with that of others reported in the literature.

## Introduction

As part of an ongoing effort to improve the quality of patient care at the Nova Scotia Rehabilitation Centre, the services being provided to the public are being examined more closely. This paper focuses on the amputee population within Nova Scotia, as the Centre serves the entire province. The incidence of amputation in 1992 has been reported to be 43/100,000 per year. Approximately 380 amputations are performed each year in Nova Scotia; these include hemipelvectomy, hip disarticulation, trans-femoral (above-knee), knee disarticulation (including Gritti-Stokes), trans-tibial (below-knee), Syme, transmetatarsal, toe, and upper limb amputations. This yearly figure includes any revisions necessary within the amputee population in both upper and lower limbs.

The focus of this study is on trans-femoral and trans-tibial amputees, as these constitute the majority of patients seen at the Nova Scotia Rehabilitation Centre. Of these, the Centre sees approximately 90 new amputees per year, 83.8% of whom are 60 years of age or older. The feasibility of fitting patients in this age group with prostheses has been questioned frequently (Anderson *et al.*, 1987; Davis *et al.*, 1967; Holden and Fernie, 1987).

The purpose of this study was to test the hypothesis that the amputee services and the functional outcome provided at the Nova Scotia Rehabilitation Centre are comparable to other centres described in the literature. Prosthetic use after completion of the entire rehabilitation process may be a good indicator of functional outcome and may provide further justification for including patients in the process.

## Methods

At the Centre, patients enter the programme as a result of referral to the interdisciplinary amputee clinic for assessment. Depending on their medical status and level of independence, patients begin prosthetic training either as an inpatient or outpatient. The treatment regime consists of an initial phase of prosthesis fitting, followed by a schedule of group exercises and gait training in the physiotherapy department. The exercise regime includes: mat work focusing on cardio-pulmonary conditioning, trunk and limb stretching, strengthening of specific muscle groups involved in ambulation, and balancing activities in both sitting and standing. Gait training begins with the general principles of ambulation and carries through to such specific activities as transfers, stair

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Table 1. Sex, age, and level of amputation of 132 lower limb amputees treated during 24 months by the Nova Scotia Rehabilitation Centre's amputee programme

| | Men (n = 102) Age (years) | | | Women (n = 30) Age (years) | | | Total (n = 102) Age (years) | | |
|---|---|---|---|---|---|---|---|---|---|
| Level of Amputation | No | Mean | SD | No | Mean | SD | No | Mean | SD |
| trans-femoral | 55 | 64.3 | 9.9 | 12 | 65.7 | 12.4 | 67 | 65.5 | 12.0 |
| trans-tibial | 47 | 64.5 | 14.9 | 18 | 63.9 | 14.0 | 65 | 64.1 | 14.1 |

Patients with trans-femoral amputations ranged in age from 23 to 81 years; those with trans-tibial amputations were from 30 to 85 years old.

climbing, practice on various terrains and where appropriate, practice with recreational activities. Ongoing medical care is provided in consultation with the attending physical medicine specialist. Decision for discharge is based on the following criteria: independence for ambulation on gravel, stairs, incline, and carpet. The ability to get down to the floor and rise independently was an important safety requirement. Upon discharge, a routine three-month follow-up appointment is arranged.

The charts of 132 patients who had a trans-femoral or trans-tibial amputation and had completed the amputee programme in a 24-month period were reviewed. From the chart review, was obtained the patient's age and sex, the date and level of the amputation, the date of the patient's initial visit to the amputee clinic, whether the patient was referred from a hospital in metropolitan Halifax or from one elsewhere in the province, and the length of the patient's training time.

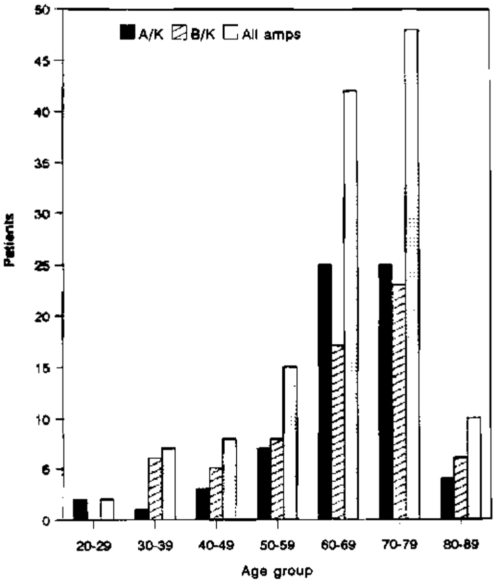


Fig. 1. Age groups of all amputees.

In order to determine the patients' functional outcome over a period of 6-24 months after discharge from the programme a questionnaire was sent to all 132 patients, requesting information pertaining to use of their prostheses, ambulation aids, and wheelchairs and to their avocational activity level. Usable data were received from 103 (78%) of the patients. Some 112 returned their questionnaires – an 85% return rate – but nine questionnaires (8% of those returned) were unusable.

## Results

The patient population consisted of 102 men and 30 women with approximately 3.4 men for every woman in the total population. The ratio was 4.6:1 among trans-femoral (above-knee) amputees and 2.6:1 among trans-tibial (below-knee) amputees (Table 1). The patients' ages ranged from 23 to 85 years, but whereas the average age of the patient population was 64.8 years, 75% of all patients were at least 60 years old, and the age distribution peaked in the seventh decade (Table 1 and Fig. 1) Referral time (the difference between the date of the amputation and that of the patient's first visit to the clinic) was 70 ± 73 days and 45 ± 33 days respectively for trans-femoral and trans-tibial amputees referred to the clinic from Halifax hospitals and 112 ± 87 days and 112 ± 79 days for patients referred from other hospitals. The average training time was 44.0 ± 26.5 days for all patients.

Sixty-seven (65%) of the 103 respondents reported that they wore their prosthesis at least nine hours per day. Another 12% wore their prosthesis at least half of the day, and only 16.1% reported that they were no longer

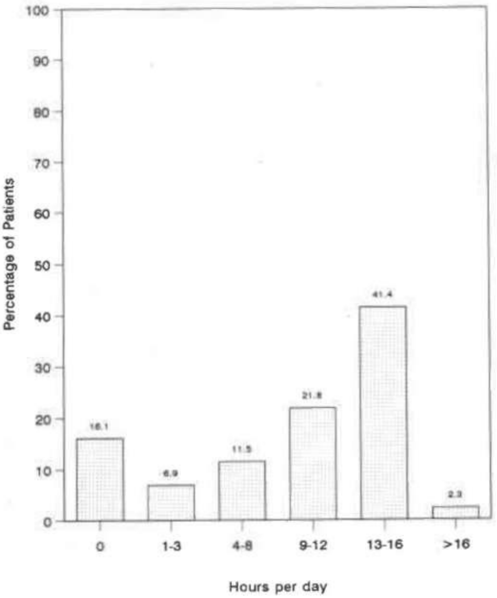


Fig. 2. Prosthetic utilization.

wearing their prosthesis at the time of the survey (Fig. 2). Most respondents reported using some type of ambulation aid (Table 2); however, nearly 75% did not use a wheelchair, whereas 18.6% reportedly used one as their only means of mobilization and 9.3% used one occasionally. Respondents reported that they regularly participated in a wide variety of activities (Table 3), with only 12.8% reporting that they had no regular activity. A comparison of functional outcome criteria from various studies is depicted in Table 4.

## Discussion

The adult amputee clinic at the Nova Scotia Rehabilitation Centre is the sole provider of prosthetic services within the province. The study group has therefore a broad-based representation of the adult population.

The male:female ratio is similar to other studies (Kegel *et al.*, 1978; Pohjolainen *et al.*, 1990; Reyes *et al.*, 1977; Weaver and Marshall, 1973).

According to current demographics, the number of older persons has increased rapidly, and this increase is projected to continue (Perreault, 1990). It is from this group that the major proportion of the amputee population comes. The average age of the subjects in this study compares with those in other studies (De Luccia *et al.*, 1992; Moore *et al.*, 1989; Pohjolainen *et al.*, 1990; Reyes *et al.*, 1977; Weaver and Marshall, 1973)

Patients should be introduced into the rehabilitation process as quickly as possible. The differences in the referral times from hospitals outside metropolitan Halifax and those from hospitals in Halifax indicate a need for further education of care-givers to refer patients early. The large standard deviation for the mean referral times in all groups reflects the multiple medical problems with which this patient population presents, and which may delay their entry into the programme. Upon entrance to the programme, after the initial amputee clinic visit, the average training time was similar for all amputees at 44.0 ± 26.5 days. The overall time from surgery to the end of training time was 127 days. There is a wide discrepancy in the literature, with published values ranging from 95 to 189 days (Katrak and Baggott, 1990; Kerstein *et al.*, 1974; Pohjolainen *et al.*, 1989;

Table 2. Use of ambulation aids

| Type | Inside (%) | Outside (%) |
|---|---|---|
| no aid | 42.9 | 28.6 |
| 1 cane | 26.0 | 35.7 |
| 2 canes | 28.6 | 28.6 |
| quad cane | 1.3 | 1.4 |
| crutches | 2.6 | 2.9 |
| forearm crutches | 1.3 | 0 |
| walker | 19.5 | 7.1 |

Table 3. Activities reported

| Type | N | % |
|---|---|---|
| daily walk | 35 | 44.9 |
| travel | 45 | 57.7 |
| shopping | 42 | 53.8 |
| driving car | 40 | 51.3 |
| housework | 41 | 52.6 |
| gardening | 19 | 24.4 |
| woodworking | 14 | 17.9 |
| cooking | 30 | 26.8 |
| fishing | 13 | 16.8 |
| camping | 10 | 12.8 |
| chopping wood | 8 | 10.3 |
| hunting | 8 | 10.3 |
| swimming | 3 | 3.8 |
| miscellaneous* | 18 | 23.0 |
| none | 10 | 12.8 |

* Miscellaneous activities include dancing, archery, boating, lifting weights, golfing, tennis, baseball, farming, latch hooking, driving an all-terrain vehicle, using a stationery bicycle, and making bicycle repairs.

Table 4. Interstudy comparison of functional outcome criteria.

| | Present | Kegel | Reyes | Pohjolainen |
|---|---|---|---|---|
| no use | 16.1% | 10.0% | 5.4% | 10.6% |
| 1-3 hrs. | 6.9% | — | — | 5.0% |
| 2-4 hrs. | — | — | 4.5% | — |
| 4-6 hrs. | — | — | — | 16.3% |
| 4-8 hrs. | 11.5% | — | — | — |
| 6-8 hrs. | — | — | 49.1% | — |
| 7-11 hrs. | — | — | — | 7.8% |
| 9-12 hrs. | 21.8% | — | — | — |
| >12 | 43.7% | — | — | 60.3% |
| no assistive device* | 42.9% | 55.9% | 2.7% | 16.3% |
| 1 cane | 26.0% | 10.4% | 40.2% | — |
| 2 canes | 28.6% | 2.2% | 5.4% | — |
| quad cane | 1.3% | — | — | — |
| crutches | 2.6% | 10.7% | — | — |
| forearm | 1.3% | — | — | — |
| walker | 19.5% | 0.7% | 5.4% | — |

* inside only.

Pohjolainen *et al.*, 1990). This may be related to such factors as the availability of health facilities and services or varying approaches to provision of prostheses.

The evaluation of functional outcomes is based on responses to the questionnaire distributed. Of the total responses, 8% were unusable with a net response rate of 77% exceeding the norm based on any homogeneous population (Dillman, 1978).

Comparing functional outcome between centres was difficult. The review of the literature revealed a great discrepancy with respect to the criteria used to determine functional outcome within the amputee population. Various centres used such activities as ambulation on stairs, ramps, public transportation, distances walked, wearing time (hours) (Pohjolainen *et al.*, 1990), driving, housekeeping chores (Reyes *et al.*, 1977), sexual activities, getting up from the floor, and living arrangements (Narang *et al.*, 1984). Our use of ambulatory aids and prosthetic wearing time is similar to the practice in several other centres (Table 4) to measure functional outcome (Kegel *et al.*, 1978; Narang *et al.*, 1984; Reyes *et al.*, 1977).

In conclusion, although the results of this two-year review reflect that similar patterns of programme effectiveness are occurring internationally, discrepancies have been identified in the literature (Table 4). The development of standardised criteria to determine functional outcomes would allow better comparisons to be made between various programmes in various geographical locations.

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# Validation of spiral CT and optical surface scanning for lower limb stump volumetry

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**Abstract**

Spiral X-ray Computed Tomography (SXCT), Optical Surface Scanning (OSS), and hydrostatic weighing methods were used to measure stump volume of trans-tibial amputees. The precision and accuracy of these methods were assessed in a validation study. A repeated measures analysis of variance statistical design was employed that required each participant to be measured (scanned) twice at each of two separate measurement sessions. For OSS and SXCT, each scan was segmented twice to determine intra-observer error. Plaster cast replicas of subject stumps were formed by certified prosthetists to serve as a reference standard. Accuracy (bias) of SXCT and OSS was determined by comparison to volumetry by hydrostatic weighing. Ten trans-tibial amputees were recruited for this study and nine completed both sessions. Plaster replica measurement precision error relative to the mean was found to be less than 1% for all modalities. The precision was slightly inferior on subjects, 1.1% and 2.2% error for hydrostatic weighing and OSS respectively, due to patient instability during measurement, but was better with SXCT where the subjects' stumps were stabilized during scan acquisition. The OSS and SXCT methods offer advantages over hydrostatic weighing and other volumetry methods since the volume data are represented digitally and can be analyzed in multiple ways. SXCT enables study of the stump and its internal tissues with the prosthesis in place. It was found that SXCT is comparable to hydrostatic weighing in both precision and accuracy. While OSS had a high precision and reproducibility, it was found to have an associated bias.

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## Introduction

The socket/stump interface is known to be the primary factor governing prosthetic fit. To support static weight bearing and ambulation, forces are transmitted from a prosthetic socket through a liner material (silicone sleeve and/or socks), through the soft tissues and patellar tendon, and into the limb skeletal remnant. A well fitted prosthesis has an optimal distribution of pressure, directing forces preferentially through pressure tolerant regions and directing force away from pressure sensitive areas. Pressure tolerance is dependent on individual and prosthesis design differences. It is widely believed that the stump can tolerate very little change in volume due to the prosthesis, therefore, any volume change to accommodate loading must be accounted for elsewhere in the socket. A review of biomechanical principles of prosthetic fitting is given in Murdoch and Donovan (1988).

Lower limb stumps consist of a bony substructure, fibrous and cartilaginous investments, and a soft tissue envelope. The soft tissue envelope is subject to both short and long term changes due to oedema, venous pooling, exercise, weight gain/loss, muscle contraction, and atrophy. In addition, this soft tissue envelope has a heterogeneous composition and is continuously deformable. Lower extremity volumetry methods have been developed to quantify and help understand the effects of oedema and atrophy (Lennihan *et al.*, 1973; Fernie *et al.*, 1978; Fernie *et al.*, 1982; Krouskop *et al.*, 1988; Persson *et al.*, 1989; Bednarczyk *et al.*, 1992). Optical surface scanners have been used in CAD CAM prosthetic design (Saunders *et al.*, 1989; Boone *et al.*, 1989; Oberg 1989; Boone *et al.*, 1994), but these reports do not include volumetric analysis of patient stumps. These non-contact methods have limited ability for atrophy and oedema assessment. Surface measurement methods (Lennihan *et al.*, 1973; Persson *et al.*, 1989) where the stump is modelled as a truncated cone, were shown to be unreliable by Fernie *et al.* (1978) and Bednarczyk *et al.* (1992). Contact contour tracers (Krouskop *et al.*, 1988; Bednarczyk *et al.*, 1992) suffer from patient motion, skin deformation from contact, and limited resolution. Fernie described and used a water bath on an elevator platform to measure cross-sectional area of the stump (Fernie *et al.*, 1978; Fernie *et al.*, 1982). All measurements were related to the distal end, and were inaccurate due to uncertainty introduced by surface tension between the skin and water as the water bath was lowered. A stated benefit of this method was to enable regional volume change assessment as volumes could be estimated for slices of a given thickness instead of a single volume measurement for the whole stump. Krouskop *et al.* (1988) attempted to compare the volumes of the socket, unloaded stump, and plaster replica on 5 subjects with contact contour tracer. The volumes determined with this method are basically an extension of the truncated cone method, with a higher number of circumference measurements. All methods described above are limited for prosthetic fit assessment as limb volumetry cannot be performed with the prosthesis in situ. A method for visualization and measurement of the stump, including its internal composition of subcutaneous fat, muscle, and bone with the prosthesis in situ is sought to aid prosthesis fitting. Spiral X-ray Computed Tomography (SXCT) has been developed for this purpose (Fishman *et al.*, 1993).

An Optical Surface Scanner (OSS) (Commean *et al.*, 1994[1]) which employs structured light was developed to accommodate lower limb trans-tibial amputees (Bhatia *et al.*, 1994[2]) and to measure distances between fiducial landmarks (Commean *et al.*, 1994[2]). While OSS does not allow discrimination of sub-surface composition nor in situ evaluation of prosthesis fit, it is a viable volumetry tool when internal information is not required.

SXCT and OSS volumetry methods define the stump in three dimensions at very high resolution. An OSS scan captures approximately 30,000 x,y,z coordinates defining the surface of the stump, over a fixed range of approximately 30cm. For a comparable 30cm Z-axis range (cranial-caudal axis) SXCT captures approximately three million volume elements known as "voxels" that represent the complete morphology of the stump. While the OSS scan range and resolution is fixed, the SXCT range and z resolution can be extended by performing multiple scans, changing the number of gantry rotations per table increment, and/or x-ray collimation. Once captured and processed into three dimensional data, computer stump models can be analyzed repeatedly. Regional volumes can be determined and SXCT volume data analyzed in terms of tissue composition, allowing registration of models by skeletal structure for analysis of regional shape, and determination of local or regional volume differences between scans. Surface and volumetric data from the OSS and SXCT are amenable to solid modelling (Bhatia *et al.*, 1994[1]; Pirolo *et al.*, 1993) enabling finite element analysis (Szabo *et al.*, 1991).

This validation study extends an initial pilot investigation performed on SXCT volumetry using phantoms and a cadaver leg (Smith *et al.*, 1995) and includes comparison with optical surface scanning as a volumetry tool in lower limb prosthetics.

## Materials and methods

Ten trans-tibial amputees were recruited to participate in this study. Informed consent was obtained from all subjects. Nine subjects completed the entire protocol. Inclusion criteria required the subjects to have been previously fitted with a permanent prosthesis and to have at least some independent or assisted ambulation. One subject did not complete the study due to personal reasons and another completed the study, but was excluded from the analysis as he had not yet been fitted with a permanent prosthesis. The mean age of the sample was 50

Table 1. Subject information for recruited amputees

| Patient | Date of Examinations (1994) | Age | Sex | Race | Side of Amp | Type of Amp | Included in Study | Prosth Type* |
|---|---|---|---|---|---|---|---|---|
| 1 | 2/18; 6/20 | 77 | M | W | L | BK | Y | P |
| 2 | 3/28; 7/1 | 38 | M | W | R | BK | Y | P |
| 3 | 2/15; 4/14 | 44 | F | W | L | BK | Y | P |
| 4 | 2/23; 4/8 | 35 | M | W | L | BK | N | T |
| 5 | 2/25; 4/12 | 31 | M | W | L | BK | Y | P |
| 6 | 3/2; 4/19 | 42 | F | B | L | BK | Y | P |
| 7 | 3/4; 4/22 | 66 | M | W | L | BK | Y | P |
| 8 | 3/15; 4/28 | 69 | M | W | L | BK | Y | P |
| 9 | 3/17; 5/6 | 51 | M | W | L | BK | N | P |
| 10 | 5/3 | 49 | M | W | L | BK | N | P |

* P denotes permanent prosthesis
T denotes temporary prosthesis

years, with 8 males and 2 females participating (Table 1). Nine of the subjects were left leg amputees and one was a right leg amputee. All subjects were at least 6 months post amputation.

Participants underwent two measurement sessions during the study in which identical protocols were followed at each session. The stump volume was measured using hydrostatic weighing (HW), Optical Surface Scanning (OSS) and Spiral XCT (SXCT) methods. Fiducial marks were used to denote an imaginary "cutting plane" which defined the enclosed volume of interest. Vertical orientation of the stump was desired, but not all patients could attain this position during the hydrostatic weighing procedure. The proximal cutting plane was therefore defined by palpation of the mid-patellar tendon and a fiducial mark made with a permanent black marker. The stump was then immersed in a water bath until the mid-patellar tendon mark intersected the water level. Two additional landmarks were then marked 120° posteriorly (one medial and one lateral) at the water level. During optical surface scanning, 6.25mm round black adhesive markers were placed atop the previously defined landmarks. It is difficult to identify marks smaller than this size with the OSS. Lead BB's (1.5mm spherical adhesive Beekley X-spots, Briston, CT) were placed over the OSS fiducial marks prior to CT scanning to define the imaginary cutting plane. In addition to the proximal cutting plane which defined the total volume of interest, a set of three additional landmarks were made with one denoting the end of the tibia at the anterior aspect and the other two placed 120° posteriorly in a plane orthogonal to the long axis of the stump. These landmarks were used in a separate investigation of distance measurements.

After the fiducial marks were identified, stump volume was measured with hydrostatic weighing. Hydrostatic weighing is based on Archimedes' principle of buoyancy (Marks, 1941) which has been shown to be accurate for measuring stump volumes (Smith *et al.*, 1995). The volume of interest was defined as the stump volume including the distal end to an imaginary plane defined by the proximal three landmarks. Repeat volume measures were obtained for determination of measurement precision.

Upon completion of the hydrostatic volume measurement the subjects were transported by wheelchair to the OSS for surface scanning. Elapsed time between HW and OSS was approximately one-half hour. The stump was towel dried and optical reference markers positioned over the previously marked fiducial points (described above). The subject was positioned in the OSS, the camera apertures adjusted for optimal exposure, and two scans recorded. The scan data were then processed to three dimensional data using triangulation and calibration methods previously described (Godhwani *et al.*, 1994). The data were visualized on a graphics workstation (Silicon Graphics 4D/340 VGX, Mountain View, CA) to ensure adequate coverage and data quality.

Upon completion of the surface scan the lead markers described previously were placed directly over the visual fiducial marks. A prosthetist then took a negative impression of the subject's stumps by plaster wrap technique. From the negative mould a plaster postive was created. The plaster positive formed with this method had small bumps which denoted the positions of the lead fiducial markers.

The subject was then transported by wheelchair to a spiral CT scanner (Siemens Somatom Plus S, Erlangen, Germany) located in Barnes Hospital (St. Louis, MO). The prosthesis was left off during all procedures to prevent disturbance of the fiducial landmarks. Approximately one to two hours elapsed between plaster wrap and spiral CT scan. The subject was oriented in a supine position on the CT table with stump resting on a foam block covered by a blanket, used to stabilize the stump during examination. A plastic rod of known volume was placed next to the stump as a calibration reference. A 32 second spiral CT scan was obtained with parameters of 120 kVp, 210mAs, 8mm collimation, 8mm table increment per gantry rotation, and a 512 by 512 sensor matrix. The scan started just below the distal end of the stump and ended above the proximal fiducial marks. The CT projection data were reconstructed at 1mm slices on a Siemens Somatom Satellite CT Evaluation Console using half scan linear standard reconstruction algorithm (Paranjpe, 1994). Two scan acquisitions per subject at each of the two measurement sessions were performed to test the precision and response stability. Completion of the spiral CT scans ended the subject's volume measurement session.

The plaster positive cast representing the patients stump was sealed with enamel paint to prevent water absortion during hydrostatic weighing. The bumps denoting the fiducial landmarks were located and clearly denoted with permanent marker. The plaster cast replicas were measured using the same protocol followed for human subjects (described above)

### Data analysis

The OSS surface model was represented in a cylindrical coordinate system with the central axis aligned longitudinally and passing through the centroid of the data. The data was re-sampled to a cylindrical grid consisting of 512 radial surface points and a 1.56mm z-axis spacing. The texture mapped OSS three dimensional surface data was cylindrically umwrapped using programmes developed in the PV-WAVE software environment (Visual Numerics, Inc., Boulder, CO). This software allowed manual digitization of the visual fiducial marks placed on the stump prior to scanning. These coordinates were used to generate a cutting plane to section the volume of interest, defined by the proximal fiducial marks. A discrete wedge may be defined from the OSS surface model using two grid points from the central axis together with four surface points. Each of the two adjacent grid points on the central axis uses two planar adjacent surface points to define the wedge. Summation of all possible wedge volumes enclosed by the stump surface and the cutting plane yields the total volume of interest. The volume in cubic millimeters was determined for the OSS surface data by this method. The process of point digitization, volume generation, and volume determination was performed twice. Figure 1

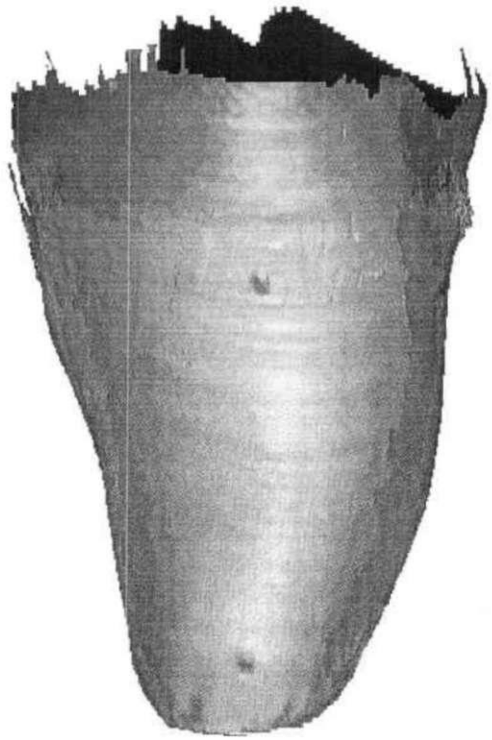

Fig. 1. Optical Surface Scan (OSS) of a lower limb stump in a frontal three dimensional view. OSS surface data is converted to a solid model by closing the stump surface with a plane through the three proximal fiducial landmarks.

shows an OSS surface data set of a trans-tibial stump with optical fiducial marks.

The SXCT volumetric stump data was imported into Analyze™ software (Robb *et al.*, 1989[1]; Robb *et al.*, 1994; Robb *et al.*, 1988; Robb *et al.*, 1989[2]) running on an imaging workstation (Sun Sparcstation 20, Mountain View, CA) for volume determination. To relax the memory requirements of the workstation and to enhance processing speed, the enormous data sets (125MB) were resampled in Analyze™ to 1mm cubic voxels, resulting in a file size reduction of about 10 to 1. Mid-point thresholding as described in Smith *et al.* (1995) was used to segment the skin surface envelope of the stump from surrounding air. This threshold also eliminated the foam block/blanket support (used to support the stump during scan acquisition) from the data. The volume of interest defined by the proximal fiducial marks was isolated by passing a cutting plane through the volume data, which intersected the three markers. The volume was then determined using the volume measuring tools available in the Volume Render module of Analyze™. Each volume measurement was done twice. Figure 2 shows a single SXCT stump scan from each of the ten subjects. The volume of interest was segmented from surrounding air and sectioned through the three proximal fiducial landmarks for volume determination. In addition, the calibrated rod volume was measured and compared to the known, to verify scan quality.

Precision relates to the ability of an instrument to consistently measure a variable over repeated trials of the same individual. Precision of the volumetry methods were tested using a precision index known as Method Error (Portney *et al.*, 1993), a measure of variance between two sets of repeated scores. Method Error is defined as:

$$ME = \frac{S_d}{\sqrt{2}}$$

where $S_d$ is the standard deviation of the difference scores.

Unlike other reliability (precision) measures, it is not affected by a low true variance among measurements as it is based on difference scores of repeated measures. To relate the error variance in the difference scores relative to the size of the mean difference the value is converted to a percentage using the coefficient of variation (Portney *et al.*, 1993):

$$CV_{ME} = \frac{2ME}{\bar{X}_1+\bar{X}_2} * 100$$

This test is applicable both within subjects and across subjects in the sample.

Validity of an instrument pertains to its suitability to measure the desired variable. For the volumetry application, the issue of validity deals with bias or systematic error as well as precision error. The precision or random error could be very small, but if measured volumes are consistently high or low from the true value (truth) then an adjustment in scale or calibration of the instruments would be required before being considered valid. This validity test obviates that the true value or "gold standard" be accurately known. For a test of stump volumes where the volume is dynamic, this test is not as meaningful. For instance, if a bias is found, is it due to systematic error in the instrument or is the change a real effect such as short term change due to swelling? To address this issue, instrument validity testing was conducted using non changing standards. These standards were comparable to the BK stumps of the subject group as they were plaster casts of the subjects' stumps taken at the time of their measurement sessions.

## Results

Of the ten subjects recruited for this study seven successfully completed the entire protocol. One subject with a temporary prosthesis did not meet the inclusion criteria, another dropped out due to personal reasons, and a third subject was omitted due to a corrupted SXCT data set. All three subjects excluded from the analysis were male resulting in a distribution of 5 male and 2 female subjects having a mean age of 52 years, a change of 2 years from the original recruited sample. The three subjects excluded from the analysis are identified in Table 1.

Each of the seven subjects was scanned twice in the OSS and twice in the SXCT to determine the error due to scanning. Each of these scans was segmented twice into the volume of interest (defined by the markers) to determine the error due to segmenting the volume of interest. The patients were brought back for a second visit where the above process was repeated to determine error due to fiducial landmark

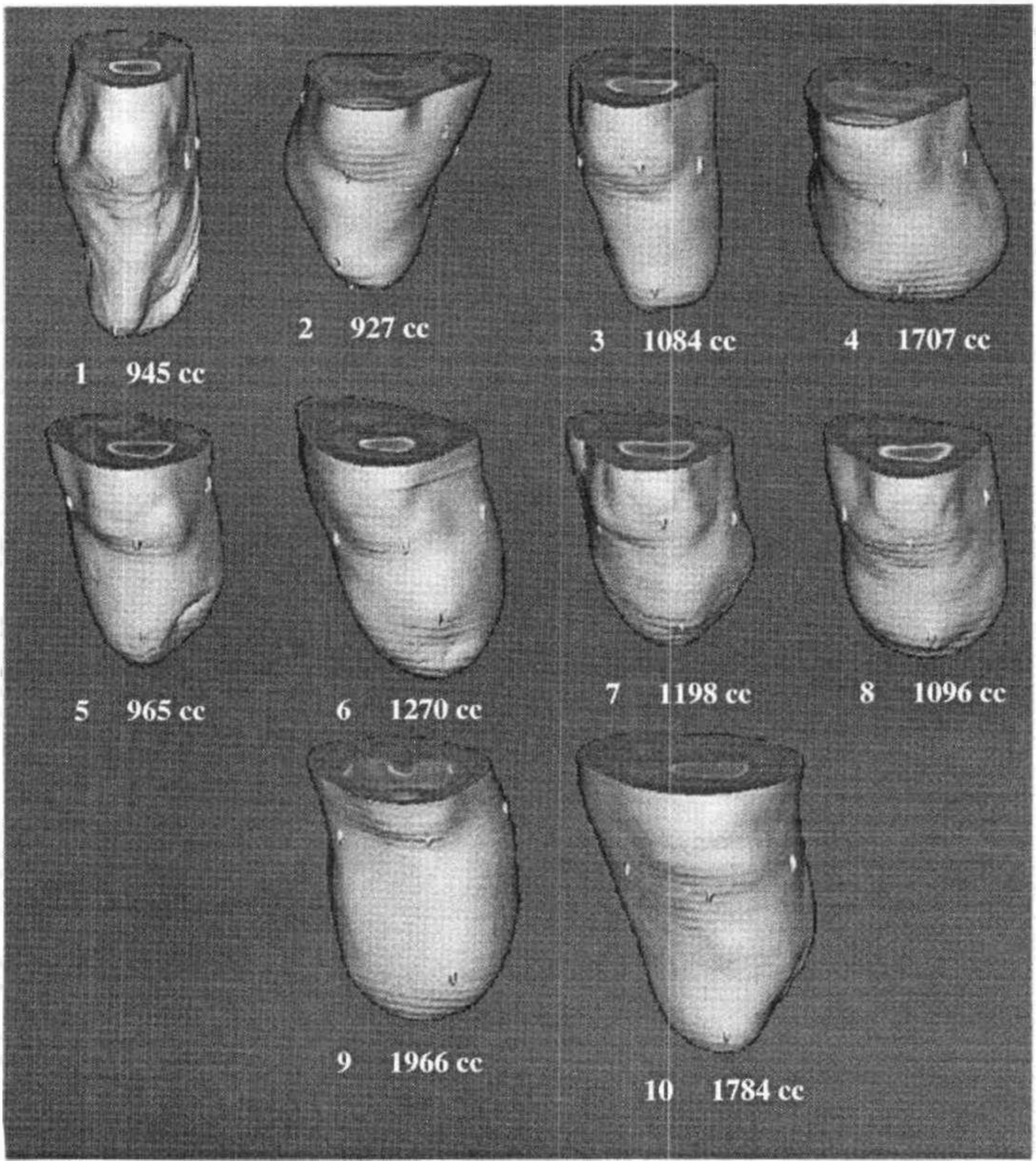


Fig. 2. Spiral X-ray Computed Tomography (SXCT) volume renderings of lower limb trans-tibial amputee volunteers. The volumes shown in the figure correspond to a volume of interest that extends from the distal end to an imaginary cutting plane through the three proximal fiducial landmarks.

placement and to test response stability over time. The hydrostatic weighing method was repeated four times per subject during each visit to assess precision error.

Table 2 details the cumulative precision error (percent) for each step in the volume measurement process for each modality. For example, random error at the segmentation level (OSS and SXCT) relates directly to intra-observer ability to align an imaginary cutting plane through the three proximal fiducial marks (which defined the volume of interest) together with any computer errors in volume determination; error at the instrument level includes errors at both the segmentation level and the instrument level (error due to instrument); error at the fiducial landmark placement level includes errors at the

Table 2. Method error (precision) as a percentage of the mean for the three methods

| | | Hydrostatic Weighing Method Error % | | Optical Surface Scanning (OSS) Method Error% | | Spiral CT (SXCT) Method Error % | |
|---|---|---|---|---|---|---|---|
| LEVEL | n | Subject | Cast | Subject | Cast | Subject | Cast |
| Segment | 28 | N/A | N/A | 0.5 | 0.5 | 0.8 | 0.8 |
| Instrument | 14 | 1.1 | 0.3 | 2.2 | 0.5 | 0.5 | 0.9 |
| Fiducial | 7 | 10.0 | 10.8 | 9.8 | 10.5 | 10.9 | 12.3 |

segmentation level, instrument level, and landmark placement level (ability to place landmarks between subject visits).

At the segmentation level, difference scores are analyzed to determine effects of the segmentation procedure on resultant scan volume. All pairs (n=28) of segmentations for a given scan (OSS and SXCT) were used for this determination. A random number assignement to select either the first segmentation or the second segmentation of a given scan (n=14) was used when determining effects of the method on volume determination. For the hydrostatic method 2 of the 4 recordings were randomly selected for analysis. Using averages of the two pairs would have resulted in a greater precision than a single number and using both pairs would be inappropriate due to a high correlation between scans (landmarks were not moved between scans). Comparing between measurement sessions (fiducial landmark level) similar consideration was given, however, only 1 of the 4 values was randomly selected (n=7). This was necessary to prevent bias due to correlated data pairs.

At the segmentation level the computerized segmentations of a single scan for the OSS and SXCT methods had 0.5% and 0.8% precision error respectively for both subject and cast data. At the instrument level for subject data there was a 1.1% precision error for the hydrostatic weighing, a 2.2% precision error for OSS, and a 0.5% precision error for the SXCT. At the instrument level for casts there was a 0.3% precision error for hydrostatic weighing, 0.5% precision error for OSS, and 0.9% precision error for the SXCT. The precision error for subjects at the fiducial landmark placement level was 10% for hydrostatic weighing, 10% for OSS, and 11% for SXCT. At the fiducial landmark placement level for the casts the precision error was 11% for hydrostatic weighing, 11% for OSS and 12% for SXCT.

A paired T-test showed a significant difference ($0.001>p>0.0005$) between scan 1 and scan 2 (instrument level) for subjects scanned in the OSS during a given measurement session. This test was necessary as the method error statistic does not detect systematic error between measurements. For other measurements no significant difference between pairs was found.

A paired T-test is inappropriate when comparing more than two group means, therefore, a single factor repeated measures analysis of variance (ANOVA) was performed on the cast data and subject data to test for significant differences in mean volume measurements between modalities. ANOVA was performed at the instrument level with the independent variable being measurement modality with three levels, hydrostatic weighing, OSS, and SXCT. A random number assignment was used to select either the first or second segmentation of a scan to prevent statistical bias due to correlated pairs of data. Tables 3 and 4 show the results of the ANOVA.

Having found a significant difference among methods, it was necessary to perform a multiple comparison test (Tukey's Honestly Significant Difference (HSD) (Portney, 1993) test for repeated measures) to determine which methods were significantly different. For the cast data a significant difference was found ($p<0.05$) between the hydrostatic method and OSS, and between the OSS and SXCT methods. No significant difference ($p>0.05$) was found between the hydrostatic and SXCT methods. For the subject data, a significant difference ($p<0.05$) was found between both hydrostatic and OSS and hydrostatic and SXCT, but not between OSS and SXCT methods ($p>0.05$).

Table 3. Repeated measures analysis of variance results between volumetry methods
Analysis of Variance on Casts

| Source of Variance | df | SS | MS | F-Ratio | p |
|---|---|---|---|---|---|
| Casts (C) | 13 | 1644092 | 126469 | – | |
| Methods (M) | 2 | 13567 | 6784 | 10.11 | p<.001 |
| Error (C x M) | 26 | 17442 | 671 | | |
| Total | 41 | 1675101 | | | |

Table 4. Repeated measures analysis of variance results between volumetry methods
Analysis of Variance on Subjects

| Source of Variance | df | SS | MS | F-Ratio | p |
|---|---|---|---|---|---|
| Subjects (S) | 13 | 1619478 | 124575 | – | |
| Methods (M) | 2 | 32671 | 16335 | 7.46 | p<.01 |
| Error (S x M) | 26 | 56913 | 2189 | | |
| Total | 41 | | | | |

Table 5. Multiple comparison test between cast group means showed the OSS to be significantly different from both SXCT and hydrostatic weighing methods.

Tukey HSD Multiple Comparison Test (Casts)
Table of Mean Differences

| | OSS | | | SXCT | | |
|---|---|---|---|---|---|---|
| | Mean Diff. (cc) | Min. Sig. Diff. (cc) | p<.05 | Mean Diff. (cc) | Min. Sig Diff. (cc) | p<.05 |
| Hydrostatic | 43.07 | 24.61 | Y | 13.64 | 23.12 | N |
| OSS | – | – | – | 29.43 | 14.35 | Y |

Table 6. Multiple comparison test between subject group means showed the OSS and SXCT to be significantly different from the hydrostatic weighing method, presumably due to short term swelling.

Tukey HSD Multiple Comparison Test (Subjects)
Table of Mean Differences

| | OSS | | | SXCT | | |
|---|---|---|---|---|---|---|
| | Mean Diff. (cc) | Min. Sig. Diff. (cc) | p<.05 | Mean Diff. (cc) | Min. Sig Diff. (cc) | p<.05 |
| Hydrostatic | 58.21 | 44.61 | Y | 60.07 | 29.09 | Y |
| OSS | – | – | – | 1.86 | 39.45 | N |

These findings are shown in Tables 5 and 6.

## Discussion

It may be seen from Table 2 that the predominant source of precision error on both casts and subjects for any of the three volumetry methods is attributable to fiducial placement between patient visits. This is explained by the inability of the observer to repeatedly place fiducial marks at identical locations between visits. The mid-patellar tendon was the only anatomic location where a fiducial mark was placed. The placement of the remaining two proximal marks was determined by the angle of the stumps in the water bath together with a subjective placement by the observer at approximately 120° posteriorly from the mid-patellar mark. While the precision of this variance was 10% or more, the mean volume of the repeat visits were nearly identical for all methods indicating no bias between visits. This further suggests the variance in volume was due to fiducial mark placement rather than a true change in the sample population over time. In fact, the sample group was chosen to contain mature stumps to reduce volumetric change over time.

At the instrument level the three volumetry methods were directly compared. Both hydrostatic weighing and OSS methods had a larger variance for subject volume measurements than for cast volume measurements. In hydrostatic weighing, patient instability presumably caused the larger error variance in volumes. When measuring the cast volume with the hydrostatic weighing method, more time and control was available in taking the measurement, thus reducing the precision error. During scanning in the OSS the cast orientation was fixed in a holding device, however, subject scanning required the patients to hold their stumps motionless in air and to assume a new orientation for each scan. For the SXCT method, surprisingly the casts had a higher precision error than subject stumps. With the SXCT method, neither the subject's stump nor the plaster casts were moved between scans and both were rested on a table during the scan procedure.

To compare between similar scans of a given measurement session a paired T-test was performed for the hydrostatic, OSS, and SXCT methods. A significant difference was found with the OSS segmentation of a single scan for cast data. The most reasonable explanation for this error is intra-observer error (i.e. a systematic error was made in segmenting the data).

Since the true volume of the stump was not accurately known, a repeated measures analysis of variance was performed, followed by a Tukey's HSD multiple comparison test to establish equivalence of methods. The ANOVA on both the cast data and subject data showed a significant difference between methods. Only the cast data was examined for equivalence of methods as that represents a non-changing standard. Lower limb stumps are in a dynamic state and subject to short term swelling that would represent a real effect and not a bias. The multiple comparison test for the cast data showed the hydrostatic weighing and SXCT methods were not significantly different ($p>0.05$), however, it showed the OSS method to be significantly different from both the hydrostatic weighing and SXCT methods. As hydrostatic weighing is an accepted form of volume measurement, it appears that the OSS has a bias or systematic error associated with it. When measuring lower limb stump volumes, comparisons should be avoided between hydrostatic weighing and OSS and between SXCT and OSS. Intermodality comparisons between hydrostatic weighing and SXCT would be acceptable.

When examining subject data the multiple comparison test showed the OSS and SXCT were both significantly different ($p<0.05$) from the hydrostatic weighing method. It is believed this is a real effect caused by short term swelling of the subject's stump over the duration of the measurement session. The hydrostatic weighing was performed immediately upon removal of the prosthesis followed within thirty to forty-five minutes by the OSS session. The SXCT examination was performed with a time lapse of approximately four hours from hydrostatic weighing. Short term swelling is most rapid upon removal of the prosthesis as the fluids pumped from the stump during ambulation return.

The SXCT precision errors may be reduced by implementing metal artifact reduction software during reconstruction or using smaller fiducial markers to reduce the metal artifact component. The precision error of the hydrostatic weighing and OSS methods may be reduced by providing the subject with better support during measurement.

Of the three volumetry methods examined SXCT is the only one capable of measuring stump volume with the prosthesis in situ. The ability to examine quantitatively sub-surface tissues in relation to the external prosthetic socket in a three-dimensional format offers to provide information on prosthetic loading distributions previously unobtainable. The volumetric data obtained with the SXCT can be saved and analyzed in many ways. The data can be sectioned into discrete slices or predefined regions for specific volumetric analysis and/or comparison to the socket volume or volume of the stump with no prosthesis. Figure 3 shows a SXCT digital projection radiograph of a trans-tibial stump with the prosthesis in place. The area covered by the two-dimensional computed radiograph is digitized and reconstructed into three dimensional data for subsequent analysis.

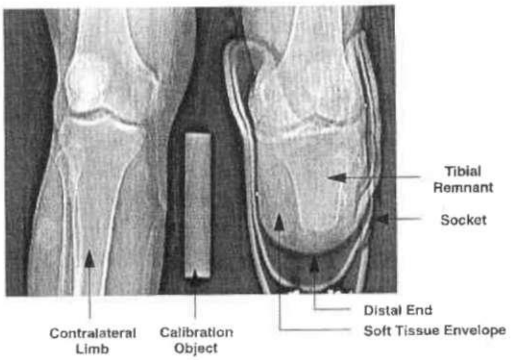


Fig. 3. Digital Computed Radiograph of a left trans-tibial stump of a female amputee scanned with prosthesis in place in the Spiral XCT scanner. This is a projection image (not a slice).

Hydrostatic weighing provides a low cost approach to limb volumetry. The process, however, is cumbersome and volumetric measurement with prosthesis in place is not possible. In addition, there are concerns regarding water borne infections when subjects have open wounds due to ulceration or other irritation. Optical Surface Scanning (OSS) would be preferred to hydrostatic weighing, but is still limited to volume measurement bounded by the external skin envelope.

## Conclusion

SXCT, OSS, and hydrostatic weighing are practical methodologies for volume measurement of trans-tibial stumps. SXCT and OSS require computer processing, but are non-contact methods and are less strenuous for the subject than hydrostatic weighing. SXCT and OSS data can be saved and regional volumes examined at later dates. SXCT offers a means to conduct detailed long term atrophy and short term oedema studies, through data set registration of unchanged bony surfaces. The ability to capture quantitative subsurface information (SXCT) of trans-tibial stumps, with the prosthesis in situ, offers a needed tool for biomechanical assessment of prosthetic fit. OSS provides a low cost tool for short term volume assessment when reference markers can be placed for registration, however, OSS was shown to have a bias when compared to SXCT and hydrostatic weighing volumetry methods and should be considered prior to comparing results. Longitudinal volumetric study of trans-tibial stumps, using fiducial markers on anatomic landmarks for registration, cannot be considered reliable.

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# The influence of orthosis stiffness on paraplegic ambulation and its implications for functional electrical stimulation (FES) walking systems

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**Abstract**

This study examines the evidence which supports the importance of maintaining relative abduction for effective reciprocal walking in high level paraplegic patients. In comparisons of orthoses, where this can only be achieved mechanically, those with higher lateral rigidity consistently showed greater levels of walking efficiency. The influence on hybrid systems of functional electrical stimulation (FES) of the gluteal muscles, where the primary function is to maintain abduction, also showed reductions in overall energy cost, reductions in upper limb effort, or both.

Examination of the effect of increasing lateral rigidity of a purely mechanical orthosis by 10% showed that significant energy cost reductions were achieved (30% reduction in Physiological Cost Index) for patients with thoracic lesions experienced in reciprocal walking.

A review of FES research suggested that for the modern healthcare sector the cost effectiveness of purely mechanical systems make them an attractive means of routinely providing the functional and therapeutic benefits of walking for high level paraplegic patients. In the prevailing climate of strict budgetry control a case is made for concentrating more research resources on improving still further walking efficiency, and resolving the outstanding problems of functionality and cosmesis in such systems for reciprocal walking.

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## Introduction

A widely held view that walking for paraplegic patients is worthwhile (Carroll, 1974; Rose, 1976; Menelaus, 1987) has led to steady improvements in performance of standard mechanical orthoses (Douglas *et al.*, 1983; Butler and Major, 1987; Campbell, 1989; Kirtley, 1992; Motloch, 1992; Lissens *et al.*, 1993). The social and economic value of these developments has been underpinned by research which shows that ambulatory paraplegic patients have half the number of bone fractures and one fifth the number of pressure sores (Mazur *et al.*, 1989). It has been demonstrated that when an orthosis is supplied in a fully controlled clinical environment the majority of patients with thoracic lesions will continue to use the device on a long term basis (Moore and Stallard, 1991). Nevertheless there remains concern that the physical effort involved in walking and the cumbersome nature of the devices required for this purpose deter many patients who could potentially benefit from this activity.

Ambitions to achieve further improvement are manifested in research and development in the separate areas of FES, mechanical orthoses, and also in combinations of these in what are termed hybrid devices.

Each of these approaches appears to have advantages and disadvantages, and all of them require appropriate compromises within the context of paraplegic ambulation (Stallard *et al.*, 1989). However, as research progresses some of the features essential to effective ambulation have become clearer. Lateral stiffness of a mechanical orthosis to enable swing leg clearance to occur easily is amongst the most important of these. The research and clinical experience of this feature has important lessons not only for future mechanical developments but also for FES research.

## Lateral rigidity in orthotic reciprocal walking

A requirement of effective reciprocal ambulation is that relative abduction of the hips is maintained (Rose, 1979; Stallard *et al.*, 1986). The ORLAU ParaWalker (Butler and Major, 1987) has been designed to achieve this by providing a high degree of lateral rigidity to permit easy clearance of the swing leg so that patients with complete thoracic lesions can walk with crutches.

To establish the effectiveness of this mechanical approach patient reviews and energy cost studies have been undertaken on the ORLAU ParaWalker system (Fig. 1), as supplied through clinical teams appropriately trained in the relevant techniques (Stallard *et al.*, 1991; Nene, 1993). These, together with other independent studies (Jefferson and Whittle, 1990; Banta et al., 1991; Bowker *et al.*, 1992; Philips *et al.*, 1995) have shown it to be an efficient device relative to other orthoses in the context of current routine practice.

Fig. 1. Patient with mid-thoracic lesion walking reciprocally with crutches in the ORLAU ParaWalker.

Straightforward mechanical analysis of an orthotic structure shows that it can be up to sixteen times more difficult to maintain abduction in adult patients than it is with children (Stallard and Major, 1993). This indicates the potential, even with the most rigid orthotic structures, for either improving mechanical stiffness or supplementing it with functional electrical stimulation of appropriate muscles to provide additional musculo-skeletal support within the structure.

## Functional electrical stimulation (FES) walking

FES has been used successfully by paraplegic patients with complete thoracic lesions in controlled environments (Kralj and Grobelnik 1973; Marsolais and Kobetic, 1987; Frech, 1987; Kralj and Badj, 1989). However, concern about the safety of such systems with their present degree of control, the energy requirement and the need to use walking frames rather than crutches has led to the development of hybrid systems in which FES has been used to supplement the function of mechanical orthoses (Petrofsky *et al.*, 1985; Andrews, 1986; McClelland *et al.*, 1987; Vorsteveld *et al.*, 1992).

A common approach has been to base a hybrid system on one of the mechanical orthoses now in routine service which enable paraplegic patients to walk reciprocally, using either a walking frame or crutches. The ORLAU ParaWalker was utilised as the basis of a hybrid system for development purposes. Results of research on that system have enabled comparisons to be made with similar hybrid approaches to walking and lead to some generalised conclusions which have important implications for further work on FES walking.

## ORLAU hybrid system research

Experimental work on six patients has been undertaken in ORLAU to establish the degree to which stimulation of the glutei on the stance side can compensate for the decreasing rigidity of the ParaWalker for adult patients.

The system consisted of a standard

ParaWalker in which the patient had been trained to walk reciprocally with crutches (Stallard *et al.*, 1986). Twin channel stimulation which delivered 60 to 100V at 20 Hz was also applied following a period of training of the glutei and quadriceps by means of a stimulator with similar specification. Training was carried out for a minimum period of 3 months on a daily basis until adequate muscle bulk had been developed (Patrick and McClelland, 1985). Patients were then fitted and trained with the hybrid system in the laboratory.

**Monitoring the effect of the ORLAU hybrid system**

Once patients had become proficient with the new system and could adequately control the timing of gluteal stimulation on the stance side via crutch handle mounted switches, two separate performance parameters were monitored and field trials were conducted.

*Crutch impulse*

The vertical ground reaction force applied through the crutches was measured. A Kistler force platform was used to record amplitude against time, as described by McClelland *et al.* (1987). The integral of force/time is known as impulse and is directly related to the effort required of the upper limb to ambulate in a reciprocal walking device. Each patient had impulse measured using the ParaWalker alone and then, after a suitable period, in the ParaWalker supplemented by the gluteal stimulation system.

In a study of six patients (Stallard and Major, 1993) it was shown that five had reductions in impulse when stimulation of the gluteal muscles on the stance side was used, and that one patient had a marginal increase (Fig. 2). The average reduction in upper limb effort required to walk when stimulation was applied, as indicated by impulse measurements, was 18.7% (including the patient in whom a small increase was recorded).

*Energy cost*

Five patients had their overall energy cost of walking monitored using oxygen uptake (Nene and Patrick, 1990). In four out of the five a decrease in energy cost of between 6% and 9% was recorded when patient controlled stimulation of the gluteal muscles on the stance side was used. The fifth patient recorded no change in energy cost. The patient in which no change occurred was the one where a marginal increase in crutch impulse was noted when FES was used.

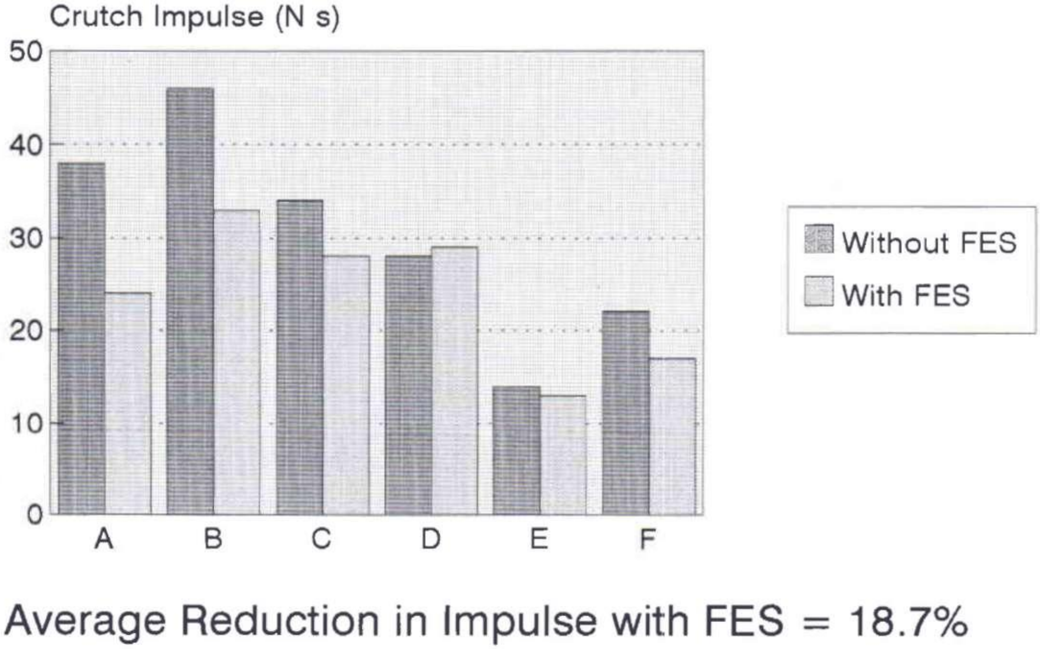


Fig. 2. Crutch impulse for six thoracic lesion patients using the ORLAU Standard and Hybrid ParaWalker systems.

*Field trials*

Following laboratory trials with the hybrid system patients were invited to try it at home and to comment on its effectiveness in general use. A common series of problems was reported by patients. These were:

1. difficulty of accurate electrode placement;
2. inconsistent stimulation;
3. the extra time it took to apply the system;
4. cross-stimulation of abdominal wall muscles when stimulation amplitude was high enough to be effective in the field environment.

All patients eventually elected not to use the option FES system and went on to use the purely mechanical ParaWalker system with what they considered adequate effectiveness.

**Measuring the effect of lateral stiffness**

Whilst the circumstantial evidence supports the importance of lateral stiffness in walking orthoses for paraplegic patients, a number of difficulties have previously prevented direct comparisons of the effect of increasing orthotic stiffness on walking efficiency. The introduction of a modified ORLAU ParaWalker '89 hip joint which has 70% greater lateral rigidity (Stallard and Major, 1992) provided an opportunity to monitor changes in patients for whom a replacement orthosis was required. Measurements of orthosis rigidity showed that the new, stiffer hip joint provided an overall increase in lateral orthosis stiffness of 10% (Stallard and Major, 1993).

**Patients**

The ORLAU ParaWalker '89 was supplied to patients for whom a new orthosis was required after a period of several years in the original ParaWalker design and where circumstances also permitted pre and post supply energy cost monitoring. Details of the three patients are shown in Table 1.

Table 1. Patient details

| Patient | Sex | Age | Level of Lesion |
|---|---|---|---|
| A | M | 28 | T8/9 Comp. |
| B | M | 31 | T12/L1 Comp. |
| C | M | 30 | T11 Comp. |

**Method**

Relative energy cost was measured using Physiological Cost Index (PCI) (Butler *et al.*, 1984). This represents the difference in heart rate between resting and walking divided by speed to give units of heart beats per metre walked i.e.

$$\text{PCI(bts/m)} = \frac{\text{walking heart rate} - \text{resting heart rate}}{\text{walking speed}}$$

This means of monitoring relative energy cost has been validated in a variety of studies (Rose *et al.*, 1991; Nene, 1993). The precise methodology of monitoring the three patients is that adopted by Nene and Jennings (1989) who established PCI for paraplegic patients using the ORLAU ParaWalker. Telemetered ECG was processed to give patient heart rates whilst resting and then walking over five timed 6.1m runs with one minute rest between each. The PCI for each of the five runs was then averaged to give overall PCI.

**Results**

Figure 3 shows a histogram of the results for the three patients. From this it can be seen that PCI was significantly reduced in each case when using the ParaWalker '89 device and that the average PCI was changed from 1.4 to 0.98, a reduction of 0.42, or 30%.

**Discussion**

The evidence that maintaining relative abduction in the lower limbs is one of the most crucial elements of ensuring effective reciprocal walking for high lesion paraplegic patients is becoming ever more irrefutable.

Results reported in this paper of relative energy comparisons in patients who used both the original ParaWalker and the ParaWalker '89 (which is 10% stiffer) provide direct confirmation of the importance of lateral rigidity in walking orthoses for paraplegic patients. Although, because of the mechanical reliability of the ParaWalker together with ethical, logistical and practical difficulties this was done on only three subjects, it confirmed previous compelling circumstantial evidence

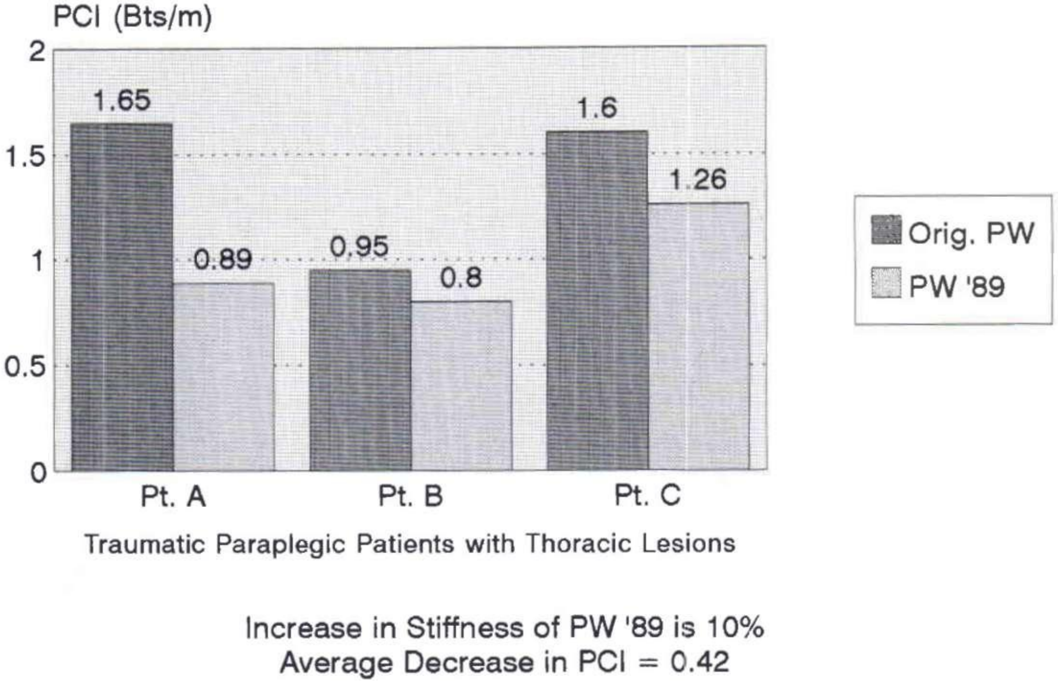


Fig. 3. PCI of three thoracic lesion patients using the Original and '89 ParaWalkers.

supporting the need to maintain relative hip abduction in walking for high lesion paraplegics.

An analysis of the results of research on the ORLAU Hybrid (FES/ParaWalker) System (McClelland *et al.*, 1987; Stallard and Major, 1993) shows that the major contribution of the gluteal stimulation is to ease swing leg clearance by increasing overall lateral rigidity in the patient/orthosis complex, thereby reducing upper limb effort. This results in reductions in overall energy cost (Nene and Patrick, 1990). The added burden of applying the additional FES apparatus led to all of the patients in the trial discarding this element for routine walking in their normal environment.

A study by Alfieri and Marchetti (1993) on a system for paraplegic patients incorporating the Louisiana University Reciprocating Gait Orthosis and FES produced an outcome in which upper limb effort was reduced, but overall energy cost stayed the same. Whilst fatigue in the arms was reduced, this system did not produce the overall energy cost savings seen in the ORLAU system. However, patients tended to retain FES for separate use to relieve spasticity, whilst discarding it for walking.

The common features of these experiences suggest that whilst FES via surface electrodes can be used to supplement mechanical orthoses and effectively reduce upper limb effort, the overall savings in energy cost are either small or non-existent. It is to be expected that energy cost savings will not match the reduction in upper limb effort since the metabolic costs of the stimulated muscle must be taken into account. Patients' unwillingness to tolerate the additional inconvenience of applying electrodes and carrying additional electronic apparatus mirrors the experience of Rushton (1992) in this regard. Healthcare purchasers question the additional cost of hybrid systems and are reluctant in many cases to sanction the necessary expenditure. Performance results suggest that this reluctance has some justification.

A single patient study involving the ORLAU ParaWalker and an implanted FES system (Nene and Jennings, 1989) produced similar patient performance results but greatly increased the cost and initial inconvenience because of the need for implantation surgery. Further patients have not been tempted by that option.

The use of FES to supplement mechanical orthoses has served to emphasise the importance of maintaining hip abduction if effective walking is to be achieved by paraplegic patients. Whilst the clinical experience suggests that the inconveniences of supplementary FES for walking generally outweigh the advantages of improved performance, there are important lessons to be learned from research in this area:

1. implanted systems will be necessary for FES to be acceptable;
2. for the additional costs and inconvenience of hybrid systems to be justified, greater improvements in performance than those so far achieved will be necessary;
3. pure FES systems will need to address the problem of producing powerful and sustained hip abduction via the gluteal muscles if they are to permit effective and acceptable walking with crutches for patients with complete thoracic spinal lesions.

Unless these problems are addressed it seems likely that FES systems will require to use walking frames for the majority of patients, rather than the more acceptable and practical option of crutches which the most effective mechanical orthotic systems permit.

For hybrid FES systems there would appear to be a case for continuing to control abduction mechanically so as to release the FES system to improve gait in other ways – for example by providing knee flexion in swing phase (Vosterveld *et al.*, 1992).

The new evidence in this paper linking orthosis stiffness directly to walking efficiency emphasises still further the importance of structural properties in mechanical ambulation devices. It is clear that pure mechanical orthoses will remain the most cost effective solution to walking for thoracic lesion patients into the foreseeable future. In a situation where cost benefit analyses will have increasing importance in the health care sector this would strongly suggest that further research and development of mechanical solutions to permit increased walking efficiency, cosmesis and convenience in doffing and donning is now more justified than ever.

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# Modified fracture brace for tibial fracture with varus angulation: a case report

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**Abstract**

Sarmiento introduced the functional fracture brace for the management of tibial shaft fracture in 1963. However, tibial angulation with varus deformity cannot be prevented or corrected by such a device. In this paper, a case of tibial shaft fracture with varus angulation treated with a modified below-knee fracture brace was reported.

## Introduction

The goals of management in tibial shaft fracture are to obtain union, to preserve ankle and knee motion, and to maintain alignment. Management of tibial shaft fracture by functional fracture bracing has been known for many years (Sarmiento, 1967 and 1970; Sarmiento *et al.,* 1980). However, varus angulation is a frequent occurrence especially in the presence of an intact fibula (Sarmiento *et al.,* 1989; Teitz *et al.,* 1980).

In the management of tibial fracture, non-operative treatment may offer the advantage of low incidence of infection (Sarmiento *et al.,* 1989). In 1963, Sarmiento introduced the functional cast for the management of fracture of the tibial shaft (Sarmiento *et al.,* 1967), and eventually modified the procedure to the use of a prefabricated functional fracture brace. His serial reports showed such bracing was an effective alternative for treatment of selected fractures of the tibia (Sarmiento, 1970; Sarmiento *et al.,* 1984; Sarmiento *et al.,* 1989). A fracture brace of such type is particularly well suited for the weather in Taiwan. Many patients with a tibial shaft fracture treated by cast fixation developed skin problems due to humid weather. However, shortening of the injured limb and the tendency for tibial angulation may still occur, especially with an intact fibula. This device cannot prevent or correct tibial angulation. Moreover, such complications were also observed by other investigators. Some researchers even concluded that single tibial fractures with an intact fibula presented an insidiously dangerous fracture pattern, particularly in patients more than twenty years old (Teitz *et al.,* 1980). When tibial angulation occurs during bracing, the brace should be discontinued and replaced by casting. Sometimes, tibial angulation should be corrected by surgical intervention (Bohler, 1965; Fernadez-Palazzi, 1969; Lottes, 1965; Sharma, 1972; Sorensen, 1969).

## Case Presentation

A 30-year-old woman fell down accidentally in downhill skiing on December 25, 1990. She was found to have a closed comminuted fracture at the lower third of the right tibial shaft with medial malleolar fracture (Figs. 1 and 2). A temporary long-leg cast was given for travelling, and changed to a functional fracture brace of Orfit material on the fifth day after injury. However, a varus angulation of 10° in the right tibia was noted ten days later after bracing (Fig. 3). To reduce and to maintain the

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Fig. 1. Picture showing patient's right tibial shaft with closed comminuted fracture and medial malleolar fracture.

Fig. 3. Picture of x-ray showing varus angulation (10°) in the right tibia ten days after fitting with conventional below-knee fracture brace.

Fig. 2. Close-up view of right medial malleolar fracture.

alignment of the tibia, a modification in moulding technique of the fracture brace was introduced.

## Modification

A wooden rod (1.5cm in diameter, 23cm in length, 45gm in weight) was embedded at the under-side of the anterior shield of the brace, just between the tibia and fibula. Trimming of the rod was undertaken for comfortable fitting. The ankle was fixed in the neutral position for the medial malleolar fracture. The brace was moulded directly to the right lower leg using thermoplastic material (Orfit, Luxilon, 3.2mm in thickness, 55-60°C in moulding temperature) after the rod was fixed snugly between the tibia and fibula by elastic bandage. The tightness of the brace was adjusted by five velcro straps at different positions (Figs. 4, 5 and 6).

## Results

Roentgenographic examination showed a significant immediate correction of the varus angulation of the tibia to less than 5° by fitting of this modified fracture brace (Fig. 7). The

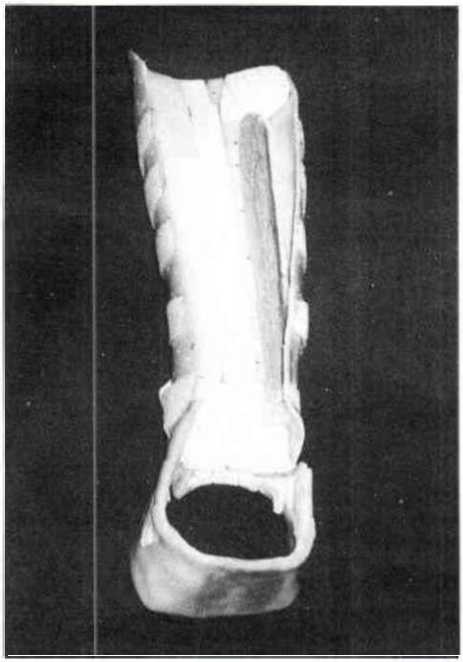

Fig. 4. Modified fracture brace was made by embedding a wooden rod at the underside of the anterior shield between the tibia and fibula.

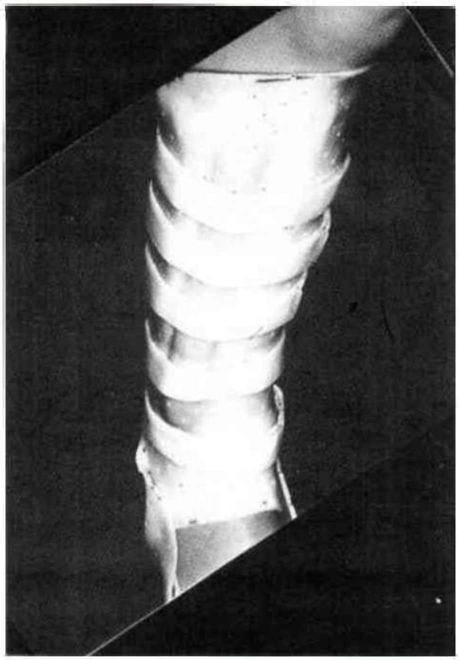

Fig. 6. The tightness of the brace was adjusted by five velcro straps.

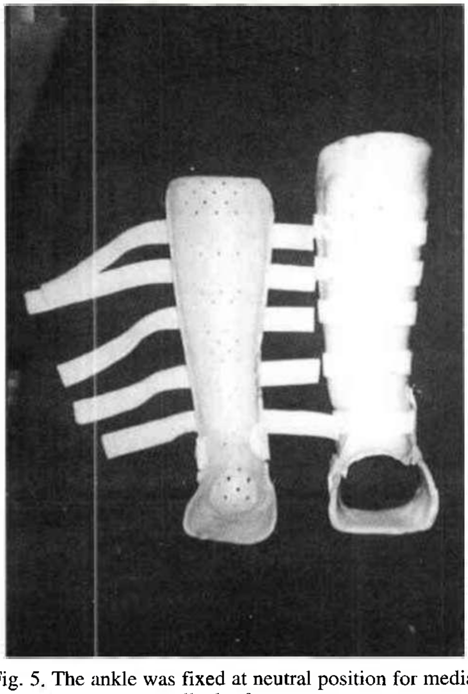

Fig. 5. The ankle was fixed at neutral position for medial malleolar fracture.

patient walked with axillary crutches with non-weight bearing of the right leg until the sixth week. Satisfactory bridging callus formation with sound alignment of the tibia was found in X-ray film at the twentieth week (Fig. 8), and the patient could walk with full weight bearing on the leg without pain. At the thirty-second week, the fracture brace was discarded. She had full range of motion at the right knee and ankle. Mild to moderate degree of muscle atrophy in the right leg was noted, but still with good muscle power. Radiographs taken one year after of the right tibia showed nearly complete bone healing with no late varus deformity (<5°) at the fracture site (Fig. 9). All joints of the right lower limb had full range of motion and no shortening of leg length was noted.

## Discussion

In this paper, the patient developed a varus deformity after fitting with conventional fracture brace for ten days. However, the angulation was successfully controlled by the modified fracture brace with satisfactory maintenance of alignment. The time course for subsequent bone healing and clinical recovery

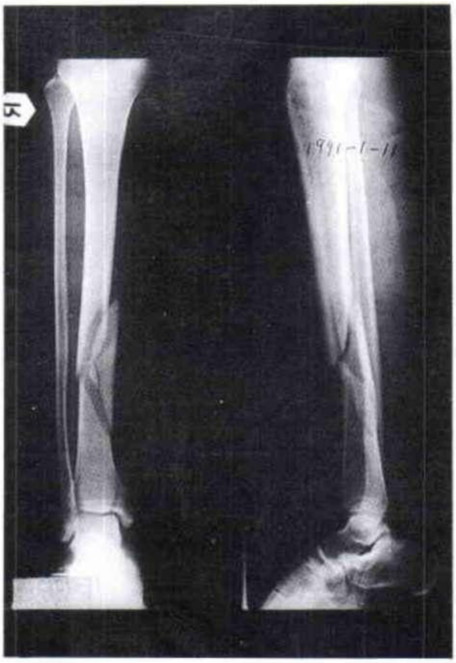

Fig. 7. Roentgenographic examination showed a good correction of the varus angulation of the right tibia (5°) immediately after fitting of the modified fracture brace.

Fig. 8. Picture showing bridging callus formation with sound alignment at twentieth week.

Fig. 9. Picture showing near complete healing at one year.

for the patient was consistent with those results reported by Sarmiento in 1989. She ambulated well consequently without any significant sequelae.

The modified fracture brace proposed in this paper was developed from the authors' successful applications of plastic anterior ankle foot orthoses (AFO) (Wong *et al.,* 1992). The fracture brace is the combination of the anterior and posterior leaf AFO (Teufel style orthosis) together as one unit (Lehmann *et al.,* 1985). The length of that brace anteriorly and posteriorly is similar to that of the patellar tendon-bearing othosis (Fishman *et al.,* 1985). This design might conform to the hydraulic principles which prevent shortening of the injured limb (Sarmiento, 1970). The embedding of the light wooden rod at the lateral underside of the anterior leaf AFO provides a constant transverse strain onto the varus angulation. It is believed that it was this straining effect that prevented the recurrence of varus deformity.

Although satisfactory results were achieved in this case with the modified fracture brace. Further investigation is needed to prove the consistency of the result.

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# Technical note
# Design and manufacture of a high performance water-ski seating system for use by an individual with bilateral trans-femoral amputations

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**Abstract**
A high performance seating unit has been produced for a water-skier with bilateral trans-femoral amputations. The system, in which the user sits whilst skiing, has helped the client to further her sporting career and enabled her to compete successfully at the highest levels of disabled water-skiing competition.

**Introduction**

The Clinical Engineering Group at Withington Hospital was asked to produce a high performance seating unit to fit onto a standard Kan-Ski water-ski (designed for use by paraplegics) for use by an individual with bilateral trans-femoral amputations. The client involved is a highly successful, competitive disabled water-skier (current UK and World Disabled water skiing champion), who was experiencing difficulties with the equipment which she was using. She felt the technique which she had adopted in order to use this equipment was both unsafe and inefficient and that if she was to remain competitve at a high level, then alternative equipment configurations would have to be sought.

Before the authors became involved in the project, the client had tried various equipment combinations, the latest being the commercially available Kan-Ski water-ski and seating system. The Kan-Ski is a high performance carbon fibre water-ski with dedicated aluminium frame seating system, designed for use by paraplegic water skiers. The seating system can be used by individuals with high level lesions as it offers a high degree of lateral support around the thorax. Unfortunately, this system proved to be unsatisfactory for this client as the seat could not be used as was intended by the manufacturer. She had to sit directly on the board, effectively trapping herself in the framework. As well as being quite unsafe, this seating position required a lot of upper body strength to manoeuvre the board, due to the proximity of the skier's centre of gravity to water level.

The task was to design and construct a completely new seating system that would easily interface with the preferred performance water-ski (the Kan-Ski) and allow complete safety combined with high level performance.

**Design requirements**

The first stage of the design process was to specify fully what was actually required by the client and to establish what parameters and restrictions the project would be subject to. A design specification was drawn up and issued to all parties concerned with the project for comments and queries. The specification included:

- geometry: client physical measurements, required seat height above water, dimensions of fixation lugs on Kan-Ski, strapping requirements etc.;
- forces: estimations of the forces experienced by the system when being towed around a

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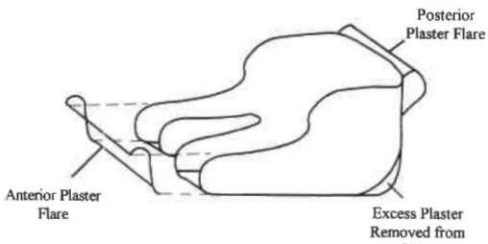


Fig. 1. Rectified seat mould

slalom course at speeds up to 58 km/h;
- materials: corrosion and safety implications (methods of failure);
- safety: getting out of the seat in an emergency;
- maintenance and cost factors.

Viewing the specification, it became apparent that the problem would be best approached by splitting it into two separate parts; i) a custom built seat (manufactured using standard orthotic techniques); ii) some means of supporting the seat and fixing it to the ski.

**Production of dedicated seating unit**

The client was cast in a "dry suit", which she normally wore when water-skiing (the suit was protected from the plaster used during the casting procedure by a polymer film wrap). During casting, she was seated on a casting couch, with her stumps lying on the head board, her body being tilted to a position which she deemed most comfortable for water skiing. The stumps and weight bearing aspects of the gluteal region, were then cast using pre-measured slabs of six-ply, 200mm thick plaster. This was followed by casting of the pelvic region and the region immediately proximal to it, with a wrap cast from 200mm thick rolls of plaster. Concave profiles were then moulded by hand over each iliac crest. The cast was then removed from the patient, sealed and filled to produce a positive mould.

Rectification of the cast involved removing a maximum of 25mm of plaster at the weight-bearing/non-weight-bearing border of the gluteal region, as well as building up the region of the cast accommodating the stumps, so that the seat was the same length as its supporting frame. Flares were then added to the front and back of the seat, enabling it to locate firmly with the cross members of the supporting frame (Fig. 1).

The seat was made by moulding 3mm low density expanded polyethylene foam to the positive cast, and vacuum moulding polyethylene-polypropylene copolymer over it. This was then trimmed to shape and all sharp/rough edges removed.

During the fitting sessions, further trimming was undertaken and a region between the stumps was marked out and a high density expanded polyethylene foam "pommel" fitted therein. This incompressible pommel allowed a more intimate fit between stups and seating system and thus added to the feedback that the client received when using the system on a water-ski.

Straps made from neoprene and nylon webbing were attached to the frame to hold the client firmly in the seat. These straps were placed over each stump, crossing over through a hole in the seat immediately behind the pommel and attaching to the diagonally opposite side strut of the frame. The client was held within the seat by these straps using a Velcro fastening. This fastening could be rapidly removed if the occupant so required, hence allowing an expedient escape from the seat in emergencies.

**Production of supporting framework**

The final frame design basically consisted of two portal frames at the front and rear supported by cross-bracing members onto which the seat would attach (Fig. 2).

It was assumed the frame would be subjected to a maximum force of 3 times occupant body weight in a vertical direction and shearing force of 2 times occupant body weight in a horizontal

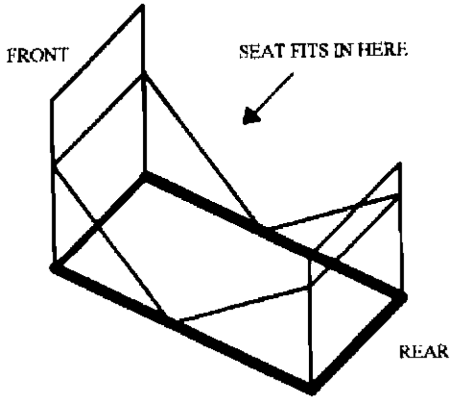


Fig. 2. Seating system framework


Fig. 3. Completed water-ski seating unit

(front to rear) direction. These estimations of the forces were calculated for the seating system when being used on a slalom course only, and not for jumping or trick skiing (a condition stipulated in the specification).

1 inch diameter 16 gauge BS1474 HE30TF aluminium alloy tubing was chosen for the main frame members as it has a high strength to weight ratio and good corrosion resistance properties. The tubing framework was welded onto a rectangular base (constructed of the same alloy but in flat bar form) which had a series of 10 vertical holes drilled into it down each side. The frame could then be placed over the lugs provided on the Kan-Ski and fixed down using wing nuts. The series of holes allowed the skier to adjust the frame to the most suitable position.

A comparison of proof stress against maximum estimated stress in the frame members was used to calculate factors of safety for the frame members. The lowest factor of safety was 4.5 for the vertical members of the rear portal frame. This was deemed acceptable as in the worst case of maximum loading scenario, the structure would show visible signs of deformation before actual failure occurred.

The hollow members of the frame were filled with polyurethane foam. This ensured improved flotation but more importantly ensured that water would not collect inside the frame, thus reducing the chances of corrosion.

## Outcome

After manufacture, the 2 components of the seating system were brought together during a fitting session with the client (Fig. 3). The system was tested for fit and comfort and the overall construction and form was discussed further, allowing both designer and client to anticipate any problems that may occur during use. When all concerned were satisfied with the system, the first prototype was issued for use.

The system was first used by the client during a training week in the USA. During this week the client regularly used the system and by the end of the week was using it at full speed on the slalom course. Some problems were experienced by the client when cracks appeared in the polyethylene-polypropylene seat and it eventually broke at the attachment point to the frame. This mode of failure had been anticipated by the design team and no injuries were sustained by the client. The client was able to reattach the seat to the frame using Nylon webbing.

On return to the UK, the client reported her experiences to the team and the whole seating system was inspected to assess any damage. Apart from the seat, the system showed no other signs of failure.

A new seat was made using a slightly thicker (4mm) polyethylene-polypropylene copolymer and new straps attached. A slight reduction in the height of the rear portal frame was also made to allow a more functional seating position.

**Conclusion**

A high performance water-ski seating system which allows the client to compete at the highest levels has been produced and has been in continual use by the client since April 1994.

No further problems have been experienced and the user is again competing successfully.

This successfully completed project has been achieved through collaboration and co-operation between the Clinical Engineering Group at Withington Hospital and the School of Prosthetics and Orthotics at Salford.

**Acknowledgement**

The authors would like to thank Mr. Stefan Jedut at Salford and Mr. Robert Williams at Withington Hospital for their assistance with this project.

Prosthetics and Orthotics International, 1995, 19, 124–127

# Technical note

## Driving appliances for upper limb amputees

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**Abstract**

The advice given to upper limb amputees in the United Kingdom with regard to the use of driving appliances has often been somewhat variable. At best a full rehabilitation package has been provided, including the fitting of the appliances to the person's vehicle and contact with the driver's instructor, to the other extreme of issuing driving appliances to patients with no instruction at all. Though upper limb amputations are not a relevant or prospective disability, all drivers with a "limb disability" are legally required, in the UK, to declare changes in their physical state to the Driver and Vehicle Licensing Agency. This study examines the current usage of driving appliances. It was found that the level of upper limb loss has little effect on where the driving appliance is placed or on any other aspect of driving method used.

### Introduction

Upper limb amputations are much less common than lower limb amputations and in most regional prosthetic centres in the UK the ratio of presentation is approximately 1 upper limb amputee to 25 lower limb amputees (Department of Health and Social Security, 1986). Though congenital deficiences are different from amputations, both in terms of the psychological effect and the prosthetic acceptance and usage, in clinical practice the prosthetic management is similar for both. For the purpose of providing appropriate appliances to assist in driving, the prosthetic management of the transverse congenital deficiency is identical to that of the amputee.

In the UK the driver and Vehicle Licensing Agency's (DVLA) licence categorically states that a driver should inform the agency if there is any physical change in his or her condition. However, a survey on diabetes and driving revealed that about a fifth of all diabetic drivers had not informed the DVLA or their motor insurers of their diabetes (Saunders, 1992). No such survey has been carried out on the amputee population. The majority of patients do drive or return to driving after upper limb amputation and are usually under the care of the multidisciplinary team at the regional or sub-regional Disablement Services Centres.

There is no comprehensive literature available to give to upper limb amputees which would advise them where to place a ball appliance on the steering wheel, how to change gear or use the hand brake, or how these details vary with level or site of amputation (UK Forum of Driving Assessment Centres, 1991[1,2]). For example should the left trans-humeral amputee place the steering ball in the same position as a right trans-humeral amputee? The advice given has often been varied and inconsistent. The authors carried out a prospective study of upper limb amputees to ascertain how these patients were currently using the driving appliances provided and whether an appropriate pattern of usage could be determined. In the following descriptions it should be remembered that in the UK vehicles drive on the left side of the road and consequently are right hand drive with the hand brake and gear change normally operated with

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the left hand. It might be presumed that in countries where left hand drive vehicles are the norm, a similar study would give results which would be a mirror image of those outlined in the following pages.

## Method

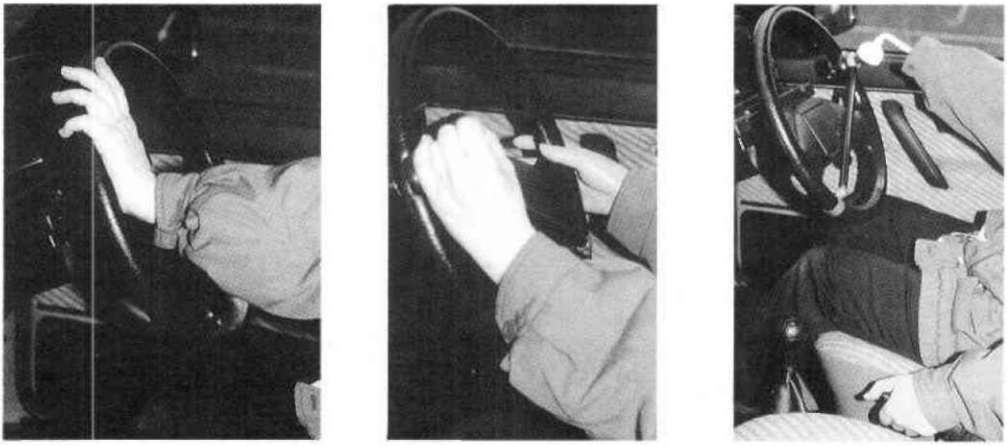
Sixty adult upper limb amputees (more than 17 years of age) consecutively attending the prosthetic clinic were asked to complete a structured questionnaire. In some cases assistance was given. Two forms were used, one asking questions specific to the left amputee and the other for the right amputee. After an initial and successful pilot study of 10 patients the study was extended to include a further 50 patients. All the questionnaires were completed during the clinic attendance itself, hence there were no non-responders.

## Results

Of the 60 questionnaires completed, 2 contained insufficient or conflicting information, and another supplied the patient's name, level and site of loss with the statement, "I do not drive". These 3 questionnaires were therefore removed from the results. Readers are reminded that driving was on the left side of the

**Table 1. Methods of steering for right amputees (34)**

| Method | Number of Patients | Comments |
|---|---|---|
| No steering ball used | 7 | Of these patients, 6 steady the steering wheel with a prosthetic hand when changing gear or using the hand brake. The other, a forequarter amputee, drives an automatic. All 7 steer by palming the wheel round with their sound hand. (See Fig. 1a). |
| Steering ball mounted on the left side of the wheel (between 9 and 10 o'clock position) | 13 | All of these patients use the ball in their sound hand. Of these 11 drive automatics. The other 2 drive manuals one steadying the wheel with a passive hand and the other with her trans-radial stump, whilst changing gear or using the hand brake (See Fig. 1b). |
| Steering ball mounted on the right of the wheel (between 12 and 5 o'clock position) | 14 | Of these 14 patients none uses the sound hand on the steering ball, except when parking. Automatics are driven by 2 of the amputees and the 12 drivers of manuals all control the wheel using a prosthetic appliance on the steering ball whilst changing gear, while steering and when operating the hand brake. (See Fig. 1c). |



(a) (b) (c)

Fig. 1. Methods of driving for right amputees.
(a). No motoring ball used.
(b) Ball mounted on the left side of the wheel (between 9 and 10 o'clock position).
(c) Ball mounted on the right side of the wheel (between 12 and 5 o'clock position).

road, and hence the vehicles were right hand drive. Of the 57 patients who responded, the number of trans-radial and trans-humeral patients, and their choice of gearbox was as follows:

| | Automatic | Manual |
|---|---|---|
| Trans-radial (41) | 12 | 29 |
| Trans-humeral (16) | 9 | 7 |

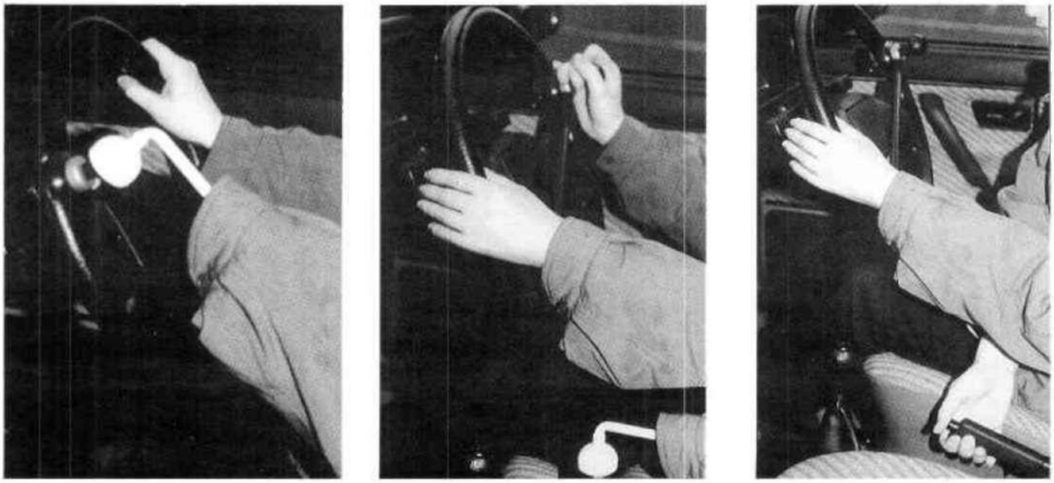
Of those amputees using an automatic gearbox, most had Automatic Driving Licences only. One of these patients, who has a shoulder disarticulation, had passed an advanced driving test and now instructs others. Only one patient was actually required by the DVLA to drive an automatic, even though he had driven previous to his amputation and was a trans-radial amputee. Of the 57 amputees the distribution of amputation by side was:

| | |
|---|---|
| Right upper limb amputee | 34 |
| Left upper limb amputee | 23 |

The methods of steering used by right and left upper limb amputees are shown in Tables 1 and 2.

**Table 2. Methods of steering for left amputees (23)**

| Method | Number of Patients | Comments |
|---|---|---|
| No steering ball used | 8 | Of these patients, 7 change gear with a passive prosthetic hand and 1 operates the autoshift with a passive prosthetic hand. All 8 steer by palming the steering wheel round with the sound hand, though some added that they also use the prosthetic hand on the wheel at all times. |
| Steering ball mounted on the left side of the wheel 10 o'clock position) | 3 | All three patients drive automatics, and 2 of them use a cup on stem appliance in the prosthesis to help steady the wheel. One added that he also uses the steering ball in the sound hand in certain circumstances, and the third patient, a trans-humeral amputee, always uses the ball in his sound hand. (See Fig. 2a). |
| Steering ball mounted on right side of the wheel (between 12 and 4 o'clock position) | 12 | All 12 patients use the steering ball in their sound hand. Of these 5 drive automatics. Of the 7 drivers of manuals, 6 use a passive prosthetic hand, and one uses a cup on stem appliance to change gear. (See Fig. 2b). All 14 drivers using manual gearbox vehicles operate the hand brake by reaching across with their sound hand, resting the prosthesis on the wheel whilst they do so (See Fig. 2c). |



(a) (b) (c)

Fig. 2. Methods of driving for left amputees.
(a) Ball mounted on left of wheel (10 o'clock position).
(b) Ball mounted on the right side of the wheel (between 12 and 4 o'clock position). Passive hand or cup on stem (inset) used to change gear.
(c) Hand brake operated by reaching across with sound hand.

**Discussion**

In this study, modifications to vehicles were found to be rare, and only involved minor changes to switch controls, such as indicators being moved from left to right of the steering wheel. It was interesting to note that the ratio of automatics to manuals is very similar for both left and right amputees.

Left upper limb amputees, when they use a steering ball, mostly set it up for use with the sound hand, (about 2 o'clock position). They change gear with the prosthetic hand or appliance. To operate the hand brake they reach across with their sound hand, resting the prosthesis on the steering wheel whilst they do so.

Right upper limb amputees driving manual gear shift vehicles set up the steering ball, if they use one, in such a position that it can be used in conjunction with an appliance in the prosthesis, between 1 o'clock and 5 o'clock position). This enables gear changing to be carried out with the sound hand.

Since many of the patients' limb deficiencies were congenital in origin, it is difficult to determine whether left or right dominance played any part in the way in which the appliances were used, but it may account for some of the methods of usage that fell outside of the patterns outlined above.

Those driving automatic vehicles mostly set up the steering ball so that it can be used with the sound hand, irrespective of the side of amputation.

**Conclusion**

The level of upper limb loss following either an amputation, or a congenital deficiency, has little effect on where the driving ball appliance is placed, or on any other aspect of the driving method used, with the exception that the higher the level of loss, the more likely it is that the patient will choose a vehicle with an automatic gear box. This seems fairly logical as neither steering nor gear changing can be easy using a prosthesis when the amputation is above the elbow level. Using an automatic gearbox frees the sound hand for steering, and obviates the need to use the gear shift whilst on the move. About two thirds (70%) of trans-humeral amputees use automatic gear box vehicles as compared to less than half (44%) of trans-radial amputees.

A quarter (26%) of the patients in this study did not use a steering ball (all of these were males) and of the remaining three quarters (74%) issued with a ball and clamp for use with a prosthesis, over half (46% of the total) use the ball in the sound hand. Since the steering ball was designed for use in conjunction with the prosthesis, it does beg the question as to whether it really is a satisfactory shape for use with a sound hand.

The question of the use of a passive prosthetic hand or a cup on stem appliance to operate the gear lever and hand brake also needs to be addressed.

With the recent introduction of car safety designs like air bags inflating from the centre of the steering wheel, in cases of impact, the inappropriateness of many current driving appliances becomes evident. New designs need to take these factors into account, in addition to the pattern of usage of the driving appliance.

Lastly it would be beneficial to have a succinct and diagrammatic instruction booklet to provide upper limb amputees with relevant updated information and advice in order to assist them in driving, thus helping to reach their optimum rehabilitation potential.

**Acknowledgement**

The authors would like to thank Mrs. J Kay-Sportelli for typing the manuscript.

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UK Forum of Driving Assessment Centres (19912). How to get behind the wheel: information for amputees wishing to drive a car.- Carshalton, Surrey: UK. Forum of Driving Assessment Centres.

# Calendar of Events

## National Centre for Training and Education in Prosthetics and Orthotics
## Short Term Courses 1995-96

### Courses for Physicians, Surgeons and Therapists

| | |
|---|---|
| NC504 | Lower Limb Orthotics; 20–24 November, 1995 |
| NC502 | Upper Limb Prosthetics and Orthotics; 27 November–1 December, 1995 |
| NC512 | Orthotic Management of the Foot; 4–5 December, 1995 |
| NC505 | Lower Limb Prosthetics; 22–26 January, 1996 |
| NC510 | Wheelchairs and Seating; 25–27 March, 1996 |
| NC217 | AFO for the Management of the CP Child; 24–26 April, 1996 |
| NC511 | Clinical Gait Analysis; 1–3 May, 1996 |
| NC506 | Fracture Bracing; 7–10 May, 1996 |

### Courses for Orthotists and Therapists

| | |
|---|---|
| NC512 | Orthotic Management of the Foot; 4–5 December, 1995 |

Further information may be obtained by contacting Professor J. Hughes, National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, Curran Building, 131 St. James' Road, Glasgow G4 0LS, Scotland. Telephone: (+44) 141 552 4400 ext. 3298, Fax: (+44) 141 552 1283, E-mail Annette.Hepburn@strath.ac.uk

**5-8 September, 1995**
2nd Leeds European Rehabilitation Conference: Neurological Rehabilitation, Leeds, England
Information: Dr. A. Cutts, Rheumatology and Rehabilitation Research Unit, The University of Leeds, 36 Clarendon Road, Leeds, LS2 9NZ, England.

**8-10 September, 1995**
4th Scientific Meeting of the Scandinavian Medical Society of Paraplegia, Oslo, Norway.
Information: Congress Secretariat, 4th Scientific Meeting of SMSOP, c/o Sunnaas Hospital, N-1450 Nesoddtangen, Norway.

**12-16 September, 1995**
10th Asia Pacific Regional Conference of Rehabilitation International, Jakarta, Bali.
Information: Secretary, 10th ASPARERI, H Hang, jebat 11-2 Blok FIV, Kebayoran Baru, Jakarta 12120, Indonesia.

**17-21 September, 1995**
Medicon '95. 7th IFMBE Mediterranean Conference on Medical and Biological Engineering, Israel.
Information: Haim Azhari, DSc, Secretary of the Organising Committee, Julius Silver Institute of Biomedical Engineering, Technion-Israel Institute of Technology, Haifa 32000, Israel.

**19-23 September, 1995**
American Orthotic and Prosthetic Association Annual National Assembly, San Antonio, USA.
Information: Annette Suriani, 1650 King St. Suite 500, Alexandria, VA 22314, USA

**20-23 September, 1995**
17th Annual International Conference of the Engineering in Medicine and Biology Society, Montreal, Canada.
Information: Dr. F. Roberge, Montreal University, POB 6128, Station A, Montreal, Quebec H3C 3J7, Canada.

**6-9 October, 1995**
Annual Fall Meeting of the Biomedical Engineering Society, Boston, USA.
Information: Kate Strauss, BMES Fall Meeting, 45 Avon Rd., Wellesly, MA 02181, USA.

**9-12 October, 1995**
2nd Australian Conference on Technology for People with Disabilities, Adelaide, South Australia.
Information: Rehabilitation Engineering, Regency Park Centre, PO Box 209, Kilkenny, SA 5009, Australia.

**10-13 October, 1995**
3rd European Conference on the Advancement of Rehabilitation Technology, Lisbon, Portugal.
Information: ECART 3, National Secretariat for Rehabilitation, Av. Conde Valbom 63,1000 Lisbon, Portugal.

**27-28 October, 1995**
International Conference on Clinical Movement Analysis, Enschede, The Netherlands.
Information: ACON Convention Services, PO Box 560,7500 Enschede, The Netherlands.

**17-22 November, 1995**
American Academy of Physical Medicine and Rehabilitation Annual Meeting, Orlando, USA.
Information: AAPMR, Tel 312-922-9366, USA.

**20-24 November, 1995**
Engineering and Physics in Medicine, New Zealand.
Information: Dr. R. Jones, Medical Physics and Bioengineering, Christchurch Hospital, Christchurch, New Zealand.

**30 November-2 December, 1995**
Orprotec 95, Valencia, Spain.
Information: FMI Avenida de las Feiras, s/n E-46035 Benimamet, Apartado 476, E 46080 Valencia, Spain.

**1996**

**22-24 February, 1996**
Annual Scientific Meeting of ISPO UK National Member Society, Harrogate, England.
Information: Mr. D. Simpson, ISPO '96, National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, Curran Building, 131 St. James' Rd., Glasgow G4 0LS, Scotland.

**22-27 February, 1996**
Annual Convention of the American Academy of Orthopaedic Surgeons, Atlanta, USA.
Information: AAOS, 6630 North River Road., Rosemont, IL 60018-4226, USA.

**2-3 March, 1996**
Annual Scientific Meeting of British Association of Prosthetists and Orthotists, Glasgow, Scotland
Information: Mrs. H. Thomas, BAPO, Dunoon and District General Hospital, Dunoon, Argyll PA23 7RL, Scotland.

**22 April-5 May, 1996**
18th World Congress of Rehabilitation International, Auckland, New Zealand.
Information: Mrs. Bice Awan, Accident Rehabilitation and Compensation, Insurance Corporation, PO Box 242, Wellington, New Zealand.

**10-19 May, 1996**
8th International Mobility Conference, Trondheim, Norway.
Information: IMC 8, Tambartun National Resource Centre, N-7084 Melhus, Norway.

**12-16 May, 1996**
1st Mediterranean Congress of Physical Medicine and Rehabilitation, Herzliya, Israel.
Information: Congress Secretariat, c/o Ortra Ltd., PO Box 50432, Tel Aviv 61500, Israel.

**9-13 June, 1996**
10th Nordic-Baltic Conference on Biomedical Engineering, Tampere, Finland.
Information: Soile Lonnqvist, Ragnar Granit Institute, Tampere University of Technology, PO Box 692, Tampere, Finland.

**11-14 June, 1996**
4th International Conference on Human Services Information Technology Applications, Rovaniemi, Finland.
Information: NAWH, Husita 4 Bureau, PO Box 220,00531 Helsinki, Finland.

**12-15 June, 1996**
13th Interbor International Congress of Prosthetics and Orthotics, Oslo, Norway.
Information: Congress Secretariat, XIII Interbor, Congress Conference AS, PO Box 7609 Skillebekk, N-0205 Oslo, Norway.

**12-16 August, 1996**
3rd Paralympic Congress, Atlanta, USA.
Information: 3rd Paralympic Congress, Atlanta Paralympic Organizing Committee, Youth and Community Programs, 1201 West Peachtree Street NE, Suite 2500, Atlanta, GA 30309-3448, USA.

**28-31 August, 1996**
10th Conference of the European Society of Biomechanics, Leuven, Belgium.
Information: Biomechanics and Engineering Design Division, Katholieke Universiteit Leuven, Celestijnenlaan 200A, B-3001, Heverlee, Belgium.

**16-20 September, 1996**
18th World Congress of Rehabilitation International, Auckland, New Zealand.
Information: Convention Management, PO Box 2009, Auckland, New Zealand.

**13-16 October, 1996**
1st International Conference on Priorities in Health Care, Stockholm, Sweden.
Information: Priorities in Health Care, Stockholm Convention Bureau, PO Box 6911, S-102 39 Stockholm, Sweden.

**1998**

**28 June-3 July, 1988**
9th World Congress of the International Society for Prosthetics and Orthotics, Amsterdam, The Netherlands.
Information: Congrex (Holland) B.V., Keizersgracht 782,1017 EC Amsterdam, The Netherlands.
Tel: +31 20 626 13 72. Fax:+31 20 625 95 74.

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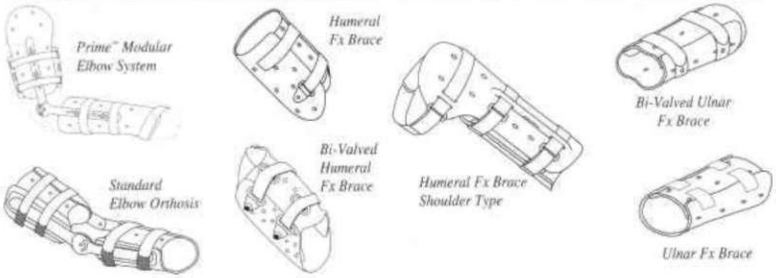

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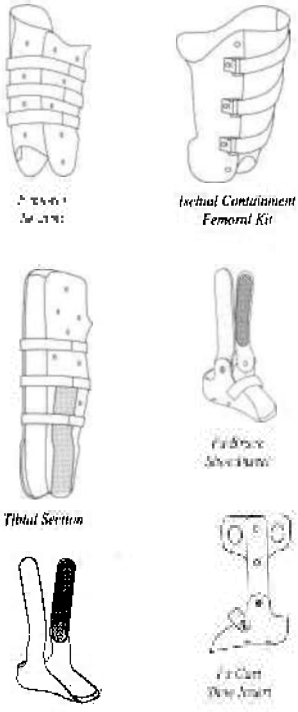

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Ischial Containment Femoral Kit

Tibial Section

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Solid Ankle Shoe Insert

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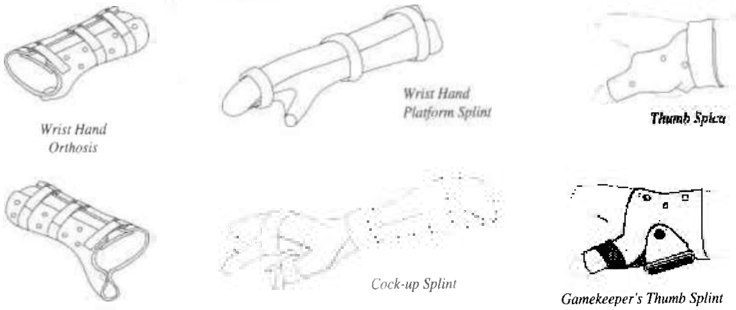
## HAND ORTHOSES



Wrist Hand Orthosis

Wrist Hand Platform Splint

Thumb Spica

Wrist Hand Thumb Orthosis

Cock-up Splint

Gamekeeper's Thumb Splint

## FOOT ORTHOSES

Ankle Foot Orthosis (AFO)

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