The Journal of the International Society for Prosthetics and Orthotics

Prosthetics and Orthotics International

August 1997, Vol. 21, No. 2

Otto Bock

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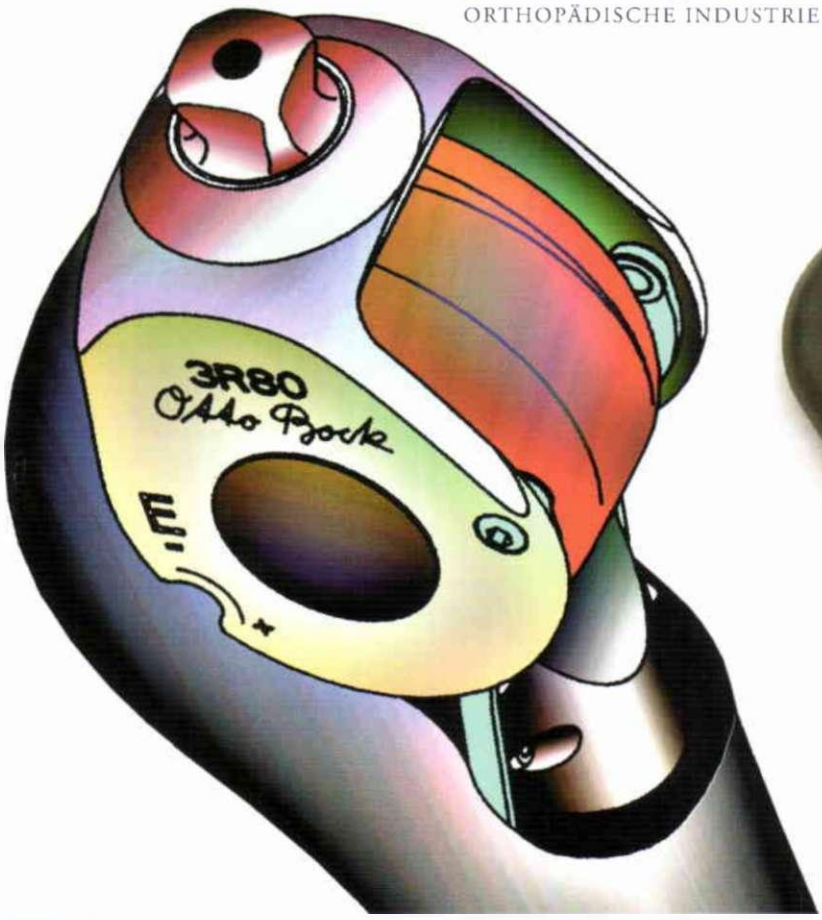

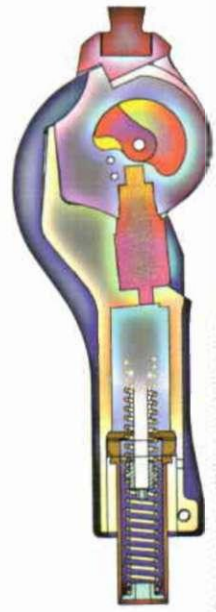

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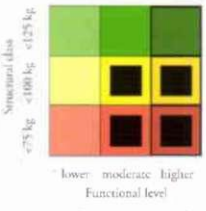

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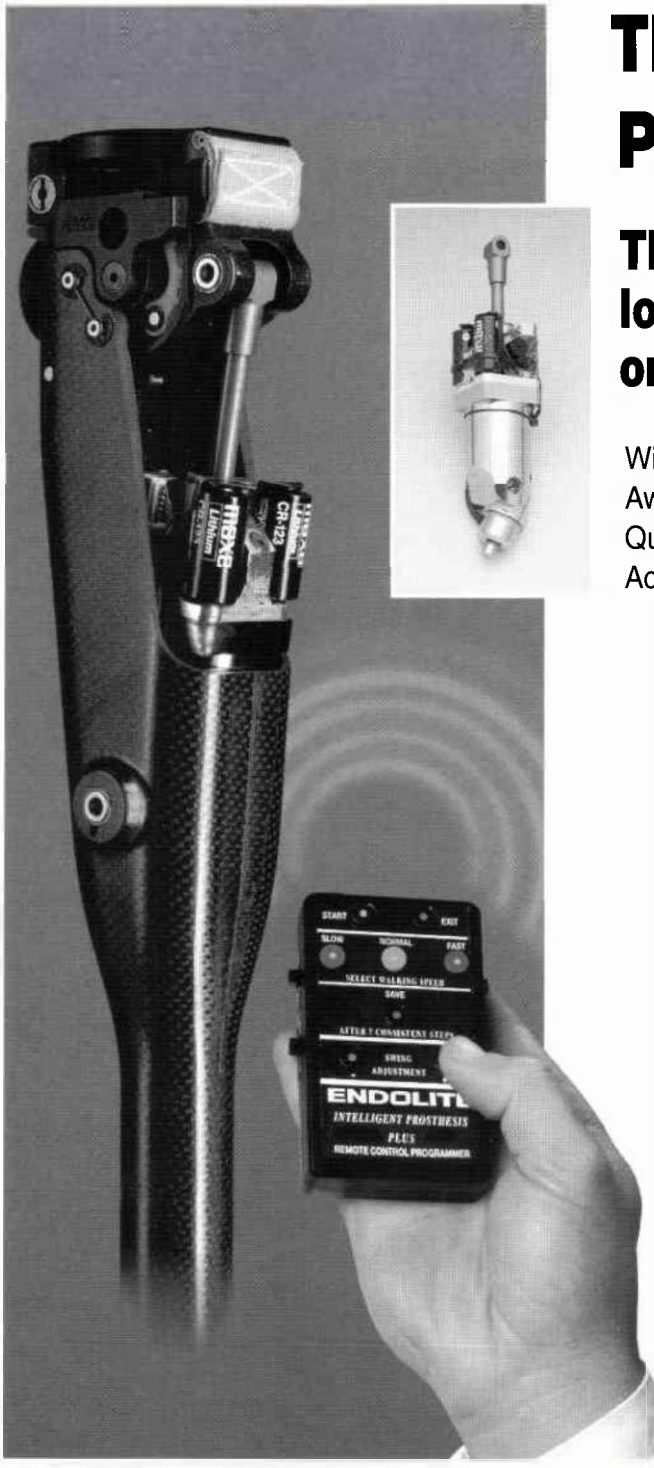

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Editorial

To those who read the reports which appear regularly in this Journal from Executive Board meetings, it must be apparent that more effort and time is expended on education than on any other single area. Education is of course a cornerstone of the Society's Constitution and in view of the enormous shortfall of prosthetist/orthotists in the developing world which is highlighted in report after report from the international agencies, the Society's emphasis is not surprising. The reason for the shortfall is obvious. There are not enough training programmes. And there are not enough training programmes because of limited resources. The Society on its own cannot change this but it can change attitudes to the importance of adequate levels of training and it can encourage and foster those training programmes which do exist.

For many years there has been agreement that in the developed world the future for prosthetist/orthotist education is at degree level. That of course must be the ultimate aim throughout the developing world but for the meantime it is not an attainable goal. On the other hand many well-intentioned agencies, not so long ago, promoted a level of training in their projects which in the worst examples was little better than primitive. Over a series of meetings the Society developed a compromise solution which promoted the idea of the Orthopaedic Technologist; a professional with a structured training programme lower than degree level but still able competently to address the needs of disabled people.

To foster the idea of the orthopaedic technologist and to give agencies and national ministries confidence in the level of training, the Society developed a system of recognition of courses. This involved submission of a questionnaire, an inspection visit usually timed to coincide with final examinations and if the Society was satisfied, and frequently subject to conditions, recognition would be granted for a three-year period.

This process has done much to stimulate the raising of standards and increasingly the Society's standard is being seen by all the major agencies as the appropriate and acceptable one. The Society has now taken a further step with the production of an Information Package for the Orthopaedic Technologist which specifies in detail the Professional Profile of the individual, the Learning Objectives of the course and gives guidelines on an acceptable Examination Process. Schools which satisfy the content of the Information Package, still subject to inspection, will be recognised as before but further, the graduates of the programme will be Registered by the Society and entitled to describe themselves as ISPO Registered Orthopaedic Technologists. It is thought with some confidence that this will further enhance the professional status of this important group of workers and encourage more programmes to seek recognition.

The Society and in particular the Education Committee is not ignoring the needs of the developed world and is working on the production of a similar Information Package for the Prosthetist/Orthotist. This is not for the many nations which have adequate training but for the many which still aspire to an appropriate level but need reliable guidance as to the detail of a programme which satisfies an international standard. However, much effort will continue to be needed in tackling the problems of the developing world and in seeking to further increase the emphasis on training there. The need continues to be a staggering one but with enlightenment and persistence progress is being made.

John Hughes
Chairman, Education Committee

Prosthetics and Orthotics International, 1997, 21, 78-84

International Conference on Orthopaedic Technology Wuhan, People's Republic of China

4-9 NOVEMBER, 1996

Introduction

The International Conference on Orthopaedic Technology was organised by the steering committee on behalf of the German Foundation for International Development (Deutsche Stiftung für Internationale Entwicklung, DSE) and the German Agency for Technical Cooperation (Deutsche Gesellschaft für Technische Zusammenarbeit, GTZ). The stated aim was to produce a framework document and in particular a "declaration" which would guide the development of orthopaedic technology activities in the non-industrial world into the next millennium.

Attendance at the conference was by invitation. It brought together workers from GTZ projects around the world, representatives of major international governmental and non-governmental agencies and identified experts in orthopaedic technology and health care provision.

In particular, the conference focused on rehabilitation as part of primary health care including community-based rehabilitation; the financing of prosthetic and orthotic care in developing countries; education and training; methods and procedures in appropriate technology; objectives and performance indicators of orthopaedic care systems. Prior to the conference, all participants and other selected experts were invited to submit their views as to the important issues in the above areas identifying lessons learned in completed and current projects, offering opinion on priorities and formulating guidance based on experience. From the response of the participants a draft declaration was produced and edited by the steering committee.

This draft declaration formed the basis for discussion at the conference which was programmed in sections corresponding to the areas of interest identified above. Following expert presentations in each area, the participants were divided into syndicates each of which was asked to endorse, comment on or amend the section of the draft declaration under consideration. Syndicate reports were produced and discussed in plenary session, and combined and edited to form a definitive section of the declaration which was once more discussed in plenary session to approve the finalised version.

The Wuhan Declaration is thus an amalgam of the participants' individual opinions, widely and thoroughly discussed against a background of informed presentations, forged into a coherent document and finally discussed and approved by the participants in the form presented here.

The Wuhan Declaration addresses the whole range of issues facing the aid and other agencies working in this field. It covers the entire spectrum from identification of need through project implementation to hand-over. It does not of course provide all the answers; indeed it raises many questions. It does however constitute an invaluable guideline, representing as it does the combined opinion of one of the most informed and involved groups ever brought together in this field.

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Wuhan Declaration

1. Rehabilitation as part of primary health care including community-based rehabilitation.

It is universally acknowledged that, under the Alma-Ata Declaration, primary health care services that are preventative, curative, promotive and rehabilitative should be made available to the community.

1.1 In developing countries, acknowledging the scarcity of health care providers, Primary Health Care services should be rationally distributed with equity, affordability, appropriateness and accessibility.

1.2 Health authorities should ensure that the community is aware of, and takes responsibility for utilisation of, the services available.

1.3 Basic orthotics and prosthetics services are integral to rehabilitation. However, in developing countries those services which are not readily available should be reviewed to enable services in line with Primary Health Care/Community-Based Rehabilitation to be provided.

1.4 Governments should embrace prosthetics and orthotics services within a national health care plan and develop national policy that would ensure the delivery of appropriate and adequate rehabilitation services.

1.5 There should be a closer coordination between all the government ministries and non-governmental organisations who are involved in the rehabilitation and the reintegration into the community of people with disabilities.

1.6 Collaboration between prosthetics and orthotics professionals and other health care providers should be systematised.

1.7 Rehabilitation services should be available at community level. The community rehabilitation /primary health care worker at this level must receive training in:

- early detection of health problems within the community;
- identification of the need to see a doctor, surgeon or prosthetics and orthotics professional for provision of a prosthesis or orthosis;
- the correct use of prostheses and orthoses and should be able to recognise and identify problems associated with fit and functional failure.

1.8 Through community-based rehabilitation, communities can make contributions to:

- providing information on disability within the locality;
- providing information on the use of prostheses and orthoses;
- providing a base for rehabilitation, training and the improvement of independent life and work capacity;
- helping people with disabilities obtain employment;
- facilitating participation in activities within the community;
- helping people/children with disabilities to be re-integrated into society/school;

- helping prevent disease and disability.

1.9 The training of community rehabilitation workers with respect to prosthetics and orthotics should be provided by trained prosthetics and orthotics professionals.

1.10 Educational material for use in primary health care services should be prepared by prosthetics and orthotics professionals.

1.11 People with disabilities should be consulted on and involved in the establishment of prosthetics and orthotics rehabilitation services.

1.12 Women should be encouraged to become prosthetics and orthotics professionals and community rehabilitation workers in order to improve access to services by all patients.

1.13 Health authorities should regularly monitor and evaluate the services provided, including those at community level.

2. Financing prosthetics and orthotics care in developing countries.

2.1 Before establishing projects or supporting national programmes external agencies should initiate discussions with national governments or agencies with respect to funding and financial viability. There should be an expressed commitment by national government or institutions to assure financial sustainability and take over programmes supported by external funding. In addition to the basic funding which includes capital, running, staffing and consumable costs, funds for training are essential. The government should make a strategic plan and define the roles of non-governmental organisations and the private sector.

2.2 Before establishing a project, a situation analysis should be carried out in order to set the aims, objectives and scope. There should be equal access for the whole population.

2.3 From the start of an externally assisted project national government must be involved in planning (including workforce planning).

2.4 From the start of planning prosthetics and orthotics services in a country, suitable funding mechanisms leading to the development of a sustainable system should be identified. Sources of funding include national or regional government, national and international organisations, individuals and patient contributions.

2.5 There should be a system in place for the costing, auditing and budgeting of all aspects of prosthetics and orthotics financing, including cost incurred by patients.

2.6 A primary aim of externally supported projects which establish, upgrade or improve service units and/or training institutions should be the handing over of responsibility to national or local agencies. The main objective should always be to make sure that the services handed over are adapted to the national resources and management capacities available in such a way that the necessary conditions for sustainability are established.

2.7 Inclusion of prosthetics and orthotics services into the national health care funding system should be promoted. This may be financed by government, insurance or other means.

2.8 Conflict and post-conflict situations could justify service provision on humanitarian grounds, that is not negotiated with a functional national government. However, when the political and socio-economic situation improves, all the above must be considered.

2.9 In order to ensure good financial and economic management externally supported projects must:

- consult and advise competent authorities in price setting adapted to the social and economic conditions of the country;
- investigate ways of reducing cost through:
  - establishing local or regional manufacture of components
  - central fabrication
  - decentralisation of care (as close as possible to the patient);
- define minimum standards of quality;
- introduce quality control;
- seek more economic material acquisition, either locally or regionally;
- develop appropriate prosthetics and orthotics care systems;
- provide equity of services by ensuring that those who are able to pay subsidise those who are not.

2.10 In order to be more cost-effective, consideration should be given to:

- recycling of components, parts and materials;
- using low cost locally available materials.

2.11 In order to promote the services and improve the working conditions, the following should be implemented:

- "awareness" programme about prosthetics and orthotics within countries amongst decision makers, commerce, educated class, and the media;
- formation of a national rehabilitation society;
- establishment of a national professional society;
- enhancement of the service structure so that professionals have good job opportunities after training.

2.12 Whenever possible, patients should contribute to the cost. Charges should be made based on a standardised method of costing. Partial payment by the patients should be according to their capacity to pay, including the possibility of total exemption.

2.13 Encouragement should be given to the establishment of private sector workshops in a country, taking account of the economic and social factors prevalent. These workshops could provide an acceptable standard of service and be integrated into, or coordinated with, the public sector service. It would be possible to influence acceptance of moving towards the private sector by:

- setting up a legal framework including accreditation of professionals;
- clarifying the regulatory function of the government or public bodies;

- business management courses;
- development and implementation of a system of quality assurance;
- financial assistance for private activities, e.g. credit, etc.

2.14 Performance indicators in the provision and use of prostheses and orthoses should be established.

2.15 Inter-agency cooperation between the different organisations working within a country or region should be promoted. Consumer groups and the private sector should be included.

2.16 Externally funded initiatives for the provision of services should not jeopardise those services already existing, but support them.

3. Education and training in prosthetics and orthotics in developing countries.

3.1 The ISPO classification of prosthetics and orthotics professionals is endorsed:

- Category I - prosthetist/orthotist, orthopaedic meister
- Category II - orthopaedic technologist
- Category III - prosthetics/orthotics technician
- Category IV - workshop assistant

however, ISPO is urged to re-examine this categorisation with a view to the inclusion of other workers in the field.

3.2 Training and education of orthopaedic technologists (Category II) should be the immediate focus in order to provide a long term and improved service for people with disabilities. With regard to the education and training of the orthopaedic technologist:

- the education and training syllabus must be oriented to the internationally recognised professional profile and the ISPO/WHO guidelines from Alexandria;
- if there is no national or regional professional profile existing, it is mandatory that one should be introduced before setting up an education and training programme in order to establish professional recognition and a service structure;
- the contents of the education and training should be periodically reviewed and updated to include new developments in the field, including appropriate technologies and their effects in the development of prosthetics and orthotics services;
- education and training programmes should aim to produce professionals capable of adapting to the use of different technologies;
- the education of teachers is mandatory. Teachers should be Category I and their education and training should start before establishing the local educational training programme. A programme of education and training of teachers should be finalised and internationally recognised by ISPO as a guideline to responsible authorities;
- within the training and education activities there should be more emphasis on:
  - clinical management of patients
  - appropriate technologies
  - quality control/endurance of devices

- financial management (cost calculation)
- staff development/management;

- besides the regular education and training programmes, possibilities should be given to allow others to be upgraded or enter the profession. For example:
  - short courses leading to complete education and training
  - modules for upgrading
  - a final professional examination of acceptable standard independent of the education and training received.

3.3 An external assessor/examiner should take part in the final examinations and sit on the final examination board in order to strengthen the education and training courses and to guarantee their international standard.

3.4 Any education and training programme should be carried out within the educational system of the country and preferably in association with an educational facility which can provide an ongoing infrastructure, administration and teaching faculty.

3.5 The education and training programme should be recognised by the appropriate national authorities as well as being recognised by ISPO, in order to meet international standards.

3.6 Training should be open to students from other countries to fully utilise training capacities and to generate income.

3.7 Clinical activities should be undertaken by orthopaedic technologists (Category II) with access to more senior experienced personnel (Category I) for the treatment of difficult cases and for purposes of professional development.

3.8 The contents of the education and training programme should be such that the orthopaedic technology graduate (Category II) can work independently within his/her area of experience, adapt creatively to situations encountered in the working environment and cooperate with other medical and paramedical professionals involved.

3.9 Continuing professional development of all staff involved in prosthetics and orthotics is essential.

3.10 The need for prosthetics and orthotics technicians (Category III) is great and consideration should be given to their education and training.

3.11 Sustainability of an education and training programme should be planned at the beginning and achieved before the externally funded project has come to an end.

4. Appropriate technology : methods and procedures

Appropriate technology is a system providing proper fit and alignment based on sound biomechanical principles which suit the needs of the individual and can be sustained by the country at the most economical and affordable price (ISPO, 1996).

4.1 An appropriate technology should meet the physical needs of people with disabilities within their environment and economic situation.

4.2 Local production of components from locally available materials should be encouraged.

4.3 Component manufacturing units should be functionally separate from clinical facilities.

4.4 Priority should be given to providing prosthetic and orthotic appliances which are made from locally produced or available components.

4.5 The use of different methods, procedures and materials should be investigated, tested and evaluated.

4.6 New technologies, such as CAD CAM, may be adaptable to developing countries and should be further investigated to study their effects on:

- manpower requirements;
- production;
- quality;
- cost;
- delivery time.

4.7 The adoption of standards and the use of quality management in the production of prosthetic and orthotic components and devices is recommended.

4.8 ISPO/WHO should be encouraged to develop standardised protocols to evaluate new technologies.

4.9 Experiences related to appropriate technologies and the use of materials should be properly documented and made available to others.

4.10 Regional collaboration in respect of local component production should be encouraged.

5. Assessment of needs and performance indicators.

5.1 Prosthetics and orthotics projects should be developed against the background of a national needs assessment.

5.2 It is necessary to develop performance indicators to evaluate effectiveness both during and at the end of a project.

Reference.

ISPO (1996). Report of ISPO Consensus Conference on Appropriate Prosthetic Technology for Developing Countries, Phnom Penh, Cambodia 5-16 June 1995./edited by HJB Day, J Hughes, NA Jacobs. - Copenhagen: ISPO.

Prosthetics and Orthotics International, 1997, 21, 85–91

The population of users of upper limb prostheses attending the Oxford Limb Fitting Service

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Abstract

There is no central database that records the changing provision of prostheses in the United Kingdom. Experience suggests there have been some shifts in the population, particularly in the past decade. Because the detailed records of these changes are contained in the patients' medical records it is difficult to assess the substance of these data except on an individual basis; the larger picture requires the sifting of a centre's or many centres' data. This paper describes the analysis of one such set of records at the Oxford Limb Fitting Centre. It relates the profile of the population that attends the centre to the general population, and compares the information with that obtainable from other sources. The possible causes for the findings are discussed.

Introduction

Prostheses and related care are provided over many years. Patients may change geographic area, treatment form and medical team, many times. Their medical records often bear testimony to the changes that the individual and the prosthetic service have undergone. However, there is no central repository in the United Kingdom for detailed information concerning the limb deficient population. The information is spread throughout the service, in the localised databases for different aspects of the subjects' life. For example, the number of a certain type of prosthesis supplied in the area and its time of use is probably best found in the records of the supplier of the prosthesis and it is therefore commercially sensitive data. Some national data were previously available in the Amputation Statistics for England, Wales and Northern Ireland (Ministry of Health, 1967; Ministry of Health and Social Security, 1968-1979 and 1980-1988), but this ceased to be collected in 1989. So figures for the national population of prosthesis users are now difficult to obtain. In addition, there is no requirement to record any limb deficiencies at birth, and other sources of information, such as disability grants, do not report much details (McDonnell *et al.*, 1988). Therefore, the review of the population (whether locally or nationally) can only be achieved by reviewing the records of individual patients. This is one such review.

A second purpose for this audit was to establish if there were any patterns of prosthetic usage in the region. There are many factors that dictate if a person will use or reject a prosthesis. Though it is impossible to predict precisely why a person will find a prosthesis unhelpful, when a new design of prosthesis is contemplated it is important to gather as much information about use patterns as possible, in order to create a design that is widely acceptable. The study of the records of the user population of this particular centre formed part of a knowledge acquisition phase of a project to design an advanced prosthetic hand under the European Union's TIDE initiative (Technology Initiative for the Disabled and Elderly). It was made in the spring of 1992 in order to assess the utilisation of the service by patients, as well as

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and WWW.TAGISH.CO.UK/LOSH.

in preparation for a more detailed survey of the active user population attending the centre (Kyberd *et al., in preparation*). The population studied is that which attended the Oxford Limb Fitting Centre in the Nuffield Orthopaedic Centre, National Health Service Trust, Oxford, UK.

As this was an audit of the records from the Oxford Limb Fitting Centre, the data derived corresponds only to this sub-group of the limb deficient population of the UK and generalisations should be made with caution. These patients were targeted because the aim of this audit was to gather information prior to the design of a prosthesis.

Method

A list of the names of all the individuals who are recorded as receiving upper limb prosthesis care from the centre was obtained from computer records. No other details are retained on computer. The individual records were read and specific details were recorded:

- age;
- gender;
- cause and level of loss;
- type of appliance supplied, both the current device and previous systems;
- estimated level of usage.

For the purposes of the survey, the date of last contact was recorded. The approximate level of use of their principal prosthetic arm was judged from the frequency of the requests to maintain or service the arm and its components. A small number of records (7) was untraceable. These are not included in any of the following statistics. Periodically, the files of the centre are reviewed and reminders are sent to all patients who have not had any contact with the centre. This reminder includes a brief questionnaire in an attempt to find out why no contact has been made. This may provoke a simple written reply or a request for an appointment. In either eventuality a reply was counted as positive contact.

The data obtained were analysed and compared with findings from other studies.

Centre population

The catchment area for the centre was, at the time, the Oxford Regional Health Authority (ORHA). The ORHA comprises approximately 2.58 million individuals (1.29m males, 1.29m females) (Oxford Regional Health Authority, 1991). The centre's region can be considered to be representative of the population of the United Kingdom as a whole as it has a balance of urban and rural areas. It draws individuals from Oxfordshire and Berkshire, much of Buckinghamshire as well as areas in Wiltshire and Northamptonshire. For the purposes of this survey this will be referred to as *The Oxford Area.* Patients are referred to the centre where they are assessed by the medical, therapy and

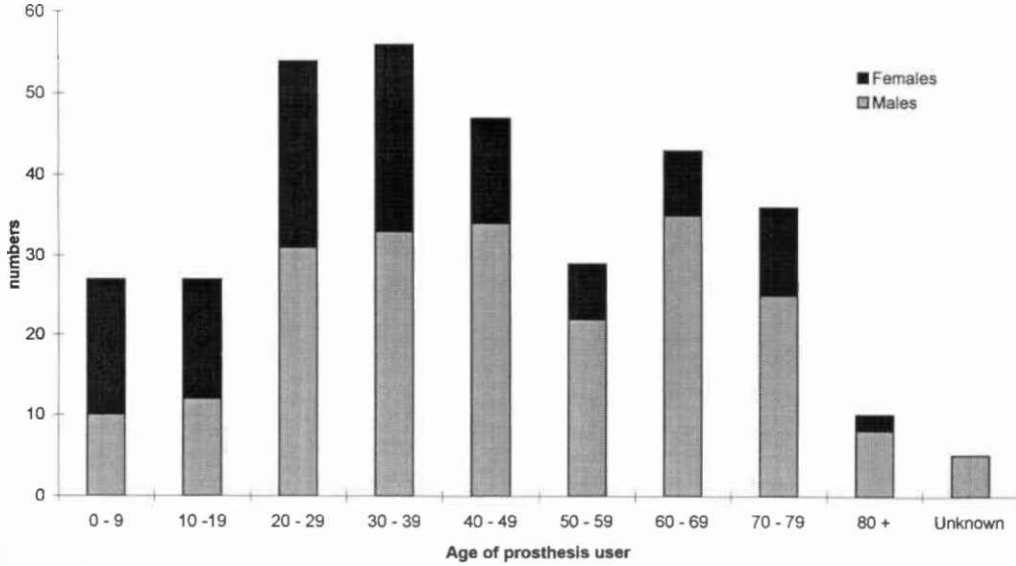

Fig. 1. The profile of the age of the population of individuals attending the Oxford Limb Fitting Centre.

Table 1. Total numbers of individuals listed for the Oxford Area limb fitting service.

Population	334	100%	plus 7 untraceable
Adult male	197	59%	
Adult female	93	28%	
Under 16s	44	13%	

prosthetic staff. Routine maintenance is carried out by contracts to the NHS. This may be carried out on attendance at the centre, or they may use the free postal service.

Results

Population

In February 1992 the Oxford Limb Fitting Centre had 341 patients listed as having attended the centre, Table 1 outlines the proportions.

The profile of the ages of the population is shown in Figure 1, arranged in intervals of ten years. It shows a similar number of patients for the first two decades, increasing for the individuals of working age and decreasing then on, with an additional peak in the years 60 to 80, when many would have been active during World War Two.

It is interesting to compare these with the available statistics of the entire regional population. The number for each age interval is normalised by dividing it by the total population in the region who are of the same age (data from Oxford Regional Authority (1991)) (Fig. 2). This gives the proportion of the population that are users of limbs and attend the centre. The result was a modest trend (r^2=0.45). The gradient of 1.2 x 10^{-7} shows that there appears to be only a small increase in the prevalence of prosthesis use with age.

Cause of loss

If the cause of loss is divided into the two broad categories of congenital and traumatic, it can be seen that the male population is dominated by traumatic amputations, of which more are right sided (Fig. 3). Of the males with congenital losses a left side bias is observable, similar to other studies (McDonnell *et al.*, 1988). The left bias in the congenital losses is shown in the female population, although there are far fewer traumatic losses. The larger number of females with a congenital loss is highly significant (p¡0.005, [Chi squared]) compared with the male population. This finding differs from other studies (McDonnell *et al.*, 1988).

Date of last contact

Figure 4 shows the date of the last contact an individual person made with the centre. A working definition for the 'active' population of users employed at the centre is that they have made contact with the centre at some point in the past two years. This definition was adopted for the study. Over 50% of the patients are therefore 'active'.

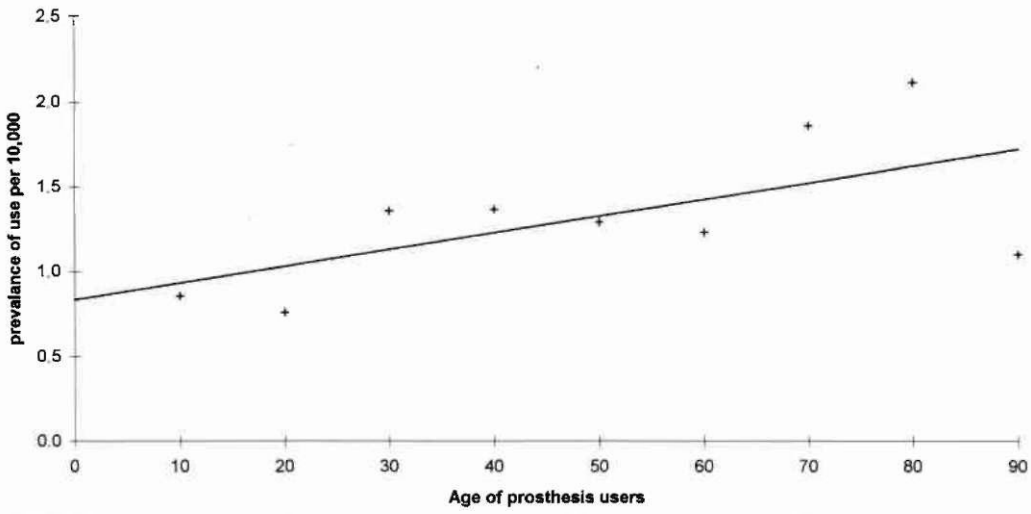

Fig. 2. Age profile of the audit population normalised by the total numbers of individuals in the same age categories within the *Oxford Area*.

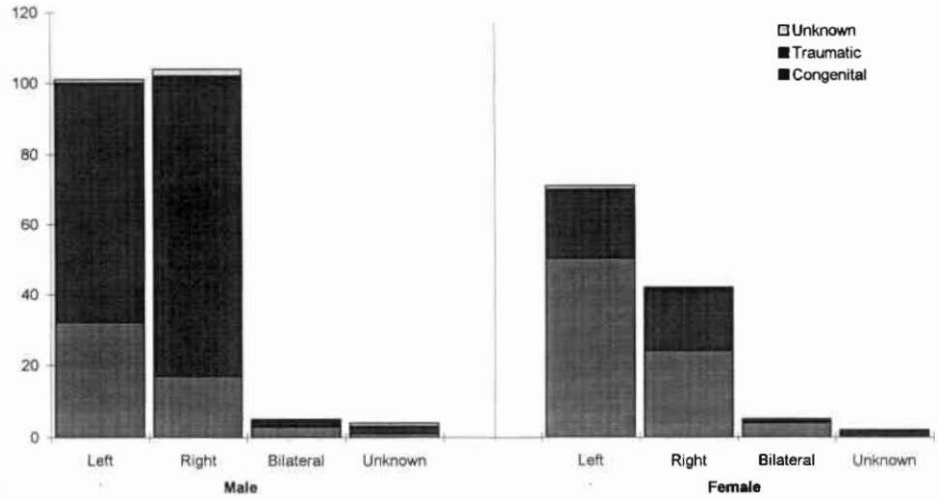

Fig. 3. The cause of loss of upper limbs within the audit population.

Principal type of prosthesis

Table 2 indicates the types of prosthesis that were used by the population. The categories are: cosmetic, working (including mechanical hands, split hooks, 'C' hooks) with myoelectric hands as a separate category. One category was judged to be the device each person principally used. This selection was based on the frequency of replacement/repair of the various devices the person was supplied with. For the entire recorded population the numbers of users of cosmetic and working hands are similar (133 to 162), if the active criteria are applied the number of working prostheses becomes proportionally higher.

Level of loss

Table 3 shows that in the population studied the largest single group are those with a trans-radial absence (163, 49%). The division of active and non-active users shows that there is no particular bias towards one group or level. In addition Table 4 shows a comparison with studies made at other centres.

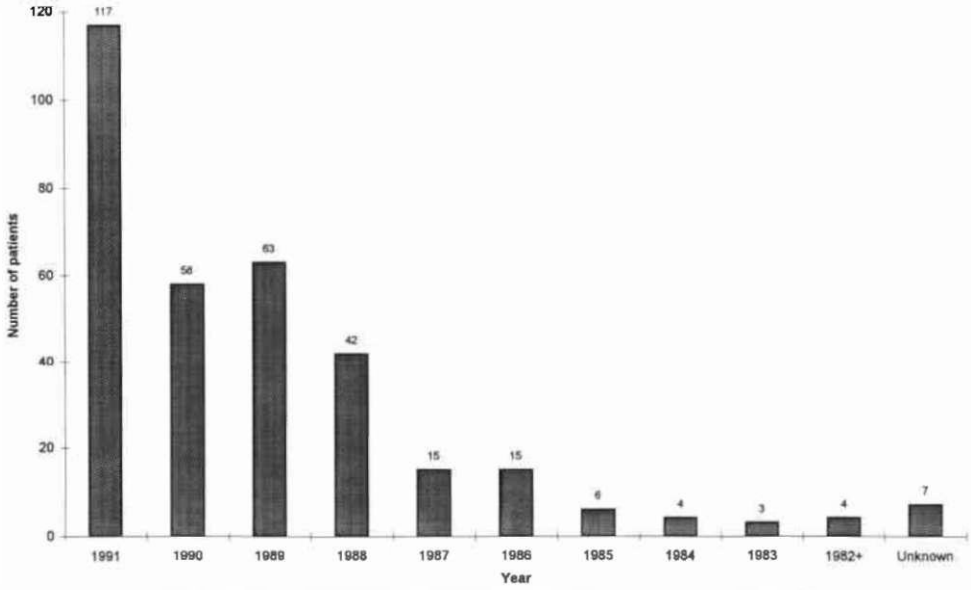

Fig. 4. Year of last contact of persons attending the Oxford limb fitting service.

Table 2. Types of principal prosthesis used, showing the proportions of users related to their most recent year of attendance.

Type	2 Years	4 Years	Total
Cosmetic	65	44	133
Working	91	45	162
Myelectric	9	4	13
Unknown	8	11	26

Table 3. The levels of loss of the centre's population with the proportions of the most recent attendances shown.

Level	2 Years	4 Years	Total
Trans-humeral	55	28	97
Trans-radial	88	49	163
Partial hand	32	27	71
Unknown	0	1	3

Discussion

Population

A comparison between the patient group and the population of the region shows only a small increase in the prevalence of prosthesis use with age (the gradient represents a rate of only 1 in 810,000 per year) and there appears to be no major increase with advancing years. This contrasts strongly with amputations of lower limbs. The dominant cause of lower limb loss over the age of 50 is related to vascular problems, which generally increase with age, and so a similar graph for these patients would probably show an increasingly steep curve. By contrast, the upper limb amputations are more often caused by work and recreational activities; this is also likely to explain the gender profile.

The gradient of the normalised population is a measure of the change in the number of individuals who choose to use the service; the increase with age is slight. This measure includes those who recently suffered an amputation plus others who may not have wanted to use the service before, minus those who have stopped using the service.

The straight line fit for the data crosses the y-axis at 7.4×10^{-5}. This represents the population of individuals at the age of zero, i.e. those with congenital absences. This suggests a ratio of 1 in 3,500.

Cause of limb loss

The bias towards the male population, with the greatest numbers being right side from traumatic amputation agrees with other studies. This distribution reflects the tendency for a greater proportion of males to undertake more dangerous work and leisure activities. The bias towards the left side within the congenital losses again is in agreement with other studies (McDonnell *et al.*, 1988). However, the larger numbers of females with congenital loss on both left and right sides differs from other studies. It is not clear why this should occur. Once again these figures only reflect the losses that can benefit from prosthesis provision.

Date of last contact

The definition of an active user is an individual who has made contact with the centre in the past two years (this includes 50% of all patients); 80% of individuals made contact in the previous 4 years. According to Fraser (1993) Addenbrookes Disability Service Centre (DSC) considers three years to be the point that defines their active users, if this is taken for

Table 4. Comparision of the percentage of the levels of loss between the *Oxford Area* population and of other surveys. The data for losses at the shoulder are not availble in two surveys, this will tend to increase the proportion at the next highest loss.

Level	Fraser (1993)	Burroughs and Brook (1985)	Silcox et al (1993)	Atkins and and Meier (1988)	Burger and Marinček (1994)	Oxford (1994)
Shoulder disarticulation	8	21	-	5	1	0
Trans-humeral	33	14	25	23	21	29
Trans-humeral	33	52	68	60	65	49
Hand/wrist	25	14	7	12	13	21

Oxford, then 238 persons (73%) are included.

The term 'active' is not intended to assess if the device is passive or active, simply if the prosthesis is used sufficiently to require replacement/repair. It is accepted that upper limb usage is a variable activity and some individuals will return much less frequently although they remain successful users of their devices. The difference in the use profiles of such devices may have a bearing on the time between repairs.

Principal types of prosthesis

There appears to be little difference in number between those issued with cosmetic and working prostheses (48% to 40%). When the active user definition is applied the balance shifts towards the working devices (52% to 37%). By four years the balance between the two groups mimics that for the total population (48% to 40%). This is consistent with the idea that the 'active' users would appear to employ their working prosthesis more often and so the components will wear out more quickly and thus need replacing earlier.

Level of loss

The comparison of these data obtained from the Oxford population with those derived from other surveys conducted in developed countries in the past 15 years shows a broad agreement. The variation in survey findings is less in the categories which have many individuals (i.e. trans-radial loss) and more in the extremes which are both rarer and less well served by current prosthetic technology (e.g. shoulder disarticulation level). Some of the data are derived from surveys of the users of prostheses so, for example, persons with a partial hand loss may not be included in such surveys. Indeed this could suggest that the data from the Oxford Area population are more representative of the genuine population as this sample is of all the users of the centre whether they employ a prosthesis or not.

Overall population

The population attending the Oxford limb fitting centre is not atypical compared with other groups. In the present economic climate with reductions in resource allocation it is important to know what population is likely to derive most benefit from the healthcare services. This investigation attempts to provide some data to aid the informed planning of upper limb provision.

A larger male population is also found elsewhere, for both congenital and acquired loss (Royala *et al.*, 1974; Gregory-Dean, 1991). This is hardly surprising as it is likely to be due to the combined contributions of the individuals' lifestyle/occupation to the traumatic cases. Congenital absence is domination by left trans-radial absences (McDonnell *et al.*, 1988). From these data the comparison of the proportions of gender, type of loss, side and type of prosthesis used shows a dominance of users in favour of male, traumatic loss, right sides and working prosthesis.

The overall numbers of myoelectric hands fitted to adults are small. The provision of myoelectric hands to adults has become routine only recently. In the initial stages of their introduction the provision was made at a few expert centres, such as Queen Mary's Hospital in Roehampton, London. Thus, some of those who were attending Roehampton may well be continuing to attend those clinics rather than switching to Oxford when it began to offer the service.

The figures for new attendees at centres throughout the country show a real trend downwards over the last 20 years (Chappell, 1992). The causes for this trend are unclear. As there is no evidence to suggest that the number of congenital deformities has changed appreciably, the reasons for the reduction can either be due to: (i) the improved medical treatment following trauma, or (ii) that fewer people are being referred to the service. One further possible reason for the fewer traumatic injuries is the decline in heavy industry in the United Kingdom in the past generation, exposing fewer individuals to risk of injury.

McDonnell *et al.* (1988) attempts to derive a measure of rate of congenital absence from a variety of reports from developed countries. From UK government statistics they suggests a range from 1 in 8,400 to 1 in 10,000; McDonnell derived from this a figure of 1 in 9,400. From the Oxford population normalised by age (Fig. 2) the zero offset of 7.3×10^{-5} corresponds to a congenital absence rate of around 1:13,500 for the Oxford data. This is between 20 and 40% lower than the above estimates. As there has not been a legal

requirement for birth abnormalities to be recorded, it is possible that a number are not included in the derived figure. In addition, the level of absence is not recorded in the statistics so the higher reported rate may include many persons with a partial digit absence. Therefore the figure derived in this study is more likely to be a measure of the rate of absence for individuals that can use prosthetic services.

Conclusions

The population of users of upper limb prostheses attending the Oxford limb fitting centre reflects similar trends to those found at other centres throughout the world. The largest single group is men with right-side trans-radial, traumatic losses. However, in contrast with other studies, the larger group of people with congenital absences is female.

The distribution of cosmetic and working prosthesis is evenly spread over the entire population, but users of working prostheses tend to return to the centre more frequently.

There appears to be little increase in the proportion of the number of amputations with age which results in an increase in prosthetic use. In addition, the rate of congenitally produced abnormalities requiring use of a prosthesis are in the region of one in 13,500 live births.

Acknowledgements

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Epidemiology of lower limb amputees in the north of the Netherlands: aetiology, discharge destination and prosthetic use

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Abstract

The aim of this study was to give a retrospective review of all lower limb amputations performed in the 3 northern provinces of the Netherlands in 1991-1992. Assembled data were compared with the existing information in the National Medical Register (NMR) over the same period.

With the participation of all regional hospitals, 473 lower limb amputations from transpelvic to transmetatarsal level were identified. Of the amputations 94% were performed for vascular pathology, 3% for trauma, and 3% for oncologic reasons. After surgery a prosthesis was provided to 48% of the amputees.

The actual number of performed amputations exceeds the number of amputations registered by the NMR by 9%. Incidence rates of lower limb amputations in the Netherlands are 18-20/ 100,000 over the last 12 years. These numbers are lower than in other areas and show no sharp decrease in frequency compared with other countries in Western Europe.

Introduction

The number of recorded lower limb amputees with an amputation level ranging from transpelvic to tarsometatarsal has not changed in the Netherlands over the last ten years. The total number of primary lower limb amputations is about 2,000 per year in a national population of 15 million people. (Central Bureau of Statistics, 1995; SIG Health care information, 1994). Recent Dutch studies (Dwars, 1990; Hoofwijk, 1990; Hoogendoorn, 1988) collect data from the National Medical Register (NMR). These NMR data concern all major lower limb amputations performed in the Dutch hospitals.

Data collected by the NMR are compiled by the operating surgeons and hospital staff who record these data. Total results are then gathered by the NMR.

These national data only show the number and level of amputation operations, sex, age and average hospital stay.

There is limited knowledge about the cause of amputation, discharge destination and follow-up of these amputees in the Dutch population.

It is known that only 15% of the lower limb amputees in the north of the Netherlands are admitted to a rehabilitation centre for prosthetic fitting and training (Rommers *et al.*, 1996).

The north of the Netherlands comprises 3 provinces; Groningen, Friesland and Drenthe. The total number of inhabitants in this region is 1.6 million people. This study was aimed at identifying the total group of lower limb amputees who underwent surgery in the north of the Netherlands. Aetiology of the performed amputations, discharge destination and prosthetic fitting would be studied. The actual number of amputations performed in the north

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of the Netherlands would be compared to the total number amputees registered by the NMR in the same area. This comparison gives information about the reliability of the national data. Local and national data would be compared with previously published international studies about amputation surgery of the lower limb in the literature over the last 10 years.

Methods

National data

National data over the period 1982-1993 was obtained from the NMR.

This included all national recorded lower limb amputation operations, their amputation levels according to the codes used by the NMR, age and sex of the operated patients. The frequencies of amputation levels at trans-femoral (TF), knee disarticulation (KD) and trans-tibial level (TT), in the north of the Netherlands, were separately given by the NMR.

Local data in the north of the Netherlands

All surgeons in the 18 hospitals in the north of the Netherlands (one university and 17 general hospitals) co-operated in this study, after confirmation by the medical staff or the management of the local hospital. The operating surgeons identified their lower limb amputation operations performed in the years 1991 and 1992.

This included research of computer databank information where available, and was completed by a check of the operating theatre record books. This data was compared to the number of operations coded to the national health service and the insurance companies. This was done in order to have a complete overview of all performed amputation operations in both years.

After completing the patients lists the medical records were gathered by the hospital staff and were studied by the authors.

The operation level and the number of operations performed on each patient was checked in the operation records written by the operating surgeons. All additional information was gathered from the medical correspondence from the surgeon to the patient's general practitioner (GP). When necessary additional information concerning prescription of a lower limb prosthesis and rehabilitation, was obtained from the hospital's rehabilitation physician or from correspondence by the nursing home doctor.

After completion of the data according to a checklist of date and location of the lower limb amputation, discharge destination and rehabilitation programme, all identifying patient and hospital information was removed from the checklist and data was entered anonymously into a database computer. Afterwards data was only comparable by province location, in the same way the national data is published. This was in order to assure the required privacy of patients data according to rules given by the medical-ethical committee of the University Hospital Groningen.

Re-amputation was defined as amputation at a more proximal anatomical level than a former amputation. Operation procedures at the same anatomical level were defined as stump revisions and therefore not counted in this analysis.

Bilateral amputations in the same operation session are counted by their individual amputation operation.

Results

Studied population

During the studied period in the north of the Netherlands 473 lower limb amputation operations were performed. This included all

Table 1. Patient characteristics of all 473 lower limb amputations.

Sex	Male	61%
Age	≥65	79%
Amputation level	HP+HD	1%
	TF	35%
	KD	12%
	TT	47%
	Foot	6%
Reason for amputation	Vascular	94%
	Oncology	3%
	Trauma	3%
Re-amputation		17.5%
Prosthesis		48.3%

HP: Hemi-pelvectomy; HD Hip disarticulation; TF: Trans-femoral; KD: Knee disarticulation; TT: Trans-tibial; Foot: Transmalleolar to tarsometatarsal.

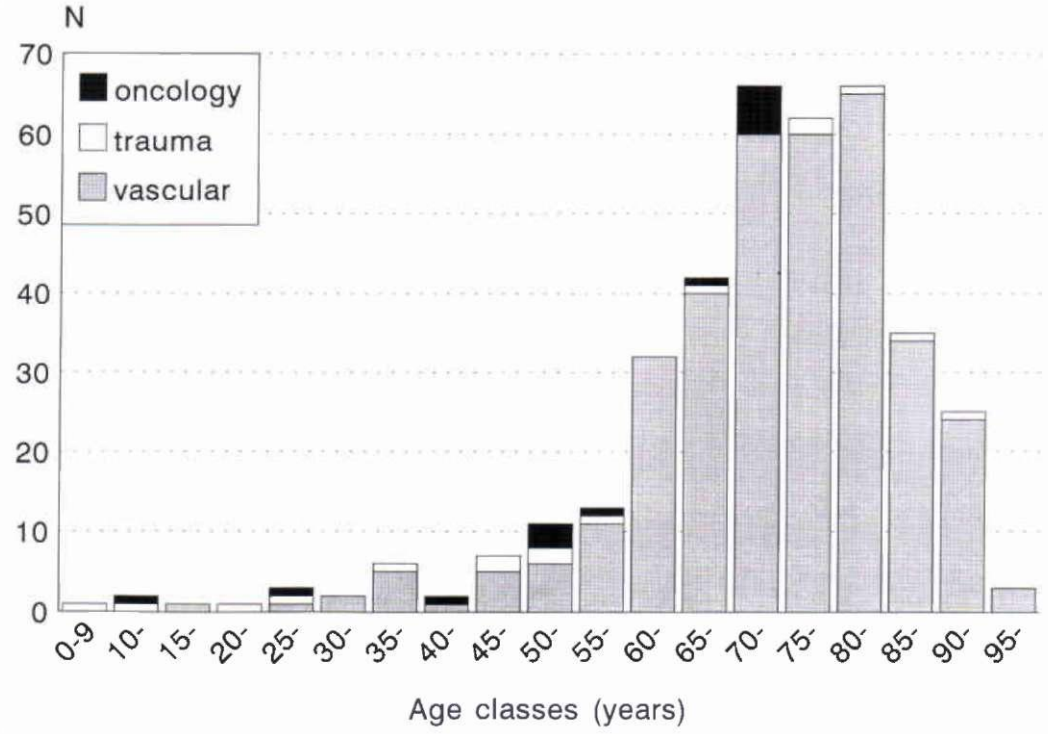

Fig. 1. Age distribution of all (N=415) lower limb amputees with the reason for amputation.

amputation from trans-pelvic to trans-metatarsal level. The age distribution of all patients related to the cause of amputation is given in Figure 1. Patient characteristics are shown in Table 1.

In this geographic area 94% of the operations were performed as a result of vascular problems. The number of diabetic patients could not be clearly identified using the data and was therefore omitted.

All 473 amputation operations were performed on 415 patients. Of these operations 17.5% were re-amputations.

Of these 473 amputations 448 lower limb amputations (94%) were performed at trans-femoral (TF), knee disarticulation (KD),and trans-tibial (TT) level.

Discharge destination and prosthetic fitting

Of all 415 operated amputee patients 47 (11%) died during the hospital stay, due to complications and co-morbidity. Information about the discharge destination after hospital stay was not always available from the studied data. In a minority of cases the name of the nursing home was not stated in the discharge letter. In other cases a home for the elderly was mixed up with a nursing home or even with the rehabilitation centre. By careful study of the clinical notes and the admission data in nursing homes in 96% of the cases the valid discharge destination could be found (Fig. 2). The majority of patients (40%) in the area were discharged to a nursing home. As found in a previous study in this territory (Rommers *et al.*, 1996) only 16% of the patients were discharged directly to the rehabilitation centre and 3% of the amputees were transferred to a home for the elderly. Not all patients stayed in their primary discharge location but were relocated to their home, rehabilitation centre or nursing home when appropriate.

A total of 191 patients (48%) were fitted with a prosthesis. This included the bilateral amputees who were provided with two appropriate prostheses. Of these 191 patients 39% received their first prosthesis during nursing home stay, 33% were treated in a rehabilitation centre and 22% were seen in outpatient clinics in the regional hospitals and were fitted with a prosthesis at home. Some 5%

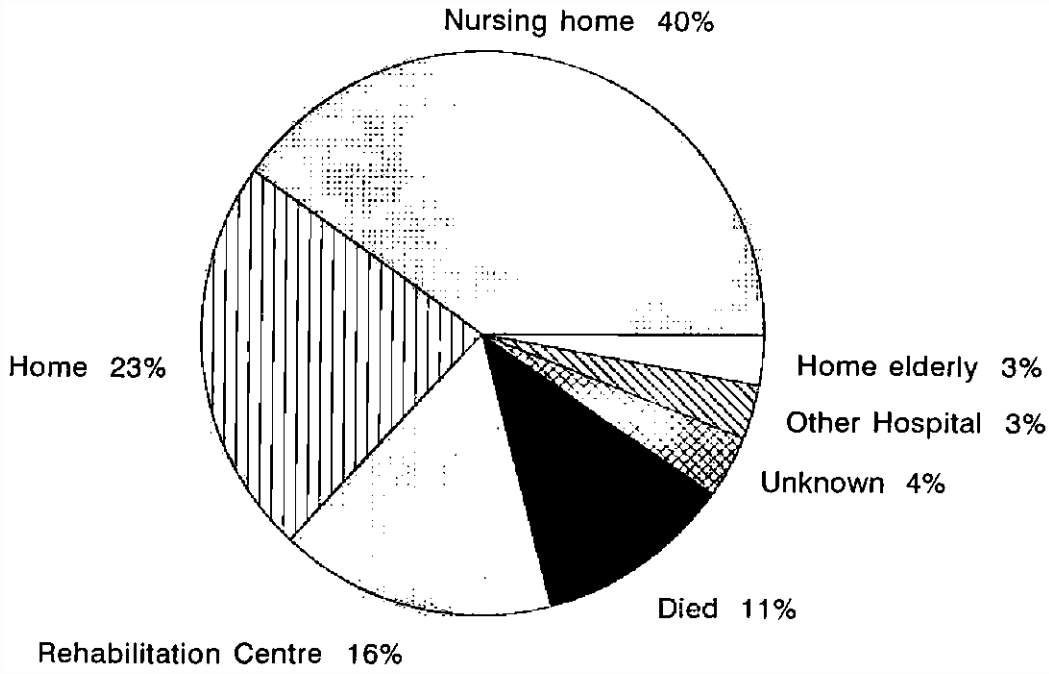

Fig. 2. Discharge destination of all (N=415) lower limb amputees. Discharge from operation hospital: north of the Netherlands 1991-1992.

were rehabilitated in the operating hospital or in another hospital in the region and 1% stayed in a home for the elderly and were supplied with a lower limb prosthesis in the local hospital.

National data compared with data in the north of the Netherlands

National data over the period 1982-1993 was obtained from the NMR in Utrecht, The Netherlands. These data show a slight increase in the incidence of lower limb amputations from 18/ 100,000 in 1982 to 20/ 100,000 in 1993. The level of amputation in the lower limb is becoming more distal over the studied years (Fig. 3). There was a sharp decline of TF amputations after 1982. The number of TT amputations is increasing as well as the number of foot amputations. The incidence of KD amputations varied, but averaged 1.8/ 100,000 for 1985-1993.

In the last few years the incidence of foot amputations (transmalleolar to transmetatarsal) increased to 3.3/ 100,000. The actual number of lower limb amputations nation-wide is about 3,000 per year instead of the 2,000 primary amputations recorded and published by the NMR (SIG Health care information, 1994).

Data collected from the NMR in the North of the Netherlands showed a number of 409 amputations in the period 1991-1992.

These included all TF, KD and TT amputations in the area registered by the hospital staff and collected by the NMR. These results are coded under NMR in Table 2.

Table 2. Comparisons between registered amputations by the NMR, and the observed amputations (OBS) in this study: 1991-1992, the north of the Netherlands.

	NMR				OBS			
	TF	KD	TT	Total	TF	KD	TT	Total
1991	74	19	106	199	83	30	99	212
1992	79	21	110	210	82	26	128	236

TF: Trans-femoral; KD: Knee disarticulation; TT: Trans-tibial.

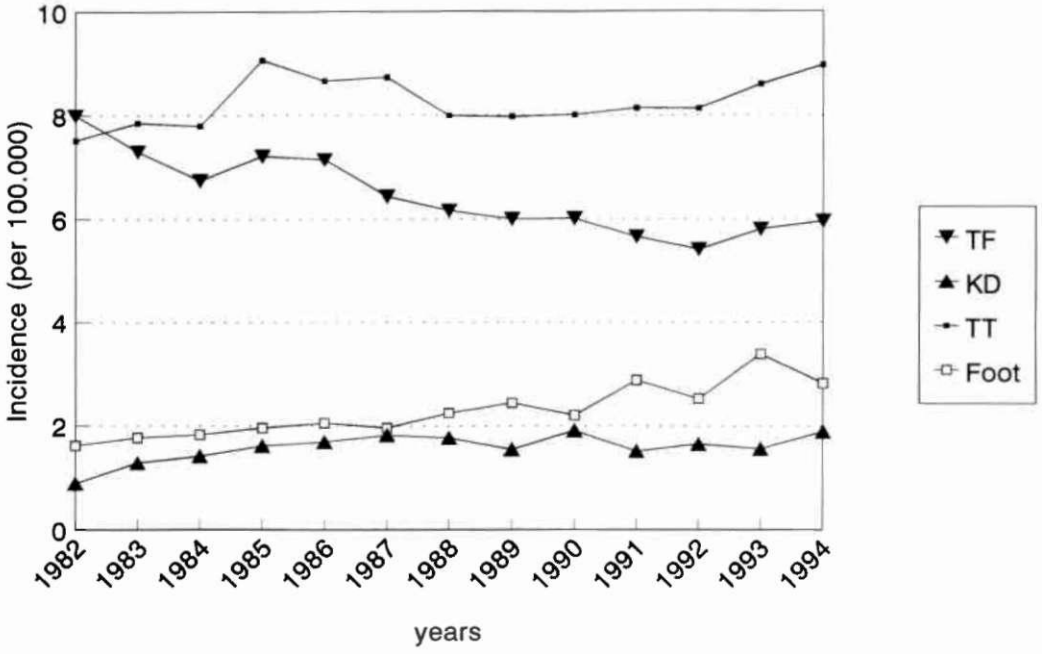

Fig 3. National data in the Netherlands 1982-1994. Incidence per year for the described amputation levels. TF = Trans-femoral; KD = Knee disarticulation; TT = Trans-tibial. Foot = transmalleolar to transmetatarsal.

The 'observed' data (OBS) from this study showed a difference of (212-199)/212 = 6% compared to the NMR data over 1991. In 1992 there was a difference of (236-210)/236 = 11% between reported and actual number of amputations.

Comparing the individual amputation levels, differences appear more significant. For instance the difference between the number of performed KD and TT amputations between NMR and OBS data is remarkable.

Differences are approximately (30-19)/30 = 36% for the number of KD amputations in 1991 between NMR and observed data. Differences between NMR and OBS are (128-110)/128 = 14% for the number of TT amputations in 1992.

It is thought that registration problems in the different hospitals are the cause of this difference. There is an overall difference of approximately 9% over the years 1991-1992.

Discussion

In this study the national data of lower limb amputations given by the NMR was compared with observed data over 1991-1992 in the north of the Netherlands. A control of this data is important since national data is used for policy making in the national health service and for study of the extent of certain diseases in the population.

The authors are certain that this study gives reliable results compared with the national data. It was shown that about 94% of all lower limb amputations are due to vascular pathology. This data is in accordance with data published by Ebskov (1992) of the Danish population. All 18 hospitals in the region participated in this study. All admitted patients with any kind of insurance are included in the study. Therefore there is a complete overview of data of the north of the Netherlands in the period 1991-1992.

In other studies (Dawson *et al.*, 1995; Pohjolainen and Alaranta, 1988; Siitonen *et al.*, 1993; Tunis *et al.*, 1991; Wahlberg *et al.*, 1994) only limited data is available or refers only to a single hospital. Apart from the data in this study only three studies (Alaranta *et al.*, 1995; Jones, 1990; Pohjolainen and Alaranta, 1988) give an overview of all vascular, trauma and oncological amputations in the studied populations (Table 3). The other studies reported here only take into account the

Table 3. Studies published in literature over the last 10 years.

Population	Study period	Cause of amputation	Source (population size)	Incidence /100,000 in general population
Southern Finland Pohjolainen & Alaranta (1988) Alaranta *et al.* (1995)	1984-1985	VTO	hospital area (1.1 x 10^6)	1984: 32.5 1985: 28.1
Eastern Finland Siitonen *et al.* (1993)	1978-1984	V	local area (253,000)	men: 33.9 women: 17.3
Australian States Jones (1990	1981-1985	VTO	3 states	1984: 23.6
Maryland USA Tunis *et al.* (1991)	1979-1989	V	1 state	28-32
Sweden Wahlberg *et al.* (1994)	1987-1992	V	hospital area (250,000)	1987: 26 1989: 16
Switzerland Enzler (1994)	1979-1989	V	½ of national data (3.1 x 10^6)	1979: 7.7 1989: 14
Denmark Eikhoff (1993)	1977-1990	V	national data (5.1 x 10^6)	1977: 25.8 1983: 32.2 1990: 23.1
Denmark Ebskov (1992)	1978-1990	V	national data (5.1 x 10^6)	1978: 30 1983: 34.5 1990: 25.0
The Netherlands Rommers *et al.*	1982-1993	VTO	national data (15.1 x 10^6)	1982: 18 1993: 20

Cause of amputation: V = Vascular; T = Trauma; O = Oncology. Source: studied area and population at risk in the study. Incidence: number of amputations/ 100,000 in the studied area.

vascular group and even exclude the bilateral amputations (Hoogendoorn, 1988), or foot amputations (Ebskov *et al.*, 1994). In a number of studies adequate information about the amputation levels are not stated at all.

Some 4% of patient data was not complete due to incomplete medical notes or because clinical records were not available for detailed study.

During the research of medical notes it was found that most of the registration in the hospitals is done by administrative personnel. We found differences between the performed operation and the registration by the hospital. Mistakes were made in amputations in the foot region, where the different anatomical levels were mixed up. Another source of bias were the operations at TT and KD level. In several cases it was stated that an amputation of the lower limb was performed. A TT amputation was coded but the actual operation performed was a KD amputation. In the case of several surgical procedures during an admission, sometimes not all performed operations were stated in the discharge letter to the GP which is the main source of information for the registration of medical data.

The discharge destination as stated in this letter is coded for the NMR. The authors believe that the institution code as used by the NMR is more reliable than the use of the names of the individual institution because of the risk of confusion. In several cases institutions were inaccurately recorded and this is one of the causes for the errors in the information about discharge destination after surgery. Previously

the authors found (Rommers *et al.*, 1996) that the actual admission frequency to the rehabilitation centre far exceeded the registered number of transferrals by the NMR to the rehabilitation centres in this territory.

The actual number of operations performed by the surgeons were calculated using calendar dates of the operations, whereas calculation of the NMR data was done at time of discharge. Small differences in number of amputation procedures can result from this, but will be equalised.

Information about the actual prescription of prostheses and the rehabilitation process is scarce in the Netherlands. Recent studies by Dawson *et al.* (1995) and Hoofwijk (1990) give some information, but this data was solely based on vascular amputees and only included a few hospitals. The actual prosthetic fitting rate of 48% in this study is in correspondence with Christensen (1995) and Eickhoff (1993). Dawson (1995) found 80% rate of prosthetic fitting in the surviving group, but only 54/81 = 66% of all patients involved actually received a prosthesis.

The actual place where the prosthesis is fitted and prosthetic training is given can be determined by regional differences. National data as gathered by the NMR about this issue is not available.

Parameters such as physical condition, social factors, age and co-morbidity are influencing factors in determining the discharge destination. Long term prosthetic use after discharge in a prospective cohort of patients is under study by the group and will be published in the future.

Studies of national data by Alaranta *et al.* (1995); Ebskov (1992); Ebskov *et al.* (1994); Eickhoff (1993) and Pohjolainen and Alaranta (1988) show a decrease in the incidence of lower limb amputations in Western Europe.

The Dutch national data shows a relatively constant incidence over the last 12 years. The Scandinavian authors (Ebskov, 1992; Ebskov *et al.*, 1994; Eickhoff, 1993; Pohjolainen and Alaranta, 1988) conclude that the increase in vascular reconstruction procedures is responsible for the decrease in lower limb amputations. Better control of diabetes in diabetic patients could have a positive contribution in this as well. The authors think that the influence of vascular surgery over the last 15 years in the Netherlands has gradually influenced the data and therefore did not show a sharp decrease as stated by the previously mentioned authors. It is difficult to compare data and incidence rates since not all authors state the precise amputation levels.

The total number of inhabitants in the studied areas is seldom given. This study gives low incidence rates, whilst counting all amputation levels in the lower limb. The authors agree with the LEA Study Group (1995) that standardisation is needed for comparable results in Western Europe and throughout the world.

The recorded number of lower limb amputees in the Netherlands of 2,000 per year (Central Bureau of Statistics, 1995; SIG Health Care Information, 1994) is not complete. The actual number of amputations is 3,000 (incidence of 20/100,000 x population) +8.5% (relative difference between NMR and OBS data) = 3,300 lower limb amputations from transpelvic to transmetatarsal level.

The influence of ageing of the population gives rise to much discussion about the near future. Figures given by the Central Bureau of Statistics (1995) indicate a 20% increase of people of 65 years and over from 1995-2015.

If the results in the north of the Netherlands are compared with the actual change of age distribution in the population, there is no evidence that increasing age in the population gave a higher incidence of amputation surgery of the lower limb in this region in the period 1982-1993.

Conclusions

National data of performed lower limb amputations in the Netherlands is reliable within certain limits. Registration in the hospitals for the NMR is done with care, but as is concluded from this study, it is incomplete, especially regarding individual amputation levels. The actual number of lower limb amputations is±60% higher than national published data. The re-amputation rate is about 17.5% in this area and is in line with previous results.

It was found in this study that 94% of the lower limb amputations in the region were due to vascular pathology, 3% because of trauma and 3% as a result of oncology.

Some 48% of the patients were fitted with a

prosthesis during their stay in hospital, nursing home or rehabilitation centre.

Incidence rates for lower limb amputations are about 18-20/ 100,000 over the last 12 years and show no sharp decrease in frequency as a result of vascular surgery as has been shown by other authors.

Amputation registration, discharge planning and prosthetic fitting give valuable information for the adequate treatment of the lower limb amputee.

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Static versus dynamic prosthetic weight bearing in elderly trans-tibial amputees

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Abstract

The purpose of this study was to compare prosthetic weight-bearing tolerance in the standing position to the dynamic vertical ground reaction forces (VGRF) experienced during walking in elderly dysvascular trans-tibial amputees. Ten unilateral trans-tibial amputees attending an amputee clinic (mean age =67±6.5 years) were selected as subjects. Selection criteria were the level of amputation, age, medical fitness to participate and informed consent. Each participant completed five trials of standing (static) weight bearing measurement followed by 10 walking (dynamic) trials on a 10m level walkway, five trials for each limb. Static weight bearing (SWB) was measured using standard bathroom scales. Dynamic weight bearing (DWB) was measured during gait using a Kistler multichannel force platform. T-tests for dependent means indicated that the forces borne in prosthetic single limb stance (mean=0.97±0.03 times body weight (BW)) were significantly lower than the forces borne by the prosthetic limb during the first peak (weight acceptance) VGRF (mean = 1.08±0.08 BW; t = -4.999; p = 0.001) and significantly higher than the midstance VGRF (mean = 0.82±0.07 BW; t = 5.401; p<0.001). However, there was no significant difference between SWB and the second peak (push-off) VGRF generated by the prosthetic limb during walking (mean = 0.96±0.03 BW). It was concluded that clinical gait training may utilise SWB as a guide to an amputees' prosthetic weight bearing tolerance and requirements during walking.

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Introduction

Weight bearing, the act of supporting body weight during standing (static) and walking (dynamic) conditions, has been investigated in amputees using a variety of quantitative methods. Instrumentation has included strain gauges (Tibarewala and Ganguli, 1982), shoe-borne load cells (Lord and Smith, 1984; Ranu and Eng, 1987), force platforms (Hurley *et al.*, 1990; Hermodsson *et al.*, 1994; Thorburn *et al.*, 1990), and bathroom scales (Stolov *et al.*, 1971). Quantifying weight bearing provides an objective measure and numeric feedback of an amputees' prosthetic weight bearing tolerance.

That portion of weight not borne through the prosthesis during the prosthetic stance phase must be taken through the amputee's upper limbs and a walking aid. Assistive devices such as a cane (Winter *et al.* 1993) or a frame (Crosbie, 1992; Pardo *et al.*, 1993) have been used for many reasons: to reduce pressure on sensitive structures, improve balance or assist proprioception (Lord and Clark, 1991). Pardo *et al.* (1993) concluded that walking frames, crutches and canes were used to assist in supporting body mass vertically. While the cane assisted propulsion via vertical force transmitted to the upper limb, the walking frame helped to restrain ambulatory progression with no significant role in propulsion.

Clinically, it is possible to measure weight bearing using bathroom scales. Stolov *et al.* (1971) demonstrated clinical and research utilisation of static standing weight bearing

measurements using bathroom scales. The authors reported the importance of percentage body weight over absolute weight as a criterion when reporting progression of weight bearing after immediate prosthesis fitting following trans-tibial amputation. Summers *et al.* (1978; 1988) used a Double Video Forceplate (DVF) to measure biomechanical parameters of stance and balance. They concluded that simple measurements of weight distribution between the feet and maximal weight-bearing while leaning on the prosthesis could provide objective confirmation of clinical improvement and assist in retraining amputees to stand, balance, and walk.

Assessing weight bearing during standing or balancing tasks, however, is limited to relatively quiet standing situations. Measurements in stance do not assess foot placement, fluctuations in ground reaction forces, nor progression of the centre of pressure experienced during dynamic gait (Winter, 1991; Wintra and Sienko, 1988; Himann and Cunningham, 1988; Engstrom and Van de Ven, 1985). Thorburn *et al.* (1990) assessed ground reaction forces, using a Kistler force platform, generated by five trans-tibial amputees while wearing the SACH, Flex-foot, STEN, SEATTLE, and CC II prosthetic feet. The authors reported that the various prosthetic feet resulted in similar force patterns mediolateral shear force, anteroposterior shear force, and vertical ground reaction force (VGRF) during the stance phase of gait.. The second peak of the VGRF in terminal stance ranged from 97.6% body weight (BW) with the Flex-foot to 99.5% BW with the SACH foot. Andriacchi *et al.* (1977) reported the minimum amplitude of the VGRF (at midstance) to have the largest rate of change with walking speed. Peak VGRF at heel strike and propulsion were considered to exhibit a meaningful velocity dependence. Force amplitudes were found to vary linearly with velocity and were not as sensitive an indicator of gait abnormalities as temporal measurements.

Although assessment of dynamic weight bearing tolerance during gait is preferable to static assessment of weight bearing tolerance, it requires expensive equipment, specialist technicians, and expert interpretation. In contrast, tools to assess static weight bearing tolerance, such as bathroom scales, are inexpensive, readily available to the clinician and provide immediate quantitative data. However, no study was identified in the literature which related single limb static weight bearing tolerance, assessed using bathroom scales, to the forces generated by amputees during gait. Therefore, the purpose of the present study was to compare standing prosthetic weight bearing tolerance to the forces experienced during walking in elderly dysvascular trans-tibial amputees.

Methods

Subjects

Subjects included 10 elderly (mean age = 67±6.5 years) male unilateral trans-tibial amputees, secondary to vascular etiology, recruited from the Illawarra Regional Hospital Amputee Clinic. The subjects reported a mean of 2.8±2.2 years since amputation and had worn their existing prosthesis for an average of 11.8±4.4 months, approximately 12.8±2.4 hours per day. Prosthetic suspension in nine subjects was supracondylar after the fashion of Kegel (1986) and one was by a thigh lacer. Prosthetic foot componentry included seven SACH, two Seattle and one Greissinger. Concurrent disease processes reported by the subjects included six with diabetes mellitus, four had suffered a CVA, one had hypertension, two experienced myocardial infarction, two had osteoarthritis, one gout and one emphysema.

Written informed consent was completed by each subject before testing. All testing was conducted with the knowledge and consent of medical personnel familiar with the subjects' medical history and according to the University of Wollongong Human Research Ethics Committee requirements.

Static weight bearing assessment

Each subject's total body mass was measured (kg) while the subject stood motionless in the anatomical position for 2 seconds on a set of calibrated bathroom scales (130 kg capacity). Prosthetic weight bearing (kg) was then measured while the subjects stood on the scales in single limb stance on their prosthetic limb for 2 seconds. During total body weight and prosthetic weight bearing assessment the subjects were instructed to focus on an eye-level focal point to help them maintain balance. Five trials for each assessment were completed. Due to the need to bear weight through the

prosthetic limb, most subjects required assistance to stand in single limb stance. Assistance was provided by the subjects leaning on a stable fixture placed adjacent to the scale. That upper limb force placed on the support was attributed to weight-bearing intolerance or to diminish balance (Isakov *et al.*, 1992). Differentiating the contribution of weight-bearing intolerance and balance was beyond the scope of this study. The numeric value in kilograms recorded from the scale during unilateral prosthetic stance was therefore defined as the prosthetic weight bearing tolerance. The mean scores for total body weight and prosthetic weight bearing were then substituted into equation (1) to calculate static weight bearing (SWB) tolerance:

$$\text{SWB} = \frac{\text{prosthetic weight bearing}}{\text{total body weight}} \qquad (1)$$

SWB was expressed in body weight units (BW) for ease of comparison with dynamic weight bearing values.

Dynamic weight bearing assessment

Dynamic weight bearing (DWB) was assessed by quantifying the VGRF generated as the subjects walked at a self-selected velocity over a calibrated Kistler multichannel force platform (type 9281B, 600 mm x 400 mm) embedded midway along a 10 m wooden walkway. The force platform was mounted on a concrete pedestal and covered with a sample of granulated rubber sports surface to be level with the surrounding walkway. Forces from the eight output channels of the force platform were passed through a Kistler multichannel Charge Amplifier (type 9865AO) and recorded using a data acquisition board and a personal computer. The VGRF generated by the subjects were sampled (1000 Hz) over the time required for each subject to complete the stance phase on the test limb for five trials per limb. Adequate rest was provided between each trial to minimise fatigue. Before DWB assessment, the weight of each subject was recorded while they stood motionless on the force platform to enable later normalisation of force data relative to each subject's body weight.

Throughout both SWB and DWB assessment, subjects wore their usual walking shoes (eight wore athletic flexible shoes with a deep tread; two wore thin, rigid soled leather shoes.

From each VGRF force-time curve, three values were recorded as representative of DWB (see Figure 1):

(i) the first passive impact peak (1st Peak);

(ii) the dip at midstance (Midstance); and

(iii) the active propulsive peak of push-off (2nd Peak).

These values were expressed in units of body weight. Stance duration was then derived (ms) from initiation of weight acceptance to the completion of push-off.

Kinematics of the gait cycles

The gait of each subject was filmed from anterior and lateral views (25 Hz) using two Panasonic M7 VHS video cameras. The cameras were levelled on tripods 4.8 m (anterior view) and 2.3 m (lateral view) from the subject. The focal axis of each lens was perpendicular to the relevant plane of the force platform to minimise perspective errors (Miller and Nelson, 1973). A known scale was filmed during each trial to enable later conversion of photographic images to actual distance in metres.

Each gait trial was replayed using a Panasonic VHS video cassette recorder and monitor and visually inspected to select one representative prosthetic and normal trial for each subject based on even cadence and lack of

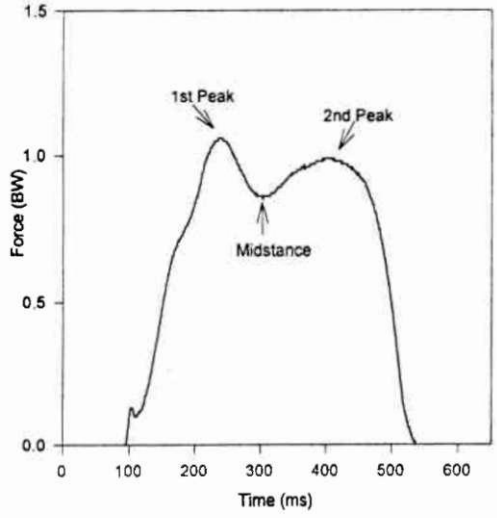

Fig. 1. Typical VGRF force-time curve generated during DWB on the prosthetic limb.

targeting. The video data were transferred from the video recorder to a 496DX-66 personal computer using a video grabber board (Creative Labs VideoSpigot) and Video/Cap software (Video for Windows 1.1). Data were captured (25 Hz) at least 10 frames before heel strike until 10 frames after toe-off to the test limb. The x and y coordinates for 15 anterior markers (vertex, clavicular acromion, humeral lateral epicondyle, radial styloid process, mid-inguinal crease, patellar superior pole, midway between lateral and medial malleoli, anterior tip of shoe; left and right sides) and 9 lateral markers (greater tronchanter, fibular head, lateral malleolus, fifth metatarsal head and tip of shoe on facing leg, femoral adductor tubercle, medial malleolus, first metatarsal head and tip of shoe on contralateral leg) representing the anatomical link system were digitised throughout the stance phase of the test limb. The digitised data were then smoothed using a fourth order low pass Butterworth digital filter (6 Hz cut-off frequency) (Winter, 1990).

The following variables were then calculated during the stance phase of the test limb from the smoothed digitised data using Digital Signal Processing (DSP) software (Andrews, 1994):

(i) velocity of forward progression, represented as the average linear velocity of the greater tronchanter marker in the sagittal plane from weight acceptance to push-off (stance phase);

(ii) lateral (weight) transference of the total body centre of gravity (COG). The total body COG was calculated utilising custom software based on Dempster's linked segment model (Winter, 1990). Lateral weight transference was defined as the horizontal displacement (cm) in the frontal plane of the COG relative to the base of support at midstance. A smaller displacement indicated less lateral weight transference of the subject's COG toward the base of support.

Statistics

Means, standard deviations and ranges were calculated for each kinetic and kinematic variable for the gait trials performed on each subject's sound and prosthetic limb and for SWB. T-tests for dependent means were then conducted (SYSTAT 5.01 for Windows) to determine any significant differences ($p<0.05$) between the forces generated during SWB and DWB. Pearson product moment correlations were also calculated to identify any significant relationships among the forces generated during SWB and gait and the gait descriptions.

Results

SWB versus DWB

The mean, standard deviation and range of values recorded during SWB and DWB are presented in Table 1. Forces borne through the prosthesis during SWB[1] (mean = 0.97±0.03 BW) were significantly lower than the forces borne by the prosthetic limb during the first passive peak VGRF (mean = 1.08±0.08; $df = 9$; $p = 0.001$) and were significantly higher than the forces borne during midstance (mean = 0.82±0.07; $df = 9$; $t = 4.999$; $p<0.0001$). However, there was no significant difference between SWB forces and the second propulsive peak VGRF (mean = 0.96±0.03 BW; $df = 9$; t =

[1]SWB forces of less than 1 indicated that the subjects required external support while standing on their prosthetic limb.

Table 1. Forces (BW) recorded during SWB and DWB (n=10).

	SWB*	Prosthetic limb DWB			Sound limb DWB		
		1st Peak	Midstance	2nd Peak	1st Peak	Midstance	2nd Peak
Mean	0.97	1.08[1]	0.82[1]	0.96	1.19	0.81	1.03
SD	0.03	0.08	0.07	0.03	0.17	0.09	0.06
Min	0.92	0.99	0.68	0.93	0.99	0.66	0.96
Max	0.99	1.25	0.90	1.01	1.48	0.93	1.15

*SWB was calculated for the prosthetic limb only. Therefore, comparisons of SWB and sound limb DWB were not conducted.

[1] indicates a significant difference ($p<0.001$) between SWB and DWB values on the prosthetic limb.

Table 2. Horizontal velocity, stance duration and lateral weight transference during the stance phase on the prosthetic and sound limbs (n=10).

	Velocity (m.s.$^{-1}$)		Stance duration (ms)		Lateral weight transfer (cm)	
	Prosthetic	Sound	Prosthetic	Sound	Prosthetic	Sound
Mean	0.52	0.51	483	499	3.82	4.49
SD	0.18	0.15	214	224	0.96	1.33
Min	0.19	0.20	353	366	2.83	2.03
Max	0.74	0.67	1064	1113	5.60	6.06

0.629; p = 0.545) generated by the prosthetic limb during walking.

Strong significant negative correlations were found between the first passive peak VGRF and midstance VGRF for both the sound limb (r = -0.950) and the prosthetic limb (r = 0.860). This result indicated that a high first VGRF peak was associated with a low midstance VGRF.

Gait Descriptors

The mean, standard deviation and range of values calculated for the three kinematic gait descriptor are presented in Table 2. There was no significant difference between the prosthetic and sound limbs for the velocity of forward progression during the stance phase, stance duration or lateral weight transference onto the stance limb.

For SWB to be used as a predictor of walking ability, the relationship between SWB and dynamic gait descriptors and must be established. Correlation coefficients calculated between the gait descriptors and the forces recorded during SWB and DWB are presented in Table 3. Velocity of forward progression during the stance phase of the sound limb was significantly correlated to SWB (r = 0.680), indicating that the higher forces borne through the prosthesis during SWB were associated with greater velocity in the sound limb during gait. This relationship was not significant during the stance phase on the prosthetic limb (see Table 3). Velocity was also found to be significantly correlated to DWB at midstance for the prosthetic limb (r = -0.699) and the sound limb (r = -0.781). Therefore, greater velocities were associated with decreased VGRF at midstance. Although velocity was significantly correlated to DWB at the first peak VGRF on the sound limb (r = 0.735), it exhibited only a low positive relationship with DWB on the prosthetic limb at the first peak VGRF (r = 0.450). Velocity exhibited only a low negative relationship with the DWB at the second peak on the prosthetic and the sound limb (see Table 3). Stance

Table 3. Correlation coefficients between the three gait descriptors and SWB and DWB.

	Velocity (m.s.$^{-1}$)		Stance duration (ms)		Lateral weight transfer	
	Prosthetic (n=10)	Sound (n=10)	Prosthetic (n=10)	Sound (n=10)	Prosthetic (n=9)	Sound (n=10)
SWB[1]	0.472	0.680*	-0.649*	-0.624	0.241	0.237
DWB						
1st Peak	0.450	0.735*	-0.513	-0.565	0.132	-0.004
Midstance	-0.699*	-0.781	0.509	0.585	-0.193	-0.091
2nd Peak	-0.220	0.353	-0.233	-0.182	0.060	0.455

[1]SWB was correlated to the three gait descriptors to establish the relationship between the static weight bearing ability and dynamic gait.
*indicates a significant correlation at p<0.05 (r>0.632; *df* = 8)

duration was found to be significantly correlated (r = -0.649) with SWB on the prosthetic limb, indicating longer stance duration on the prosthetic limb was associated with lower forces borne through the prosthesis during SWB. No other significant correlations were found between SWB or DWB and either stance duration or lateral weight transference during stance for the prosthetic or sound limb.

Discussion

The VGRF recorded during free walking in the present study were consistent with those reported previously for trans-tibial amputee patients (Thorburn *et al.*, 1990). The bi-peaked vertical force-time curve characteristic during gait indicated a weight acceptance force greater than 1 BW, a VGRF less than 1 BW at midstance, followed by a second peak VGRF approximately 1 BW at push-off of the stance phase.

The mean velocity of forward progression recorded in the present study was less than mean velocities reported in previous studies of amputee gait (Hurley *et al.*,1990; Thorburn *et al.*, 1990; Winter, 1991). However, there was a large range of velocities demonstrated in the present study (Table 2). Those subjects displaying lower velocities of forward progression during stance also exhibited less defined peaks and troughs in their VGRF force-time curves. The weight bearing tolerance of the high peaks were held for short durations whereas the lesser forces were tolerated longer.

The mean values of velocity, stance duration and lateral weight transference were not significantly different when comparing the prosthetic limb to the sound limb in the small sample of subjects in the present study. In contrast, Thorburn *et al.* (1990) found stance duration to be significantly longer for the sound limb than the prosthetic limb.

The SWB test provided a practical quantitative assessment of the weight bearing tolerance of the amputee subjects on their prosthetic limb, a test suitable for use in clinical settings. Static weight bearing in single limb stance can be measured quickly and easily using bathroom scales and can be expressed as a percentage of total body weight. The functional task of shifting the amputee's weight onto the prosthesis to step onto the scale demonstrated an objective task reflecting patient confidence and comfort. Walking aid selection and gait training goal setting may be guided by the percentage of prosthetic weight bearing as a predictor for further training.

Progression of weight bearing from dual limb to single limb stance can prepare an amputee for the VGRF experienced during midstance and push-off phases of gait. In the present study, SWB was shown to be less than the VGRF generated dynamically at weight acceptance by the prosthetic limb. It would therefore be insufficient to train tolerances equivalent to SWB when attempting to prepare patients to meet demands anticipated for weight acceptance at self-selected walking velocity. The SWB test provided more prosthetic weight bearing than that exhibited dynamically during midstance, particularly in the faster walkers. Therefore, training weight bearing tolerance in unilateral stance to full weight bearing would be adequate to enable patients to cope with the VGRF generated during distance in self-selected walking velocity. Furthermore, the SWB test demonstrated equivalent VGRF to that measured at push-off. Training amputees to tolerate full weight bearing in unilateral stance on the prosthesis would therefore prepare them for the push-off phase of gait.

The DWB first peak demonstrated a positive relationship with velocity. The greater the velocity, the greater was the VGRF at weight acceptance. This first peak was also greater than the subjects' SWB measurement. In the elderly population, however, velocity of gait diminishes with increasing age. Therefore, the need to train prosthetic weight bearing to cope with high first peaks will lessen with age. Others, whose concurrent medical problems slowed them further (Himann and Cunningham, 1988; Mueller *et al.*, 1994), displayed a first peak equivalent to their body weight. In these slower individuals, the SWB value was equivalent to both the first and the second peaks of the force-time curve of the VGRF recorded during DWB.

Hermodsson *et al.* (1994) reported that the gait of vascular amputees differed from that of traumatic amputees, a difference that was not caused by reduced walking speed. The active forces during push-off on both the sound and the prosthetic leg in the trauma group were not found in the vascular group. This disparity could be an effect of the systemic disease. No distinction was made in the present study for

the underlying pathology of weight bearing intolerance, such as stump sensitivity, ill fitting socket, habit; nor concurrent medical diagnoses, such as CVA, hip replacement, and myocardial infarction. All amputees in this study had vascular etiology.

Clinical application of the results of the present study lies in the hierarchy of the weight bearing demands that were identified. Midstance requires less than body weight tolerance, push-off requires full body weight tolerance though the prosthesis and weight acceptance requires tolerance of forces in excess of full body weight. Rehabilitation may proceed to increase velocity when weight acceptance forces have surpassed 100% body weight. It is therefore recommended that rehabilitation goals for new amputees should include techniques to train prosthetic weight bearing first in bilateral stance, progress to push-off from the prosthesis and finally to weight acceptance onto the prosthesis.

In conclusion, the SWB test provides a quantitative test of weight bearing tolerance of the trans-tibial amputee on the prosthetic limb. Weight acceptance during the stance phase of DWB required a greater force tolerance than SWB. The midstance phase, however, required less force tolerance than SWB and the propulsive phase required equivalent forces to those measured in SWB. Clinical gait training may therefore utilise SWB as a guide to an amputee's prosthetic weight bearing tolerance and requirements during walking.

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A new biomechanical method for determination of static prosthetic alignment

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Abstract

A new static alignment method for trans-tibial prostheses is suggested using the individual load line as a reference.

Standing posture and static alignment of 18 experienced trans-tibial prosthetic users with good walking ability were determined and compared with 20 healthy persons. The individual load line was defined by means of the new Otto Bock alignment system "L.A.S.A.R. Posture".

The sagittal standing posture of trans-tibial amputees and non-amputees differs. Normally only a prosthesis worn by the trans-tibial amputee and dynamically aligned over an extended period of time satisfies biomechanical rules of alignment. In contrast, prostheses aligned during one session in the traditional subjective manner seem to lack any recognizable biomechanical systematics. Initial results suggest the knee centre should be 10 to 30mm behind the load line, depending on patient's weight. This knee position is independent on the type of the prosthetic foot.

Introduction

The quality of rehabilitation of trans-tibial amputees with prostheses is influenced by different factors. From the prosthetic or biomechanical point of view at least four important inter-related factors can be noted: the prosthetic socket, the type of the prosthetic foot selected, prosthetic alignment (Pinzur *et al.*, 1995; Solomonidis, 1991), and the integration of the prosthesis into the amputee's motor activity.

The present article deals with prosthetic alignment. Preliminary results from trans-tibial fittings will be presented in which the static alignment was not based on a fixed reference line but related to the measured load line.

From clinical practice it is known that the optimization of prosthetic alignment can take several weeks from the first dynamic alignment to the final definitive fitting. The dynamic optimization of the prosthesis is a very time-consuming and subjective process requiring excellent skills and many years of experience of the prosthetist. In dynamic alignment the practitioner must rely on his visual perception during the gait trials as well as feedback from the amputee, along with his experience, to refine the alignment interactively (Zahedi *et al.*, 1986).

Although many manufacturers of prosthetic components give static alignment recommendations from clinical experience using theoretical alignment reference lines, these general guidelines do not reflect individual differences. Furthermore, there is a worldwide controversy regarding different alignment guidelines (Radcliffe, 1994; West, 1987). The possibility of applying biomechanically determined measurements to prosthetic alignment is the subject of this paper.

Method

Static prosthetic alignment was determined using the Otto Bock alignment system "Laser Assisted Static Alignment Reference (L.A.S.A.R.) Posture". Figure 1 shows a trans-

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Fig. 1. Measuring the static alignment of a trans-tibial amputee with the "L.A.S.A.R. Posture" system.

tibial amputee standing on the alignment platform.

The "L.A.S.A.R. Posture" alignment system measures the vertical component of the ground reaction force acting on the force plate of the platform. Thus the patient's weight and the location of the weight bearing line in static standing with both feet on the force plate can be determined. If only one side, e.g. the prosthetically fitted limb, is standing on the force plate, the force of that side and the resultant load line will be measured.

In addition to the force plate, the apparatus contains a projection system, electronics with a stepper motor, and the service and display unit (Fig. 2). The force plate includes 4 sensor cells located in the corners of the force plate. The microprocessor determines the centre of pressure and the amount of ground reaction force. The electronics triggers the stepper motor whereupon it moves a semiconductor laser to the centre of the measured forces. Optics located in front of the laser convert the pinpoint laser beam into a bright line. The laser line is then projected on the person being measured illustrating where the centre of pressure is located. Thus, the location of the vertical ground reaction force is visibly indicated on the amputee.

If the distance between a certain point of the body and the load line is to be measured, the laser beam can be moved to this position by pressing a button on the service unit. The distance between this body point and the load line, and the vertical force vector are indicated on the display.

In prosthetics the distance between components and the load line in the sagittal plane is useful for the static alignment. For measuring posture as well as alignment, distances of the following points from the reference line, shown in Figure 3, were recorded: the middle of the shoulder, the greater trochanter, the knee centre – defined by Nietert (1977), and the lateral malleolus.

In the frontal plane, the distance to the load line was determined from the middle of the ankle joint and knee joint.

Patients

The alignment measuring process was performed on 18 trans-tibial amputees. The posture of the amputees and static prosthetic alignment were determined on the definitive, long-term prostheses customarily worn by the patients as well as on the test prostheses. The test prostheses used the same socket as the definitive prostheses; only the prosthetic feet were changed. All test prostheses were dynamically aligned by an experienced prosthetist. Several different prosthetic feet were tested in succession.

During the alignment measuring process the patients wore their customary shoes.

Tables 1, 2 and 3 list patient's data, amputation cause, and the foot types.

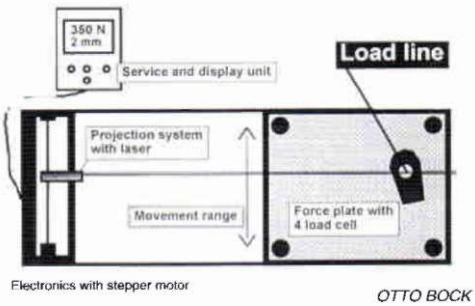

Fig. 2. Elements of the "L.A.S.A.R. Posture" alignment system.

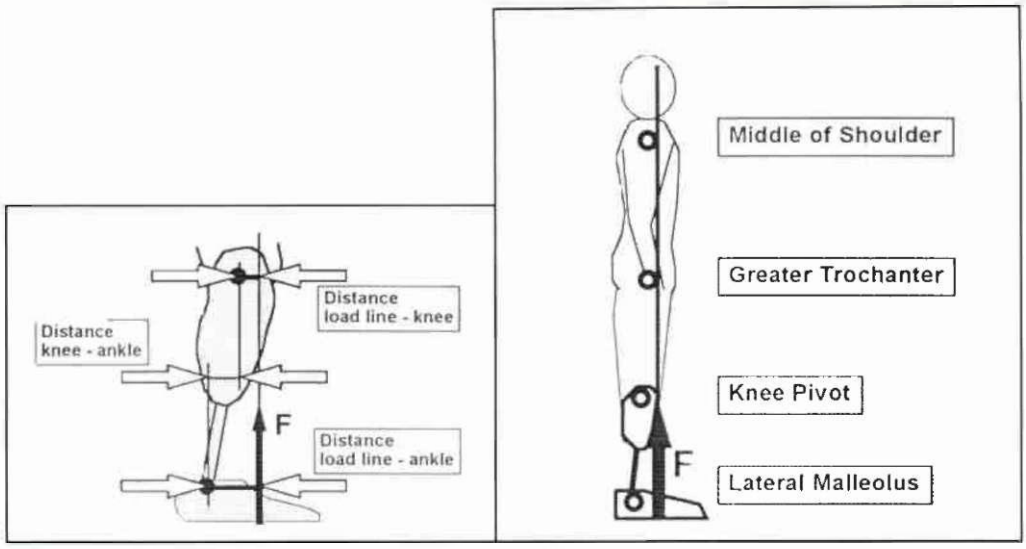

Fig. 3. Sagittal plane reference points for the trans-tibial amputee, F-ground reaction force.

For this investigation, experienced users were selected who had worn a prosthesis for many years and could walk a significant distance. For this reason, amputees with circulatory impairment were not included in the investigations. The amputees had comfortable sockets and no joint pain or range of motion abnormalities.

For reference, 20 non-amputees aged 17 to 53 years were also measured.

Results

Posture in the sagittal plane

Measurements in the sagittal plane from the middle of the shoulder, the greater trochanter, the knee joint centre and the lateral malleolus are shown in Figure 4.

The measured data show clear differences between the posture of non-amputees and the prosthetically fitted side of trans-tibial amputees. The posture of the non-amputated

Table 1. Amputation cause

Trans-tibial amputees	
Number:	18 trans-tibial amputees
Aetiology:	13 Trauma
	(2 as little child)
	3 Malformation
	1 Tumour
	1 Other

Table 3. Prosthetic feet of the definitive fitting of the trans-tibial amputees

Amputee's habitual foot type		
Number:	18 trans-tibial amputees	
Foot size:	24-28	
Foot type:	Dynamik 1D10	9
	Dynamic pro 1D20	2
	Quantum foot	1
	Multiflex	3
	Flexwalk	3

Table 2. Patient's data

Trans-tibial amputees Number: 18 trans-tibial amputees				
	Age(y)	Body mass(kg)	Height (cm)	Amp. period(Y)
Mean	39.0	77.5	177	16.9
Stand. dev.	8.7	13.8	13	10.0
Minimum	18	57	163	2
Maximum	65	106	190	36

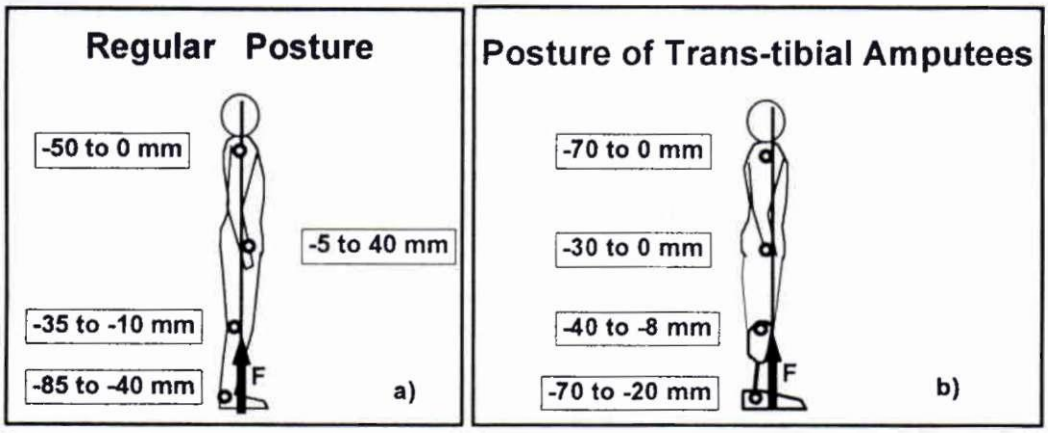

Fig. 4. Range of the positions from the middle of the shoulder, greater trochanter, knee centre and lateral malleolus; a) non-amputees, b) trans-tibial amputees with their definitive prosthesis.

side of the trans-tibial amputee is similar to that of the non-amputee.

The altered position of the hip joints seems to be most significant. Instead of a slightly anterior position of the greater trochanter with reference to the load line typical of the non-amputee, the trans-tibial amputee shifts his posture so that this anatomical point falls behind the load line (Fig. 5).

Figure 6 indicates the measured mean values and standard deviations of the measured points for all 18 trans-tibial amputees with their definitive prosthesis. The average posterior position of the ankle with reference to the load line is 50mm, that of the knee 18mm, that of the greater trochanter 16mm, and that of the shoulder 35mm.

In the frontal plane, the load lines of the trans-tibial amputee as well as of the non-amputee run approximately through the middle of the ankle joint and the knee joint of each leg.

To investigate possible interdependencies of the different joint positions, the linear coefficients of correlation were measured in pairs. For the non-amputees no relationship could be discerned. Table 4 shows on the other hand, that the trans-tibial amputees had a statistically significant correlation between the knee and ankle position, with a coefficient of correlation of R=0.53.

According to the linear correlation analysis, there are no interdependencies between the

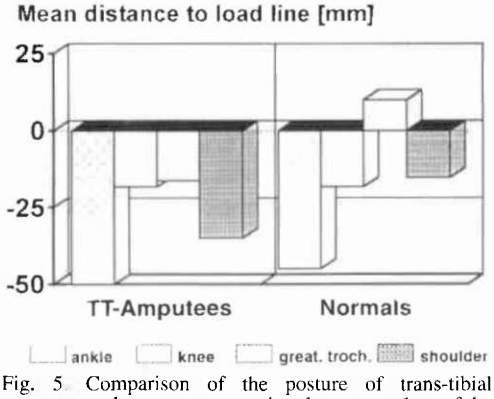

Fig. 5. Comparison of the posture of trans-tibial amputees and non-amputees using the mean values of the sagittal position of ankle, knee, greater trochanter and shoulder.

Table 4. Coefficients of correlation of the different joint positions for trans-tibial amputees

	Knee	Greater trochanter	Shoulder
Ankle	0.53	0.06	-0.31
Knee		-0.04	-0.08
Greater trochanter			0.22

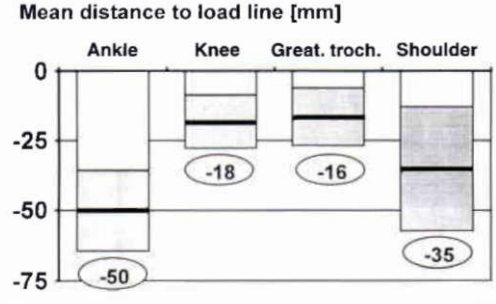

Fig. 6. Mean value and standard deviation of the position of the shoulder, the greater trochanter, the knee joint and the lateral malleolus of the trans-tibial amputees.

Table 5. Coefficients of correlation for the different joint positions versus body mass and height for trans-tibial amputeees

	Body mass	Height
Ankle	-0.20	-0.06
Knee	-0.44	0.01
Greater trochanter	-0.26	-0.12
Shoulder	0.01	-0.04

Table 6. Coefficients of correlation for the knee joint-ankle-distance versus the joint positions, body mass and body height for trans-tibial amputees

	Knee-Ankle
Ankle	-0.75
Knee	0.17
Greater trochanter	-0.10
Shoulder	0.30
Height	0.09
Body mass	-0.11

body height and the joint position for trans-tibial amputees. However, body mass has a significant influence on the distance between the knee and the load line (R= –0.44). The heavier the trans-tibial amputee, the greater the posterior position of the knee with reference to the load line (Table 5).

Static prosthesis alignment

Static prosthetic alignment in the sagittal plane is reflected in the inclination of the pylon and determined by the horizontal distance between the knee centre and the ankle.

According to the scatter diagram of Figure 7, the horizontal distance between the knee centre and the load line is independent of the horizontal distance between the knee and the ankle. The knee position with reference to the load line, and thus the acting lever of the ground reaction force at the knee joint, is independent on the characteristics of the prosthetic foot.

On the other hand, the ankle distance from the load line shows a highly significant relation to the distance between knee and ankle (R= -0.75). The sagittal plane inclination of the below-knee pylon is therefore determined primarily by the position of the foot.

Furthermore, Table 6 verifies that prosthetic alignment is neither correlated with the body measures nor with the hip position nor with posture of the upper body. The distance between knee and ankle of the trans-tibial prosthesis is influenced during standing by the variation in foot anteroposterior shift.

The distance ankle-knee, and thus the ankle position, is finally correlated with the foot type, as Figure 8 describes.

Further parameters correlating with the foot type used by the trans-tibial amputee could not be identified.

Following changes in the components of the prosthetic foot optimal prosthetic alignment assessed subjectively during one test session only did not reveal any recognizable alignment relationships.

Discussion

The lengthy and difficult process of obtaining a qualitatively satisfactory prosthetic alignment led to consideration of the fitting methodology. It is obvious that present prosthetic alignment recommendations differ from each other

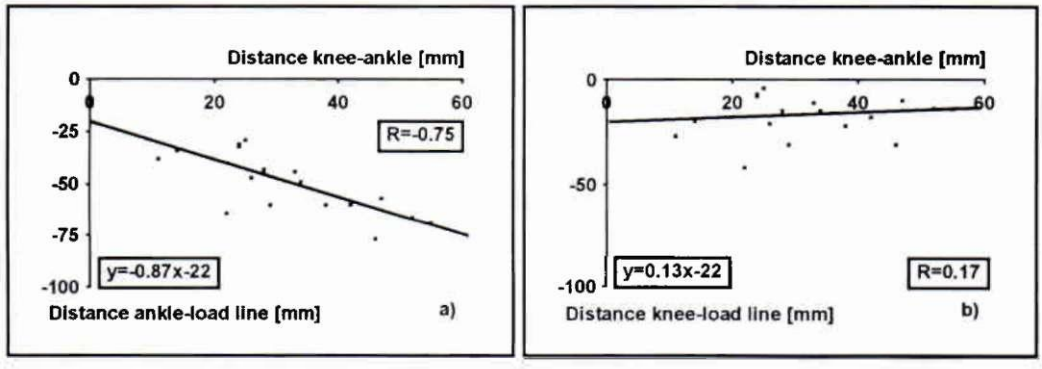

Fig. 7. Scatter diagram between the knee-ankle-difference and a) the ankle position and b) the knee position for trans-tibial amputees.

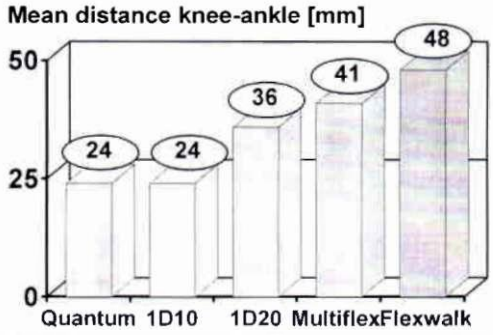

Fig. 8. Mean ankle-knee distance for the different foot types.

considerably (e.g. TKA-line, German alignment). From the biomechanical point of view, this can be understood only if prostheses contain different components. These different components have different characteristics leading to different properties during walking, e.g. different dorsal resistances of the feet in the temporal course of the gait cycle. If prostheses contain the same components and the same socket, a different biomechanical alignment with the same functional result cannot be explained. Also, in clinical practice it can be noted that prosthetic alignments become more and more similar after a longer period for acclimatization even if the prostheses were assembled before dynamic alignment according to different static alignment recommendations (Marmaras and Bach, 1995; Radcliffe, 1994; West, 1987).

Forces and moments are not visible. However, they define the fundamental function of a prosthesis during standing and walking. For optimization of prosthetic alignment, visual information about the force and moment situation seem to be desirable for the prosthetist (Wilson *et al.*, 1979). With the "L.A.S.A.R. Posture" alignment system, the vertical ground reaction force can be precisely determined and indicated on the standing patient. Measuring the static alignment then becomes possible. Forces and lever arms can be considered. This offers a new biomechanical insight; prosthetic alignment is not exclusively dependent on prior custom. With the "L.A.S.A.R. Posture" alignment system a tool is now available for making prosthetic alignment or individual posture visible in the selected plane. The basis is the vertical ground reaction force acting on the leg during standing.

Posture measurement of non-amputees shows that the ground reaction force falling along the middle of the ankle, the knee centre, the greater trochanter and the middle of the shoulder often described in the literature is not correct.

In addition, it appears that prostheses with different feet do not show any alignment consistency after brief alignment trials. Prostheses worn by the amputee for a longer time seem to have more consistent alignments. This is supported by practical experience showing that prostheses can be optimally aligned using subjective methods only over a longer period of time. Thus, the amputee is not able after only a short wearing time to evaluate the quality of the alignment (Solomonidis, 1991; Zahedi, 1986). Possibly this new procedure could offer a starting point for achieving a good fitting result in a more verifiable way and in a shorter time.

The results thus far encourage continued testing of these ideas in practice to refine the results presented here. Also applications for other areas of practice are worth considering: orthoses, compensation for leg lengths discrepancy, etc.

Non-amputees maintain their balance during standing using ankle control. The ground reaction force running behind the hip joint allows stabilization of the hip joint during standing by ligament or muscular force acting anterior to the joint. The trans-tibial amputee uses the gluteal and the posterior thigh musculature during standing for controlling stability and balance of the knee joint. Therefore, the trans-tibial amputee has not only altered gait characteristics (as known from clinical gait analysis) but, compared with the non-amputee, also different posture balance.

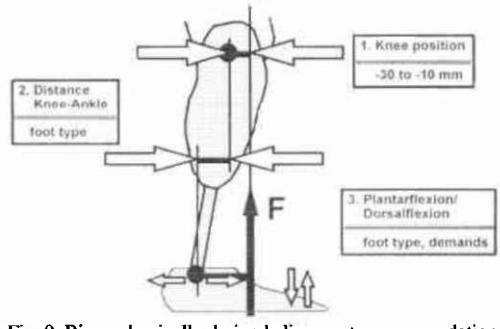

Fig. 9. Biomechanically derived alignment recommendation for trans-tibial prostheses

Conclusions

The present investigation attempted to determine prosthetic alignment for trans-tibial amputees biomechanically. As the investigation was made on a relatively small number of amputees, it requires further verifications of the results presented. However, it has suggested a new static alignment method for trans-tibial amputees' prostheses, as outlined in Figure 9:

1. determine the load line on the prosthetic side remembering that the knee centre should be located 10 to 30mm behind the load line;
2. the posterior position of the ankle (distance ankle-knee) is made depending on the foot selected (compare with Figure 8);
3. dorsal and plantar flexion of the foot have to be adjusted depending on both foot type and patient's weight (individual requirements) considering the knee location.

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The effect of changing the inertia of a trans-tibial dynamic elastic response prosthesis on the kinematics and ground reaction force patterns

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Abstract

The aim of this study was to assess, by means of gait analysis, the effect on the gait of a trans-tibial amputee of altering the mass and the moment of inertia of a dynamic elastic response prosthesis. One male amputee was analysed for four to five walking trials at normal and fast cadences, using the VICON system of motion analysis and an AMTI force plate. The kinematic variables of cadence, swing time, single support time and joint angles for the knee and hip on the affected and intact sides were analysed. The ground reaction force was also analysed. The sample size was limited to one as an example to indicate the changes which are possible through simply changing the inertial characteristics. Descriptive statistics are used to demonstrate these changes. Three mass conditions for the prosthesis were analysed m_1: 1080g; m_2: 1080 + 530g; m_3: 1080 +1460g. The m_1 condition is the mass of the prosthesis with no added weight while m_2 and m_3 were attachments of the same geometrical shape but were made from different materials. It was felt that the large mass range would highlight biomechanical adjustments as a result of its alteration. The effect on selected temporal characteristics were that as the speed increased the cadence changed and the affected side single support times as a percentage of the gait cycle were altered. The effect on the joint angles was also apparent at the hip and knee of both sides. The ground reaction force patterns were similar for all three mass conditions, though the impact peak which was evident in the intact limb was missing, indicating a shock absorbing property in the prosthesis. Clearly, changing the mass and moment of inertia has an effect on the kinematic variables of gait and should be considered when designing a prosthesis.

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Introduction

Trans-tibial amputees generally walk more slowly than their normal counterparts and the timing of the gait cycle for both trans-femoral and trans-tibial amputees is not symmetrical (Winter and Sienko, 1988; Colborne *et al.,* 1992, Barr *et al.*, 1992). Recent developments in prosthetic research have resulted in new devices known as Dynamic Elastic Response (DER) prostheses which are claimed by their manufacturers to have superior performance characteristics in gait. However this is a controversial claim with some studies noting limited biomechanical improvements. Macfarlane *et al.* (1991) observed improved kinematic symmetry in the timing of the gait cycle while others comment on the fact that these prostheses are expensive or do not offer any significant biomechanical or energy conservation benefits to the active amputee (Gitter *et al.,* 1991; Colborne *et al.,* 1992; Anzel *et al.,* 1992-1993; Perry and Shanfield, 1993). There are limited studies on the effect of changing the mass and moment of inertia on this

type of prosthesis, although it is generally understood that the inertial characteristics of a lower limb prosthesis will affect the gait of the amputee. Simulations on trans-femoral amputee models have indicated that altered inertial characteristics of the prosthesis may affect and contribute to the asymmetries involved in amputee gait (Bach *et al.,* 1993), while biomechanical studies on trans-femoral amputees have indicated that altering the inertial characteristics of the prosthesis will have a significant effect on the gait. This leads to the suggestion that these prostheses should be designed to minimise the distal weighting of the prosthesis (Tashman *et al.,* 1985) with the additional note that simply changing the mass of the prosthesis would not be sufficient to alter the swing period of the shank, rather, the centre of mass must also be moved, with a proximal move resulting in a faster swing in the lower limb. Jans and Bach (1995) advise that in the prescription of trans-tibial (TT) prostheses the inertial characteristics should be considered as the amputees analysed subjectively preferred the lighter mass and walked with increased velocity. In normal cadence walking, the swing phase is basically a passive action due to the pendular effect of the swinging limb but when the cadence is changed then the knee flexors are involved in the flexion of the knee in swing (Gage, 1991). In a study assessing the effect ankle weighting would have on normal subjects, Hale (1990) noted that asymmetrical weighting affected walking with a reduction in the single limb support time, increased swing phase and decreased stance phase for the weighted limb. Inman (1967) observed that trans-tibial amputees need a light prosthesis in early swing, so that with the reduced moment of inertia, the prosthesis could be brought to accelerate into swing easily, but that in late swing, the prosthesis should be heavy and thereby with its increased inertia would increase the acceleration and energy supplied to the system. Research on the affect of changing the inertial characteristics for different cadences is lacking in the literature. This raises the question as to the optimal mass and mass distribution of a prosthesis which may be used for different cadences. The purpose of this investigation was to assess the effect of altering the inertial properties of the prosthesis on the temporal, kinematic and ground reaction force variables of the gait cycle.

Methods

One young male amputee (body mass: 79kg, height: 1.87m, age: 24y) was assessed. An experimental DER prosthesis was fitted to a patellar-tendon-bearing (PTB) socket. This prosthesis allowed for the inertial characteristics of the prosthesis to be altered by adding a weight to the distal foot portion. The original mass of the prosthesis was m_1 1080g. Additional masses of m_2 (530g) and m_3 (1460g) were attached to the prosthesis. They were the same geometrical shape but made from different materials, m_2 was aluminium and m_3 was steel. It was felt that a wide range in the added mass would highlight the biomechanical alterations in the gait of the amputee. The addition of the masses had the effect of moving the centre of the mass of the prosthesis distally from 32.1cm, to 33.11cm for m_2 and 39.58cm for m_3. Since the shape of the prosthesis was such that the centre of gravity (CoG) of the prosthesis was located outside of it, the position of the CoG was estimated using the reaction board technique. The moment of inertia of the prosthesis was also altered from 0.156kgm^2 (m_1) to 0.245kgm^2 (m_2) and 0.53 kgm^2 (m_3). The moment of inertia of the prosthesis wascalculated as follows: the period (T) of the prosthesis was estimated by swinging the prosthesis for 10 periods from an almost frictionless pin at the position of the knee joint axis. One period for each of 10 trials was calculated and the average taken. From this the moment of inertia was calculated from:

$$I = (WhT^2)/(4\pi^2)$$

where I is the moment of inertia about the axis of the prosthesis at the knee joint; W is the weight of the prosthesis in air; h is the distance from the centre of mass to the suspension axis and T is the period of 1 oscillation.

The gait analysis on the subject was carried out using a five camera VICON motion analysis system (Oxford Metrics Ltd. Botley, Oxford, UK) and an AMTI force platform linked to an Etherbox data acquisition system and a host computer. The subject walked at each of a self-selected normal and a fast walking cadence along a 12m walkway. The subject was recorded walking through a 0.65m (X:medial/lateral direction) x 2.45m (Y: anterior/posterior direction) x 1.53m (Z: vertical direction) calibrated volume. A single force plate was located in the centre of the calibrated

volume. A trial was deemed to be successful if the subject's foot landed on the plate without any alterations to the gait pattern. Five trials for the affected and five trials for the intact sides were recorded for both cadences. During testing with m_3, fatigue was becoming evident after the collection of 4 trials at fast cadence and testing was terminated. Trials were ensemble averaged to allow comparisons across trials. One stride from heelstrike to the following ipsilateral heelstrike was deemed 100% and the averaging was carried out at the 2% level which has been deemed acceptable for walking trials (Winter, 1987). The subject was allowed 15min to become accustomed to each new mass condition before testing was resumed. The subject was required to wear shorts and reflective markers were placed on the subject's skin on the sacrum, the left and right anterior superior iliac spines, the lateral femoral condyle of the right knee, the right lateral malleolus, the head of the right second metatarsal, along the thigh in line with the greater trochanter and the femoral condyle and on the shank in line with the femoral condyle and the malleolus. On the prosthetic (left) side, markers were placed on the prosthesis lateral to the left femoral condyle, at the estimated position of the malleolus on the lateral distal aspect of the prosthesis, the estimated position of the top of the second metatarsal and along the thigh and shank as for the intact limb. Cadence, stance time, swing time, single support time, joint angles for the knee and hip on the affected and intact sides and the three components of the ground reaction force were analysed. The subject wore the same shoes for all conditions. The ankle was not assessed as the addition of the weights did not affect the prosthesis kinematics at the ankle.

Results

Stride characteristics

The means and standard deviations, in brackets, for the walking speed, cadence, stride length single-limb support and swing periods are presented in Table 1 for normal and fast walking trials.

Altering the inertial parameters of the prosthesis showed minimal effect on self-selected normal walking cadence. However, cadence at the fast condition was observed to decrease with each increase of mass.

The velocity of the subject over the plate at natural cadence varied with the alteration of the mass, with the velocity increasing as the mass increased, while at the fast cadence the velocity was consistent, with little change across the different mass conditions.

The stride increased for the faster walking condition and also increased as the mass was increased.

Single limb support decreased as the mass increased for the affected limb (left) for both cadences and concurrently, the single limb support for the intact limb increased with the increasing mass on the affected side. Single

Table 1. Speed, cadence, stride length and single support time for the three mass conditions walking at normal and fast cadences

	m_1		m_2		m_3	
	Normal (σ)	Fast (σ)	Normal (σ)	Fast (σ)	Normal (σ)	Fast (σ)
Speed (m/s)	1.47 (0.03)	1.86 (0.03)	1.57 (0.04)	1.89 (0.06)	1.61 (0.03)	1.86 (0.04)
Cadence (step/s)	86.5 (0.01)	97.1 (0.01)	87.1 (0.01)	96.0 (0.02)	84.4 (0.02)	93.2 (0.01)
Stride length (mm)	1755 (20)	1912 (33)	1811 (31)	1973 (35)	1909 (32)	2001 (35)
Single support left (% cycle)	40.49 (2.2)	39.81 (2.8)	41.58 (3.1)	38.4 (2.3)	39.62 (5.1)	37.84 (3.9)
Single support right (% cycle)	39.87 (2.6)	41.6 (2.6)	41.35 (2.3)	40.83 (3.1)	42.04 (1.5)	41.19 (5.2)

limb support was asymmetrical for both cadences and the extent of the asymmetry was affected by the mass condition. Using the lightest prosthesis, the intact single limb support was closest to that of the prosthetic single support time for fast cadence, however, for the normal cadence m_2 produced the condition where the single support was closest for both sides.

The graphs of the joint angles in Figures 1 and 2 are for one stride period, from initial contact to initial contact of the same limb. The graphs illustrate the means for the trials collected standard deviation bars are reported in the text for the key instances.

The sagittal plane hip joint angles for the intact and affected limbs for the different mass conditions are shown in Figure 1 for the normal and fast cadences. Knee joint angles are shown for the normal and fast cadences in Figure 2.

Hip joint angle

The affected hip flexion/extension pattern for the different masses was similar. There is between 40° and 50° of flexion at initial contact depending on the mass (m_1: 47° (1.2); m_2: 49° (1.4); m_3: 41° (0.7)). Prior to toe off the hip is in full extension for the lesser mass conditions, m_1 and m_2. For the greatest mass (m_3), the hip is hyperextended to a maximum of 11° (0.8). At mid-swing, the hip showed greatest flexion for m_2, and was most extended for m_3.

The mass also has an effect on the hip angle of the limb and the patterns are similar to the

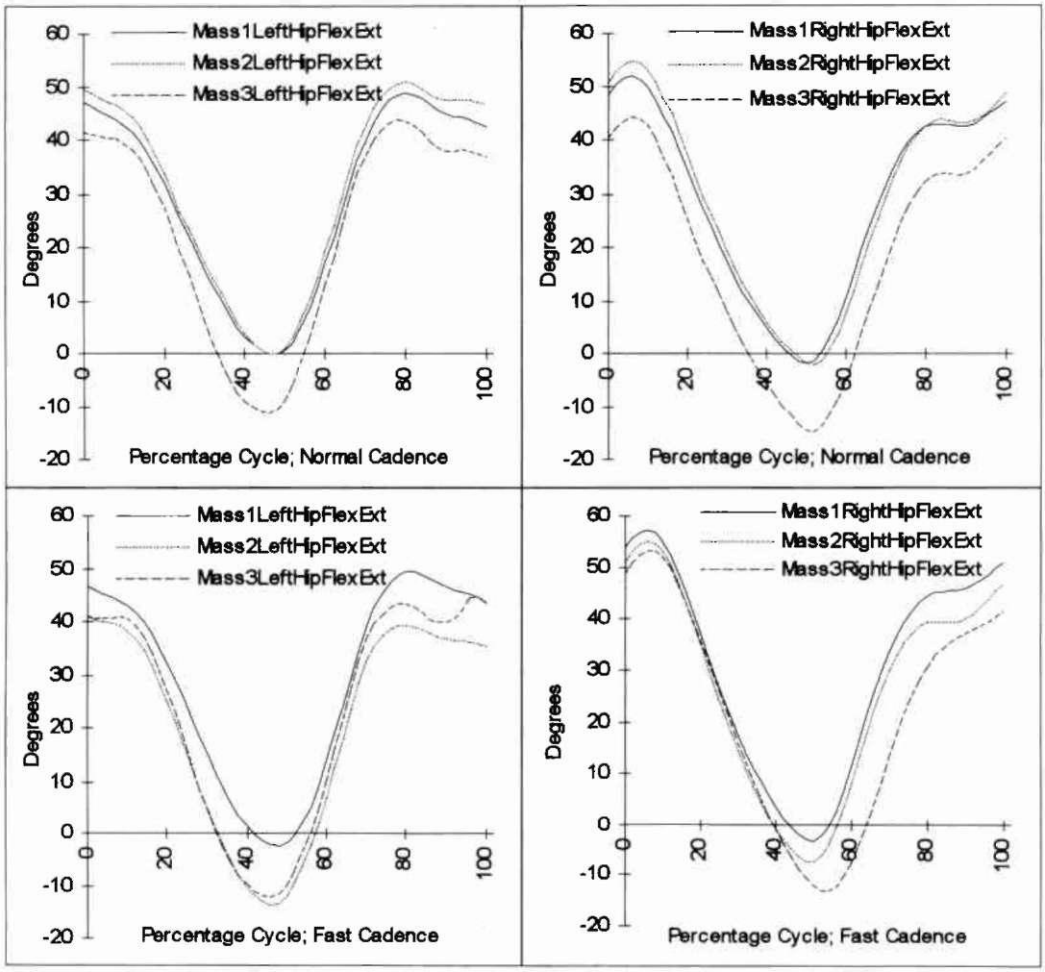

Fig. 1. Affected (left) and intact (right) hip joint angle, normal (top) and fast (bottom) cadences.

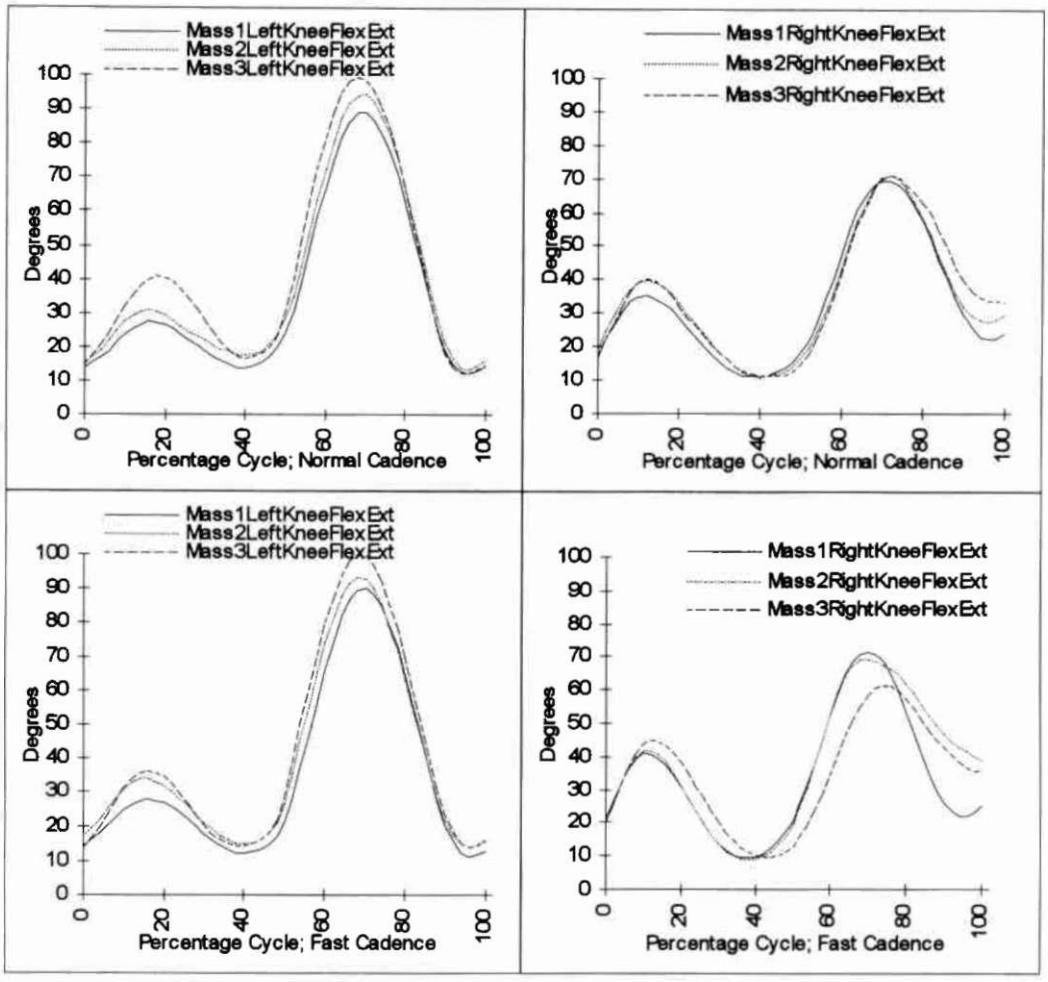

Fig. 2. Affected (left) and intact (right) knee joint angle, normal (top) and fast (bottom) cadeces.

affected limb. At initial contact the hip is flexed to 48° (0.8) for m_1, 50° (0.9) for m_2 and 40° (0.7) for m_3 and unlike the affected limb, the hip continues to flex in early stance before extending at 10% of the cycle. This is probably as a result of the action of lifting, contralaterally, the prosthesis into swing. Prior to toe off for all three mass conditions the intact hip hyperextends somewhat (2° for m_1 (0.7) and m_2 (1.3)) although it is most extreme in the case of the greatest mass at 15° (1.0). In swing the hip on both the intact and affected sides is more extended for m_3, except for the affected limb at fast cadence where it is more extended for m_2.

Walking at the fast cadence, the affected limb flexion/extension pattern for the hip joint angle is similar to the normal walking results for each of the conditions, though the magnitudes vary. The hip begins to extend immediately after initial contact for m_1, though for m_2. the hip extends more slowly from the initial contact of 40° (1.2) of flexion and begins to extend rapidly at 10% of the cycle. For m_3 the hip remains at 40° (1.4) of flexion until 14% of the cycle. For m_1 the hip continues to 2° (0.8) of hyperextension, and then begins to flex as the body prepares to lift the prosthesis into swing. Maximum hip flexion in mid to late swing is 49° (0.7). For m_2 and m_3 the hip flexion extension pattern is virtually identical until mid to late swing, when the hip becomes more flexed for m_3 than for m_2, except as indicated above for the affected limb at fast cadence.

For the intact limb at fast cadence the pattern

is again similar to the pattern at normal cadence, though the mass variation has a different effect. In this instance, the three masses produce similar results in early stance. At initial contact the hip is flexed at 50-51° in all cases. Again, similar to the normal cadence pattern, the hip flexes slightly before extending. Slight discrepancies occur in the timing and degree of maximum extension before the hip begins to flex in preparation for swing. During swing the hip of the intact limb is most flexed for m_1 and least flexed for m_3 when the prosthesis is in stance.

Knee joint angle

The knee joint angle for the prosthetic limb follows a similar pattern to the normal patterns published in the literature. At normal cadence the knee is in 14° (0.7) of flexion for m_1 and 15° (1.0) of flexion for m_2 and m_3. Maximum flexion is early stance for m_1 is 27°, m_2 31° and m_3 40° (1.2). During terminal stance and pre-swing the knee is flexing and this reaches a maximum just after toe off (m_1: 89° (1.4); m_2: 94° (1.8); m_3 100° (0.8). After reaching this maximum, the knee begins to extend in preparation for the next stance period.

At normal cadence the intact knee is not substantially affected by the mass change and follows a normal pattern. The knee is in slight flexion at initial contact (m_1 and m_3: 17° (1.3); m_2: 19° (1.4). The knee flexes through early stance to a maximum of 35° for m_1 and 40° for m_2 and m_3. The knee extends for mid-stance and again flexes in terminal stance and pre-swing in preparation for toe off. After toe off maximum

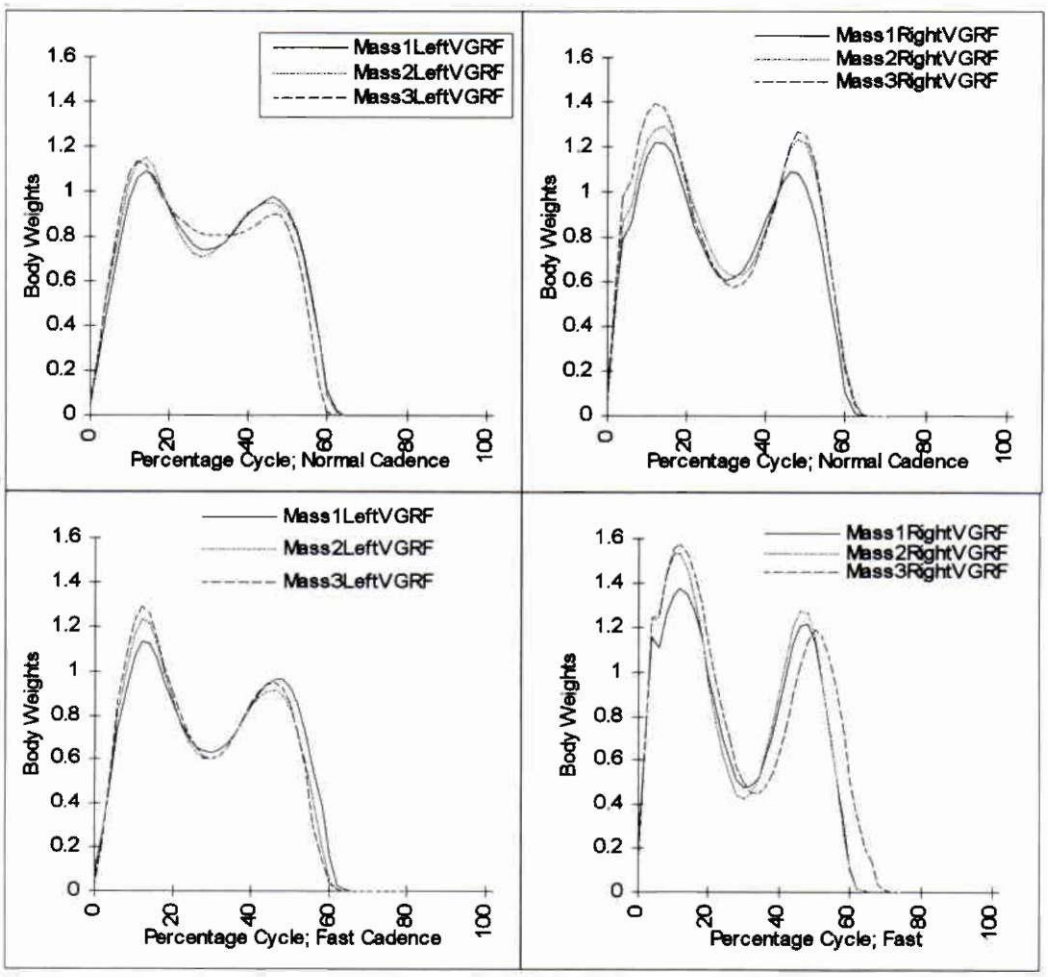

Fig. 3. Affected (left) and intact (right) vertical ground reaction force, normal (top) and fast (bottom) cadences.

flexion is reached (m_1: 69° (0.6); m_2: 71° (0.9); m_3: 71° (0.8)) and again the knee extends to prepare for the oncoming stance.

The fast cadence affects the knee flexion extension pattern for the intact limb. The knee flexes more in early stance when using the greatest mass (45°) and the extension to mid-stance is slower, reaching maximum extension at 48% of the cycle. The flexion is preparation for swing is slower, and the knee does not flex as much for m_3 as for the other two conditions. Throughout the swing period the intact limb is less flexed when the heavier prosthesis is in stance.

Ground reaction force patterns

The vertical ground reaction force (VGRF) and anterior-posterior ground reaction force (A-PGRF) are shown in Figures 3 and 4 for normal and fast cadence. The two vertical peaks typical in walking exist though the second active peak is much lower than normal, but is nonetheless typical for trans-tibial amputees.

The VGRF data is similar for all the mass conditions for the affected limb. The first active peak as the body accepts weight on the supporting affected limb is slightly higher for m_2 at 1.15BW (0.05) than for m_3 at 1.13BW (0.01) and m_1 at 1.09BW (0.01). The trough between the 2 peaks is similar for m_1 and m_2 (0.74BW, 0.71BW) and the curve rises to the second peak values which are below normal (0.97BW, 0.95BW, 0.90BW).

Under these conditions a noticeable impact peak exists for all conditions on the intact side at initial contact. The intact side first active

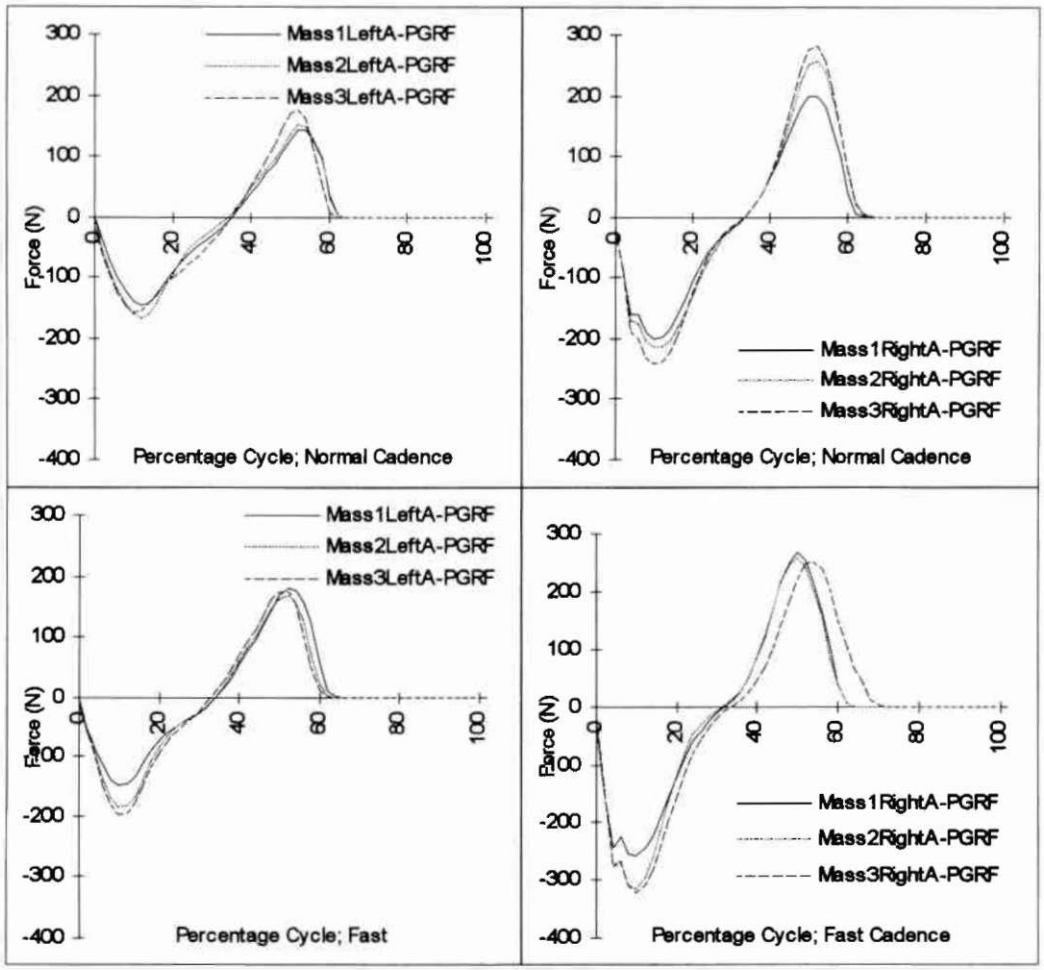

Fig. 4. Affected (left) and intact (right) anterior-posterior ground reaction force, normal (top) and fast (bottom) cadences.

peak is related in magnitude to the mass condition and is highest for m_3 (1.39BW) and lowest for m_1 (1.15BW). The trough is similar for all 3 mass conditions, as is the second peak.

Walking at fast cadence, on the affected limb, there is little alteration to the magnitudes of the first peak compared to the normal cadence, except for m_3 which has increased to 1.29BW. The trough is lower at the faster cadence as the body moves over the limb. The second peak is the same for all three mass conditions and similar to the force at normal cadence.

For the intact limb, the impact force is pronounced and again the first peak is related to cadence, with a larger magnitude than for normal cadence. The trough is lower than for the affected limb and the second peak is similar for all conditions and for the normal cadence (1.21BW, 1.27BW, 1.19BW)

There is little variation in the A-P GRF for the mass conditions ipsilaterally, however compared to the intact limb the magnitudes differ. In all cases, the braking and propulsive forces on the affected side are smaller than for the intact side.

Discussion

Varying the mass resulted in altered temporal and kinematic characteristics and ground reaction force magnitudes in the trans-tibial amputee's walking gait. This is to be expected as the altered inertial characteristics will affect the prosthesis in swing and also the accelerations at the commencement and conclusion of the affected swing phase. Across prosthetic conditions the self selected normal walking cadence was consistent, though m_2 seemed to result in a slightly faster normal cadence. The lightest prosthesis produced the most rapid cadence. The minimal disturbance in the normal cadence when the mass is altered indicates that the amputee is able to maintain his normal walking cadence over a limited distance and number of trials presumably by adapting other aspects. The reduction in the fast cadence with increased mass is similar to the findings of Jans and Bach (1995) and Hale (1990). The velocity over the plate at normal cadence increased as the mass of the prosthesis was increased and this faster velocity was achieved through a longer stride. Slight variations are apparent in the single support time. At normal cadence the single support time is nearly symmetrical for m_1 and m_2. At the fast cadence the observation that the prosthesis is in single support for a shorter time compared to the period for the intact limb corresponds with findings for normal subjects (Hale, 1990). For the greatest mass, the single support asymmetry is obvious for both cadences and this may be as a result of the amputee feeling insecure.

Hip flexion/extension in most cases follows a similar pattern with the different masses. The heavy mass caused the hip to extend more throughout the gait cycle relative to the other mass conditions. Interestingly, the right (intact) limb is also affected by the mass condition, and this is almost definitely a compensation for the action of the limb, although this compensation is not quantifiable without a joint moment and power analysis. The similar patterns for m_1 and m_2 indicate that the mass difference does not affect the kinematics of the hip of the amputee walking at normal cadence for these masses. However, the hip flexion extension pattern for both limbs at the fast cadence for m_2 was similar to that for m_3. At the fast cadence compensatory mechanisms employed by the amputee to successfully walk are evident at the hip in stance and swing. Throughout swing the hip is more flexed than expected, probably to ensure toe clearance as the prosthesis does not allow any dorsiflexion at mid-swing.

The knee patterns are not affected to the same extent by the mass changes as the hip patterns, although again the differences in the mass conditions are most obvious for the faster cadence. The knee is flexed at initial contact and continues to flex through early stance. This flexion is a result of the amputee preparing for foot-flat. During mid-stance to terminal stance and through to pre-swing knee patterns for the three masses were similar. Walking at his normal cadence, the amputee bent his knee more during swing for m_3 on the affected side.

The normal cadence intact knee pattern is not greatly affected by the mass conditions, indicating that the amputee is able to maintain normal kinematics for these conditions.

At fast cadence the intact leg is affected by the mass alterations, most notably in the timing of the knee pattern for the greater mass. The knee extends and flexes more slowly than for the other two masses both while the prosthesis is preparing for swing and while it is in swing. This is as a result of the weight imbalance

caused by the large prosthesis mass of the contralateral limb swinging rapidly, and the knee is involved in ensuring stability at the joint. The swing phase kinematics also indicate that for the heavier prostheses, the knee of the intact limb does not flex to the same extent as for the lighter prosthesis.

The vertical ground reaction force patterns vary for the different prosthesis masses also, although the difference between the intact and the affected limbs is more substantial. Anzel *et al.* (1991) suggest that reducing the load on the intact limb could reduce the effect of any bilateral pathology and possibly also delay the onset of the pathology. However in this study, for all masses, the intact foot struck the forceplate with a higher force than the affected side. The m_1 condition produced the most symmetrical forces at normal walking, although at fast walking this was not the case and the greater masses showed a higher loading force which is more symmetrical. The prosthetic limb showed an increase in the first peak from normal to fast cadence which is expected as the foot will strike the plate with a higher acceleration. The trough of mid-stance is higher for the affected side compared to the intact limb and indicates a higher acceleration of the centre of mass over the prosthesis in stance. The second vertical peak does not increase form normal to fast cadence for the prosthesis under any condition. This indicates the inability of the plantarflexors to contract and therefore the inability of the prosthesis to push off into swing, rather it is merely a support on which the body balances. The impact peak which exists for the intact limb at both cadences is missing for the prosthetic side and indicates that there is shock absorption in the mechanical construction of the prosthesis at initial contact.

The similarity of the A-P GRF for all the mass conditions in normal walking indicates that prosthetic mass does not have an effect on the braking and propulsive forces in walking. Walking fast however results in a larger braking force for both m_2 and m_3.

Conclusion

The results indicate that the inertial characteristics of the DER prosthesis affect the temporal and kinematic gait characteristics of trans-tibial amputee gait.

The joint kinematics are affected by the mass condition with the fast cadence demonstrating the most obvious disturbances when using the heaviest prosthesis. It appears that the joint kinematic disturbances can be kept to a minimum for normal cadence walking using the different mass conditions for the limited period of testing, although at the fast cadence the inertial manipulations result in kinematic and GRF adaptations to the increased loading. Most notably, the hip of both the intact and affected limbs is more extended in swing at the normal cadence and the knee seems to flex as a probable consequence, although using the m_3 prosthesis a number of trials had to be discarded as the prosthesis hit the floor in mid-swing. The intact limb is also more extended at the hip and knee in swing when using the m_3 and this may imply a lack of confidence by the amputee when the heavy prosthesis is in stance. The hip flexion in stance is much greater than for the normal population cited in the literature.

The inertial characteristics should therefore be taken into consideration when designing a new prosthesis and especially if the prosthesis is to be used over a number of cadences. The effect on the gait of the amputee is limited due to the restricted laboratory conditions for the normal cadence variables though the faster cadence indicates that a light prosthesis is more suited to the gait of a trans-tibial amputee.

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The Brian Blatchford Prize

The Brian Blatchford Prize has been established by the Blatchford family to honour the memory of Brian Blatchford. It is awarded every three years at the World Congress of the International Society for Prosthetics and Orthotics.

The Prize of £2,500 will be awarded to an individual who has an outstanding record of innovative achievement in the field of prosthetics and/or orthotics. The achievement should be related to prosthetic and/or orthotic hardware, or scientifically based new techniques which result in better prostheses or orthoses. The President, in seeking to identify the recipient of the award, will also consider nominations or applications from National Member Societies or individuals. Such nominations or applications should contain a justification together with a curriculum vitae of the candidate and should reach the President of ISPO by 1 January 1998 at the following address:

Seishi Sawamura,
The Hyogo Rehabilitation Centre,
1070, Akebono-cho, Nishi-ku,
Kobe 651-21,
JAPAN

The prizewinner shall make a presentation based on his/her work at the Closing Ceremony of the 9th World Congress, Amsterdam on Friday July 3 1998 and the paper shall be duly published in *Prosthetics and Orthotics International*. The President and Executive Board of the International Society for Prosthetics and Orthotics and the Blatchford family reserve the right to withhold the Prize should no suitable candidate be identified.

Prosthetics and Orthotics International, 1997, 21, 124–128

ICEROSS - a consensus view: a questionnaire survey of the use of ICEROSS in the United Kingdom

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Abstract

The management of the individual with a trans-tibial amputation has been strongly influenced by the increasing use of the ICEROSS socket system over recent years. Despite this growth in clinical experience, there has been very little research into its place in current prosthetic practice, and prescribing activity is largely determined by personal experience. In order to formulate the current consensus view on the use of ICEROSS, questionnaires were sent to 42 doctors and 43 senior prosthetists around the UK. The influence of 38 different factors on prescribing activity was assessed using a grading system (ranging from "primary indication" to "absolute contraindication"). An 85% response rate was achieved and no significant differences in response between the two professional groups were identified. Those factors considered by most to be positive indications for using ICEROSS were "pistoning", "shear-sensitive skin / split-skin grafts", "patient unsuccessful with supracondylar (s/c) or cuff suspension" and "insufficient suspension due to change in type or level of activity". Those considered by most to be absolute contra-indications were "ulceration / unhealed scars", "poor patient hygiene" and "poor patient commitment to prosthetic rehabilitation". This consensus of opinion is in keeping with the results of the few published audits of ICEROSS usage. There was a lack of consensus, however, about the use of ICEROSS in some situations, including skin complications.

Whilst some consensus does exist about the use of ICEROSS, the results of this survey indicate significant variations in clinical practice which serve to illustrate the urgent need for data from prospective clinical trials.

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Introduction

The Icelandic Roll-on-Silicone Socket (ICEROSS) was first developed in 1985 by Ossur Kristinsson as a system which relies on the unique properties of silicone and claims to considerably improve the weight-bearing capability of the prosthesis and the interface between prosthesis and user (Kristinsson, 1993). The system has been used increasingly in the United Kingdom over the past five years, but despite this, there has been very little research into its place in current prosthetic practice. The only published works have been retrospective audits of clinical practice (Cluitmans *et al.*, 1994; Panagamuwa *et al.*, 1995) which both identified a significant incidence of troublesome side effects, as well as certain advantages over a conventional socket system. Panagamuwa *et al.* (1996) concluded that careful patient selection was necessary to improve the effectiveness and minimise the complications of the ICEROSS system.

Prescribing activity is usually determined by a combination of several factors. These include

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Analysis of torso movement of trans-femoral amputees during level walking

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Abstract

The purpose of this study is to analyze the movement of unilateral trans-femoral amputees' gait and find patterns of compensated movement to accommodate the loss of locomotor power on one side. A 3-D analyzer system and force plate were used to measure 12 amputees.

The main focus was to find characterized movement particularly of the upper body such as pelvis, shoulder and arms. It was Saunders *et al.* (1953), who said that the human body can be purposely divided into two subsystems in analyzing man's locomotion.

This study, however is not to measure the level of function, but to evaluate the appearance of gait.

Twelve markers were placed on the subjects, and two locations for measuring points were used to measure movement of the torso in three axis (X, Y, Z) without measuring the position of the centre of gravity. The two points were at shoulder level, and at pelvic level.

Lissajou's figure was used to evaluate the movements at these two marks. The quality of the gait was determined through subjective evaluation by the author. Determination was made on two factors, one from observing the gait of the amputees and the other, patterns from their Lissajou's figure of two measuring points.

Then they were categorized as good walker or "other". Good walkers had results at shoulder level which were closer to the symmetrical Lissajou's figure of normal walkers. Yet the symmetrical pattern was not present at the pelvis level for the good walker.

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Introduction

Over the past decade there have been many developments in prosthetic components which aimed for ever higher levels of function. The most important functional loss in the trans-femoral amputee is the knee mechanism.

There are high performance knees, energy storing feet, new materials, new socket designs, with the result that with these advanced aids for amputees, they can now run and participate in many sports activities. Amputees are able to run 100 metre sprints, but the trans-femoral amputee's gait is still noticeable among normal people due to their upper body movements.

This study is to identify the characteristics of trans-femoral amputee's movements, particularly the upper body movements. The belief is that trans-femoral amputees have their own characteristic gait pattern just as patients with knee flexion contracture or fixed ankle joint have their own gait pattern.

Much research has been done on comparison between amputees and non-amputees. In this study, the goal is to find the gait pattern of trans-femoral amputees and compare this pattern between the good walkers and the "other" walkers.

The reason for the two measuring points, instead of using the usual centre of gravity, is twofold; firstly, the difficulty of finding the centre of gravity for amputees with prostheses,

and secondly, the need to find the actual movement between pelvis and shoulder as well as the entire torso. This is due to the fact that the author holds the strong opinion that amputees can never walk in the way that normal walkers can. An amputee must compensate with movement to overcome the different locomotive ability between sound leg and prosthetic leg.

Method

Equipment

A 3-dimensional Analyzer (Oxford Metrics, Vicon system 370) with 6 CCD cameras was used to measure coordinates on the subjects. The subjects wore 12 markers on their body, 6 on the left side, and 6 on the right side.

A force plate (Kyouwa Dengyou EPP-386AS) was incorporated but used only to measure the timing of the gait cycle and to identify the stance/swing phase to exact scale.

The first task of this measurement was to create conditions which would gather accurate data of the subjects' gait within a space of 2 metres (X-axis) x 2 metres (Y-axis) x1.8 metres (Z-axis), with a total length of 8 metres, including an initial 4 metres to allow the amputees to get into a steady gait, with 2 metres of walk through area after the actual measuring space, so that amputees can maintain speed while walking through the measuring space.

It was necessary to have this set up in order to reproduce as closely as possible; the usual gait pattern of the subjects.

Subjects and conditions

The selection of amputees was random, and conditions were as follows:

1. subjects must be using trans-femoral prostheses;
2. prosthetic knee mechanism must not have locked knee;
3. must have endo-skeletal prosthesis;
4. must have foam cover attached;
5. no restriction on socket design;
6. must be using most comfortable heel height for the subject;
7. must be able to walk with confidence at the most comfortable speed;
8. subjects' arms were allowed to have free swing;
9. the prostheses were in the normal alignment configuration as used by the amputees.

There was no previous adjustment on any of the prostheses before the measurements were taken. And there was no restriction on age, sex, weight, height or length of stump.

It should be pointed out that no bilateral amputees were used in this study.

Marker positions

1. left and right acromion process.
2. left and right lateral epicondyle.
3. left and right styloid process.
4. left and right iliac crest.
5. left and right outside of knee joint.
6. left and right lateral maleoli.

A total of twelve coordinates were used.

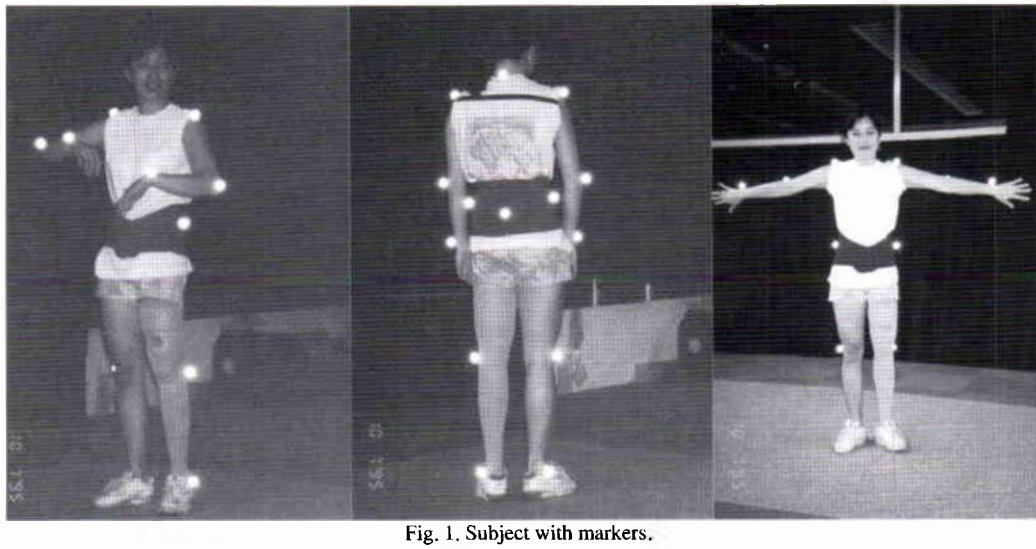

Fig. 1. Subject with markers.

Walking conditions

Twelve spherical markers, 15mm in diameter, were attached to each subject's body at the points described above with double-sided tape (Fig. 1).

Subjects were given free selection of gait speed, and they started walking 4 metres behind the measuring space. The only instruction given to the subjects was to place the right foot on the right force plate and vice versa for the left one.

This simple instruction was given to minimize interference in reproducing their usual gait, since it is rather difficult to require their trans-femoral amputees to walk along a straight line due to lack of lateral support with their prostheses.

No metronome, no step length, step width, no specific side (L or R) for entering the force plate was used so as to make amputees relaxed during the measurement period.

The first measurement was taken in the static upright position for zero calibration, and three measurements were taken at a speed chosen by the amputees.

Data treatment

A Vicon system calibrated all coordinates into a 3-dimensional matrix within the measuring space.

A 3-dimensional diagram established the centre of the measuring space as the origin of measurement, and the direction of X axis is set left to right against line of progression, Y axis is on the line of progression and Z axis is perpendicular to the floor (Fig. 2).

A 3-dimensional diagram provides sequential data, and the sampling frequency for this study was done at 60Hz, with the measuring time 4 seconds. This means that each marker will produce 240 sequential data. The data were processed by a digital low pass filter as Bryant (1984) described.

According to Cappozzo (1981) the appropriate cut-off frequency is 6Hz, and 6Hz was the figure used in this study.

Calibration

Based on smoothed sequential data, calibration was done on angular and linear relationship between the acromia and iliac crests.

Calibration was based on the observation made from behind the subjects in the frontal plane, so the starting points were set at the left acromion process and the left iliac crest.

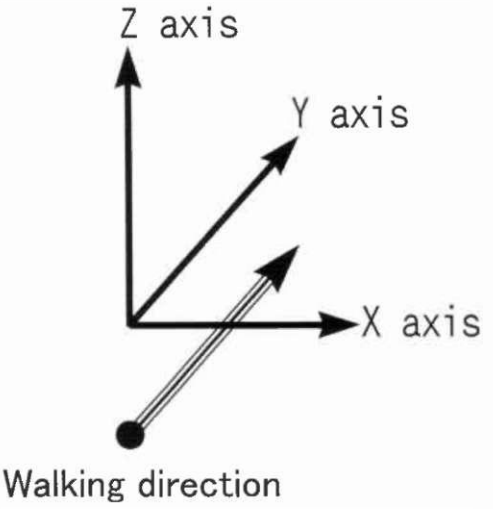

Fig. 2. Three axis of dimension.

If the left acromion process matrix is set as (X_1, Y_1, Z_1) at a given time, and at the same instance the right acromion matrix is set as (X_2, Y_2, Z_2), then the angle Θ of the line between these two can be calculated as follows:

$\Theta = \arctan (Z_2-Z_1) / (X_2-X_1)$

Referencing of the markers location on the body was established using data from the standing still position (Fig. 3).

Angulation at pelvic level was done by the same calculation.

Lissajou's figure

A Lissajou's figure was made on the mid-point between both acromion processes and iliac crests.

Often, the direction of movement of the subjects and the Y axis of the measuring space are not exactly the same, and this causes shifting of the locus on each gait cycle.

The presumption was made that subjects move from point A to point B in a straight line, and the centre is selected from the time of heel contact during gait cycle.

The coordinate of first heel contact is (X_1, Y_1, Z_1), the coordinate of the second heel contact is (X_2, Y_2, Z_2), and from this, the amount of shifting by the subject in the measuring space will be (X_2-X_1).

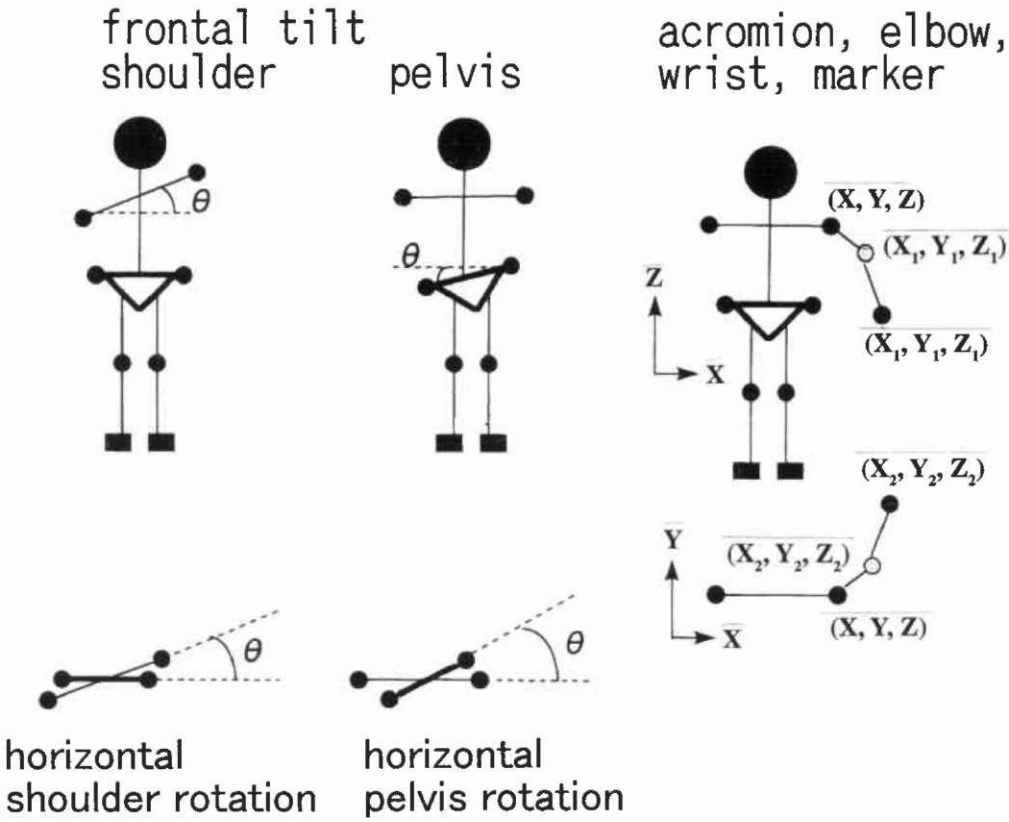

Fig. 3. Calculation of angle.

Results and discussion

The concept of "Locomotor and Passenger" was applied to this study of evaluating the gait of trans-femoral amputees. Instead of comparing the trans-femoral amputees' gait with normal gait, in this study the focus was on identifying the movement characteristics of trans-femoral amputees to compensate for their loss of locomotor system. It also, compares the good walker and "other" walker within trans-femoral amputees.

Twelve subjects were arranged in order according to the appearance of their gait. Also, evaluation of the movement of cross-over points at shoulder and pelvis level, using Lissajou's figure, was done. These two subjective evaluations had the same result.

Twelve subjectively evaluated Lissajou's figures are shown in Figures 4.1, 4.2, 4.3. From these two separate subjective evaluations, 5 (numbers 1-5) were categorized as good walkers and 7 (numbers 6-12) were classified as "others". They were numbered in order from the best to worst walkers and, these numbers were used in all graphs and material.

The walking speed of the 12 subjects was measured, but the order of their speed did not necessarily correspond to whom were the best or the worst walkers, as can be seen in Figure 5.

Lissajou's figure in frontal plane

The mid-point between the acromia and the iliac crests not only represents movement at one point, but also the movement at two points, therefore, movement at both distal parts can be measured.

This mid-point method indicates that the accuracy of measured data is very close to the method of one point measuring usually used in Lissajou's figure. But the method used in this study is also able to pick up movement in the lateral part of the body.

In this study, all the data is compared with the sound side and the amputated side, in contrast to the usual method in the Lissajou's figure of using the left and right sides. To compare the

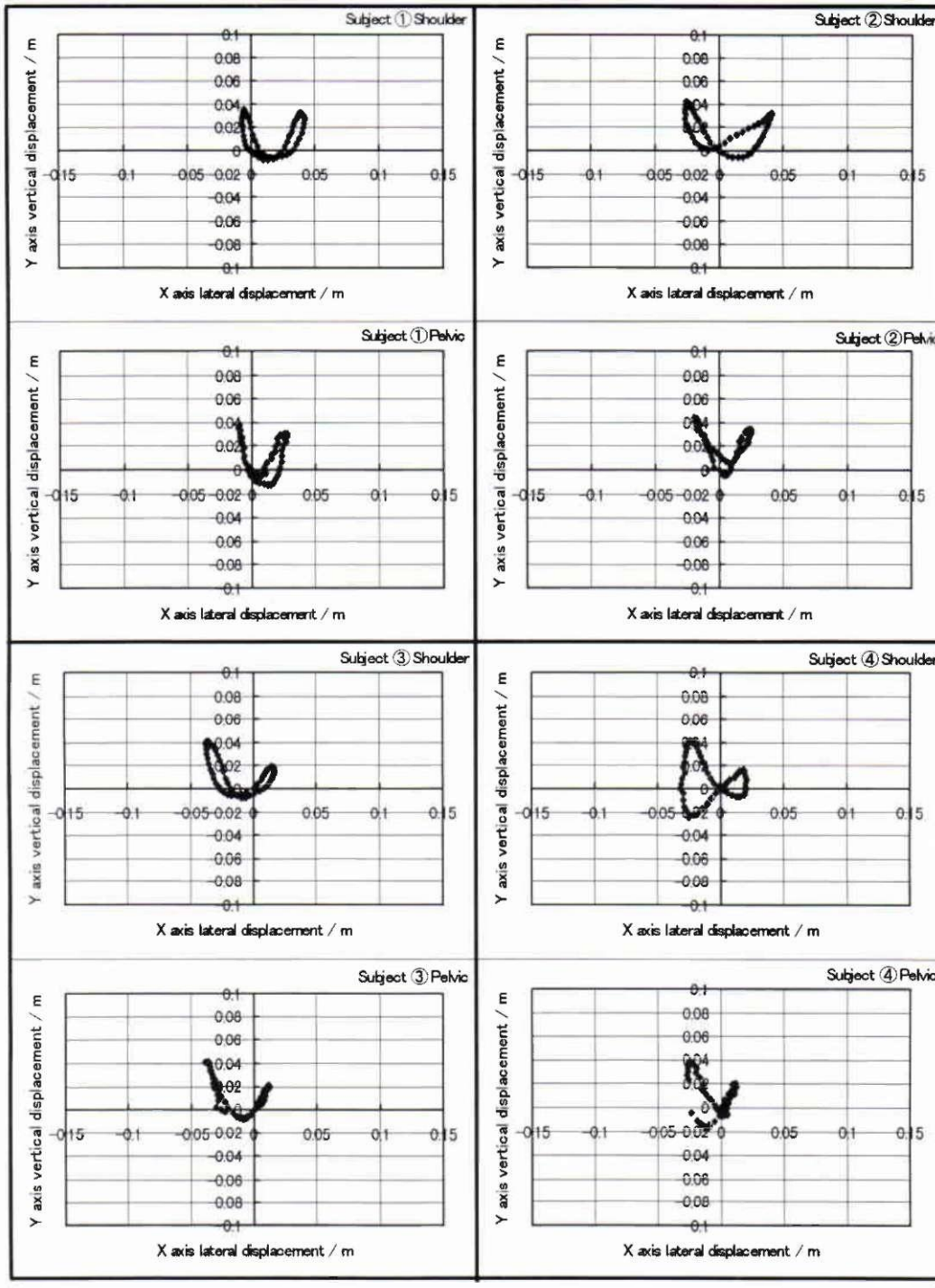

Fig. 4.1. Lissajou's figure.

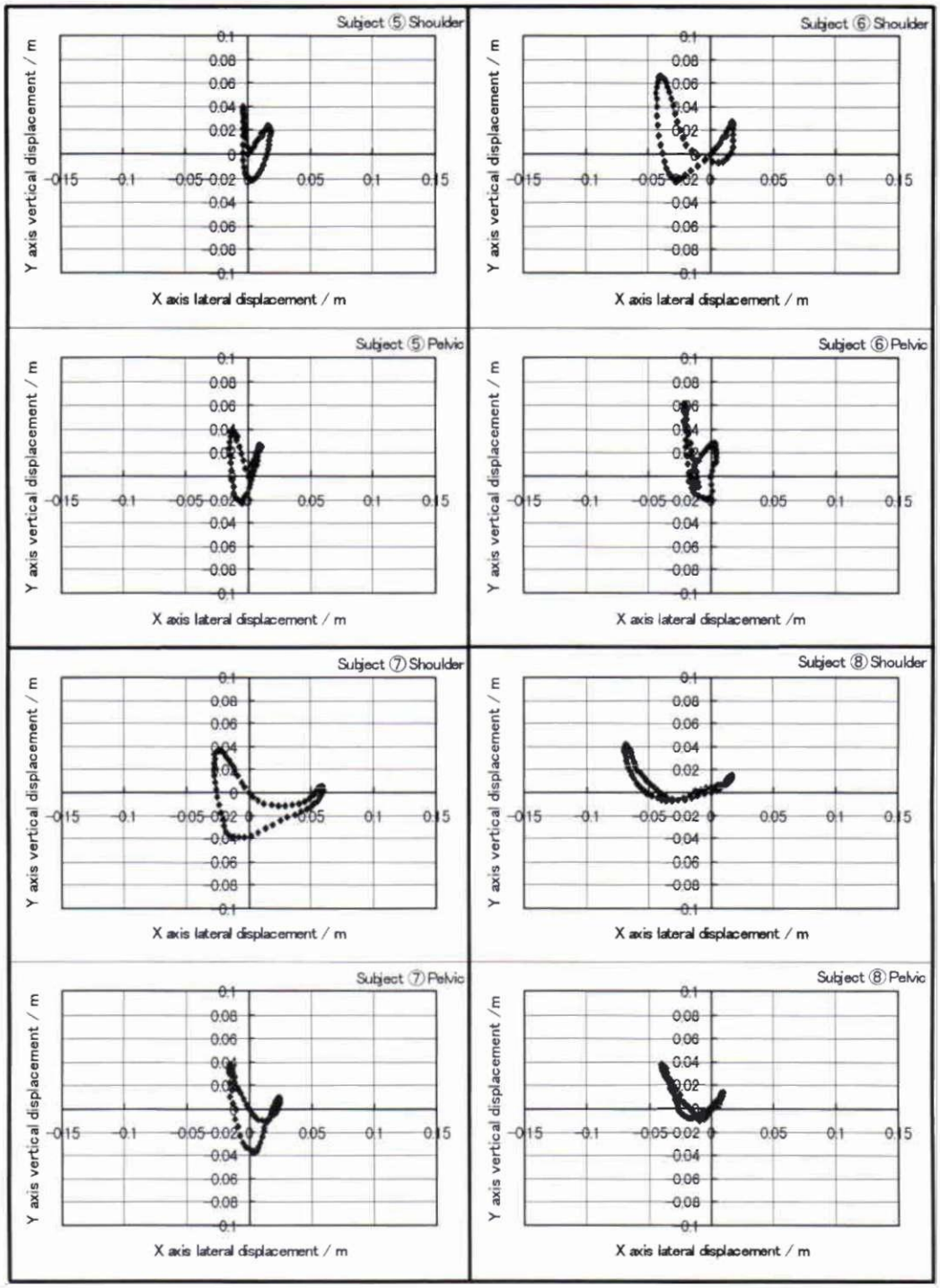

Fig. 4.2. Lissajou's figure.

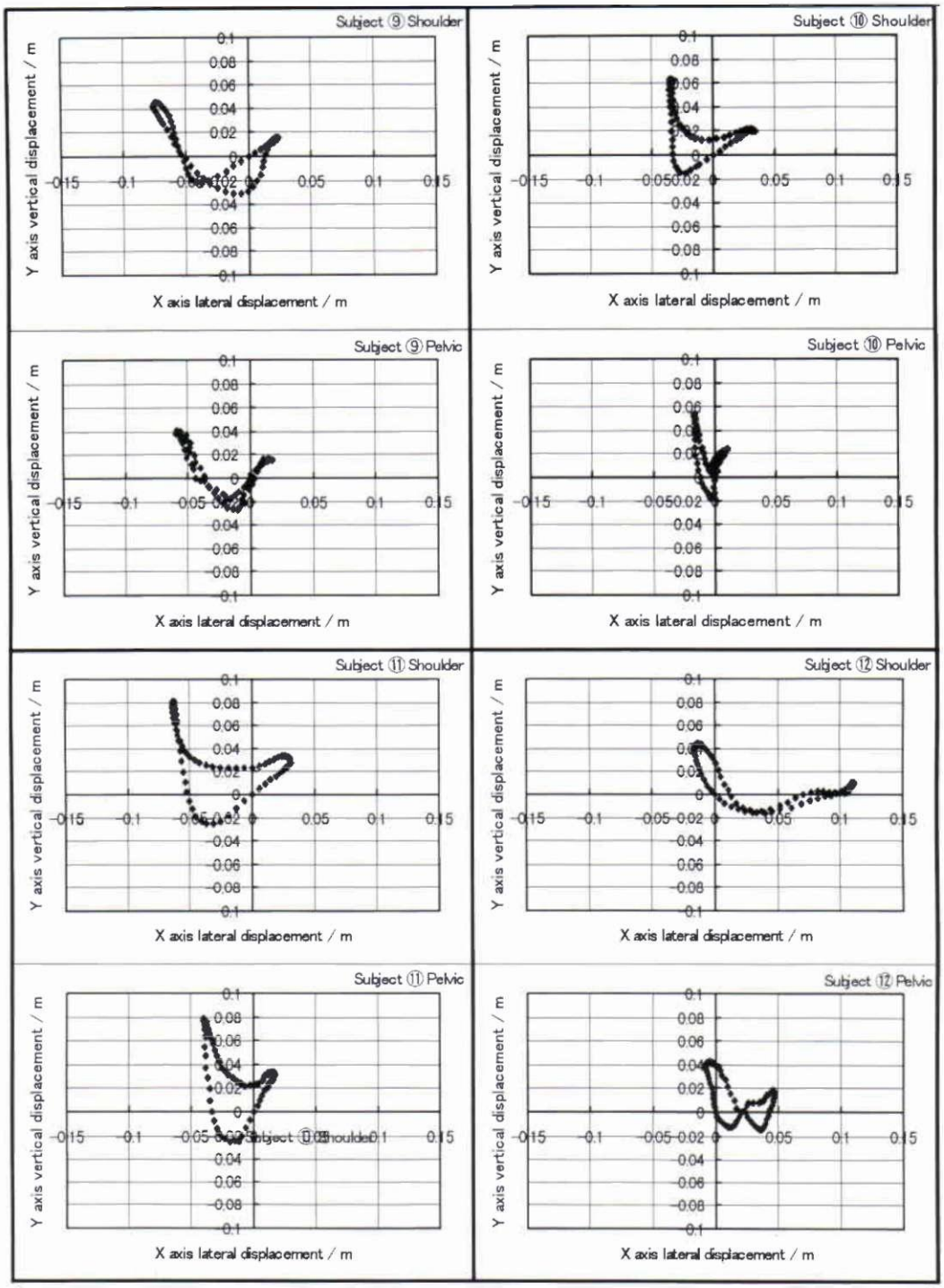

Fig. 4.3. Lissajou's figure.

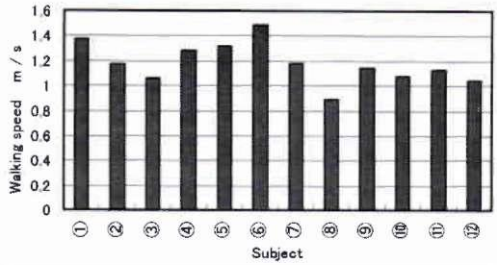

Fig. 5. Walking speed of subject.

data more easily, it has been shown with the sound side on the right and the amputated side on the left.

It is surprising that the results from the two subjective evaluations were exactly the same. All subject's names were deleted.

In the Lissajou's figures, 7 subjects showed bigger displacements.

Shoulder level

The most noticeable point of body movement is at the shoulder level, and the amount of maximum movement on Lissajou's figure was measured in the vertical plane and lateral horizontal plane.

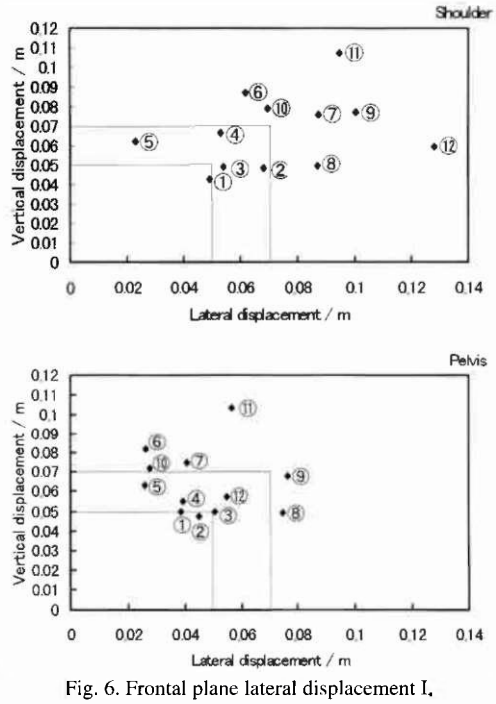

Fig. 6. Frontal plane lateral displacement I.

Figure 6 indicates maximum movements at shoulder level in two planes of all 12 subjects. The X-axis shows lateral maximum displacements and the Y-axis indicates maximum vertical displacement of the shoulder measuring point, which is mid-point between both acromion process markers.

There was only one subject within the 5cm area, but there were 5 subjects within a 7cm area, and 5 of those were the very same 5 subjects chosen as good walkers in the subjective evaluation.

Pelvic level

Using the same method as with the shoulder, there were 6 subjects in the 7cm area, and 5 of those were from the group of good walkers, and one was in the 5cm and 7cm area respectively, but from the "other" group (Fig. 6).

In this graph, X-axis indicates lateral displacements and Y-axis indicates vertical displacements.

Total amount of displacements at shoulder and pelvic level

The next comparison was the total amount of both lateral and horizontal displacement at pelvic and shoulder level (Fig.7).

A 14cm area was chosen and 6 subjects were within that area, but all the good walkers were within the 12cm area. The next closest one was nearer the 14cm line.

Differences in the amount of movement between shoulder and pelvic levels

Another comparison was done with Lissajou's figure which indicated the differences in movement between shoulder and pelvis (Fig. 7). This was a case of the pelvis being the locomotor and the shoulder the passenger.

Hypothetically, a perfectly normal walker will have his/her zero points at both X and Y axis.

The largest difference in vertical plane between shoulder and pelvis was 1.1cm and the smallest one was 0.03cm and both were from the group of good walkers.

On the other hand, lateral displacement resulted in a more dramatic figure. The largest lateral displacement between shoulder and pelvis was 7.7cm and this was in the "other" group.

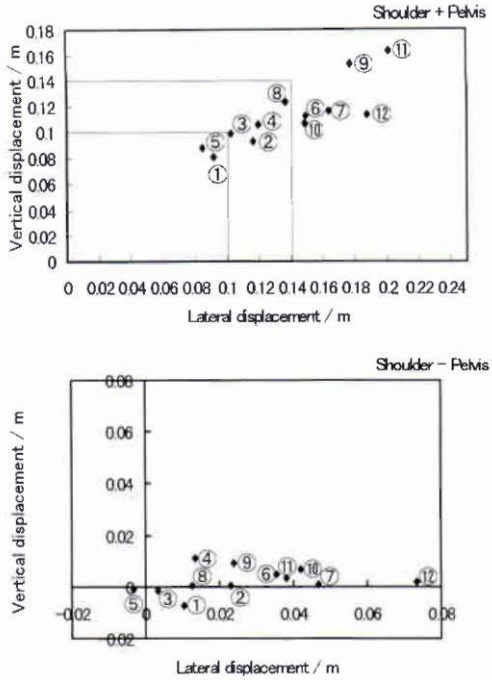

Fig. 7. Frontal plane lateral displacement II.

The smallest displacement was 0.3cm and was from the good walker group. All the good walkers were within 2.2cm lateral displacement.

Results from Lissajou's figure make it clear that the subjects chosen by the subjective evaluation left smaller displacement figures in all aspects. The one exception, subject No. 12, which appears in the good walker group in test 2, was worst in test 4 with the largest lateral displacement of 7.3cm.

Shoulder and pelvic tilt in frontal plane

These angles were calibrated from lines made from two points at shoulder and pelvic level.

Calibration was based on the observation made from behind the subjects in the frontal plane, so the starting points were set at the left acromion process and the left iliac crest.

Tilting angle at shoulder level

Figure 8 shows the maximum tilting angle at shoulder level in the frontal plane. The prosthetic side is on the right and the sound side is indicated on the left.

Even though angulations were different for subjects Nos 3, 5 and 8, the study shows that they always walk with their sound side shoulder higher than the other. Subject No. 10 indicated the opposite by having his/her prosthetic side higher all the time. It is obvious that there is a clear pattern between the good walkers and "others". Also, there was similarity in the symmetrical movement and degree of angle in the different subjects.

Tilting angle at pelvic level

Figure 9 shows the result of pelvic tilt in the frontal plane, with the right side prosthetic and the left side sound. This graph shows a pattern between the two groups; the good walkers' angle of tilt is smaller, whereas all those from the "other" group are bigger.

These two pieces of data provide evidence that good walkers are able to stabilize their pelvis during the stance phase on either leg. In other words, they are able to keep their pelvis in the right position during the swing phase of both legs.

From these results, it is quite safe to assume that good walkers can maintain the pelvis level. It is evident that good walkers either have sufficiently strong lever arm strength in their amputated hip joints or they engage in compensatory movements in the pelvis, or in a higher region of the trunk as well, such as the

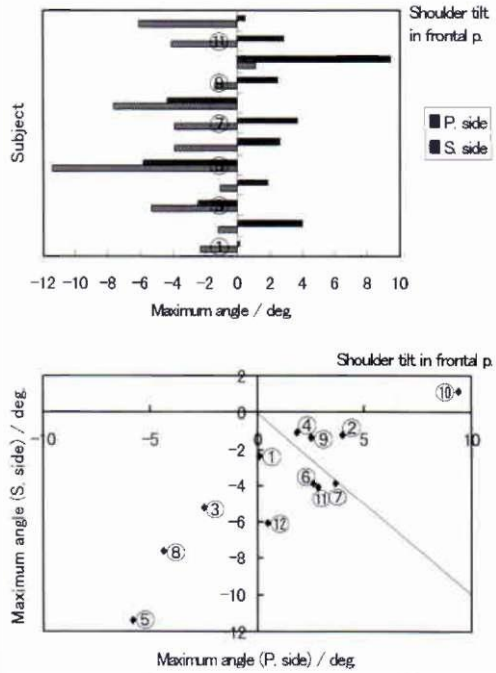

Fig. 8. Shoulder tilt in frontal plane.

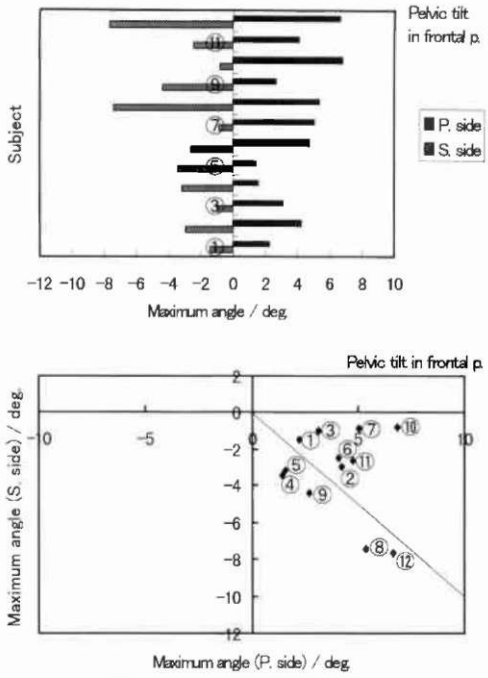

Fig. 9. Pelvis tilt in frontal plane.

shoulder. This cannot, however, be supported by the data here.

Rotation in horizontal plane

Calibration was based on observation made from behind the subjects in the frontal plane, so zero points are set at the left acromion process, and the left iliac crest.

The figures calibrated to compare the rotation in the horizontal plane need to have very careful attention.

Inman *et al.* (1981) stated that 8 degrees of rotation occurs during one gait cycle, 4 degrees of rotation each to anterior and to posterior from the mid-point. Also, on one leg there must be 8 degrees of rotation between pelvis and the foot, which is planted on the ground.

This means that the trans-femoral amputee must have rotation between the socket and pelvis, unless the prosthesis is equipped with a torque reducing unit.

Relative factors between tilt angle and horizontal rotation regarding shoulder and pelvis both have serious meaning. For the trans-femoral amputee particularly having rotation between the socket and stump is a very serious matter. For those reasons, this chapter is a key section of this study.

Horizontal rotation at shoulder level

Figure 10 shows the shoulder rotation of the 12 subjects. The good walkers had lesser degrees of movement and their movement was much more symmetrical.

In general, symmetry was evident for the "others" as well, except for subjects Nos. 5 and 10.

Horizontal rotation at pelvic level

It is quite clear that the good walker group has a pattern that can be observed in Figure 11.

The good walker group had a common pelvic movement, in that their prosthetic side pelvis never moved behind the pelvis of the sound side during an entire gait.

The "other" walkers' sound side pelvis tended to move somewhat forward of the prosthetic one.

From these points, it can be assumed that there is less movement between the prosthetic side pelvis and the prosthesis, especially in the good walker.

Also, good walkers turn more on their sound side pelvic rotation, but very little on their prosthetic side pelvic rotation.

Comparing total movement of the trunk, it is quite obvious that good walkers move their

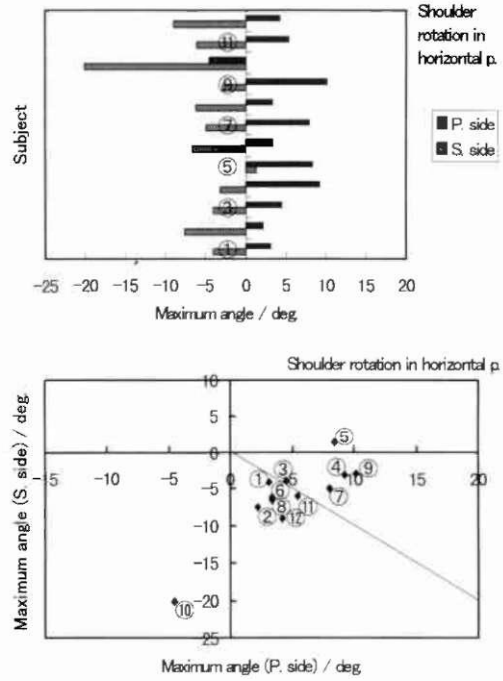

Fig. 10. Shoulder rotation in horizontal plane.

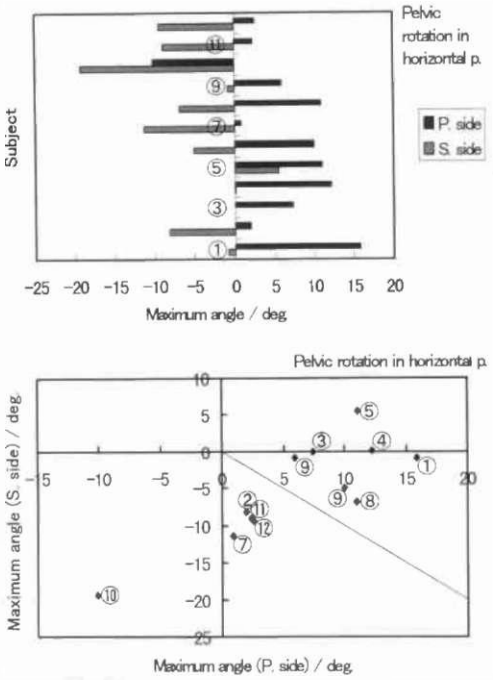

Fig. 11. Pelvis rotation in horizontal plane.

shoulders in a very symmetrical way, yet they move their pelvis rather lopsidedly, and this is one of the compensatory movements to overcome their disability with locomotion.

Step length and arm swing length

Legs and arms are a counterbalancing movement through the trunk. This is a very normal attitude in human locomotion.

When lower locomotion has unequal strength or movement, it will naturally have an effect on the arm swing movement.

Step length

Figure 12 the step length chart, shows the length of the prosthetic step on the Y-axis and the sound side step length on X-axis.

There is no pattern nor difference between good walkers and the "others". In spite of uneven locomotor strength, it is clear that trans-femoral amputees can stride the same with either the prosthetic or sound leg.

Arm swing length

Figure 13 is a graph showing arm swing length measured from the acromion process to the styloid process on the sagittal plane. This

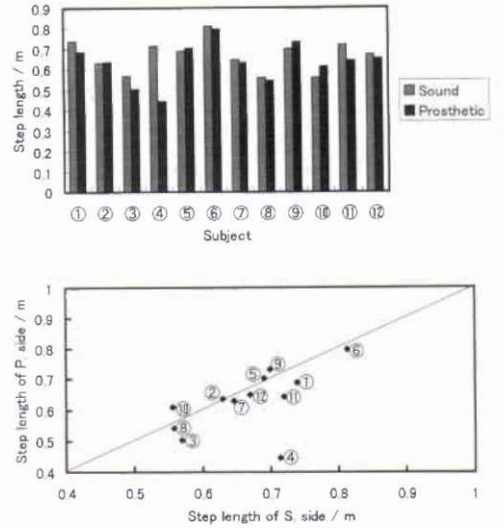

Fig. 12. Step length.

measurement did not include abduction, or adduction and did not consider elbow flexion.

Every subject in this study showed that the swing of the prosthetic side arm was larger and it moved further forward to the body.

There was no pattern and no difference between the good walkers and the "others".

The sound side arm swing was smaller for all the subjects.

Conclusion

Present prostheses have no power of their own. Their locomotive function is basically a

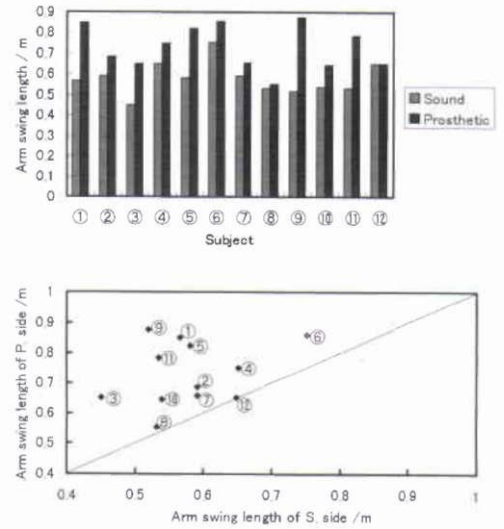

Fig. 13. Arm swing.

pendulum movement. Amputees move their stump to swing the prosthesis, which is mainly controlled by a knee mechanism.

With different locomotive abilities in their lower limbs amputees have to make compensatory movements to overcome their disabilities.

Through this study, a few points have been clarified in this regard.

Amputees who walk well have lesser movement in their gait, and most noticeable is their shoulder lateral displacement.

But, good walkers have rather lopsided pelvic movements, and it can be assumed that they are making compensatory movements.

Of particular interest, is that good walkers keep their pelvic horizontal rotation in relation to the prosthesis much smaller compared to the sound side pelvic horizontal rotation.

This has the result that, between pelvis and shoulder, there is uneven rotation between sound side and amputated side.

It was rather difficult to evaluate the data on the movement of arms due to the complex movement of shoulders (adduction, abduction), and elbow (flexion, extension). But it was clear that arm swing is a balancing factor to accommodate weakness of push-off on the prosthetic side.

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The efficacy of the one-leg cycling test for determining the anaerobic threshold (AT) of lower limb amputees

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Abstract

The aim of this study was to investigate whether or not the one-leg cycling test driven by the subject's sound leg as the exercise load method is an applicable method for determining the anaerobic threshold (AT) of lower limb amputees. To evaluate physical fitness, a graded exercise test that monitored gas exchange, ventilation and heart rate (HR) was performed in 51 unilateral lower limb amputees. AT was successfully measured for 42 out of 51 subjects, an 82.3% success rate. The average AT was 12.7 ± 2.2 ml/kg/min, and the average HR at AT point was 117.7 ± 16.2 beats/min. The average peak oxygen uptake was 20.1 ± 5.6 ml/kg/min, and the average peak HR was 145.1 ± 22.4 beats/min. The peak HR exceeded the HR at AT by an average 27.4 beats/min, which indicates that a comparatively intense exercise load above the AT level is possible. The average AT was 40.9% of the predicted maximum oxygen uptake, which seems reasonable when compared to the reports of other researchers. These results suggested that the one-leg cycling test driven by the sound limb is of use as a method for determining the AT of lower limb amputees.

Introduction

Lower limb amputees walking with prostheses, particularly those with trans-femoral prostheses, must expend considerably more energy than able-bodied people (Gonzalez *et al.*, 1974; Waters *et al.*, 1976) and as a result the physical burden on them is considerable. It should be considered as whole-body exercise. Consuming more energy makes the amputee tire more quickly than his/her able-bodied peers. In turn the amputee is inclined to reduce his/her walking. Reducing his/her walking decreases the level of fitness and makes walking even more of an effort. In this connection maintenance and increase of the level of fitness is essential to the amputee, providing a preventive treatment of hypokinetic state. In this respect exercise training is considered to be of importance for the amputee. However it should be noted that in previous prescription of exercise maximum oxygen uptake was firmly entrenched as the standard indicator of the level of fitness. For disabled peers measurement of maximum oxygen uptake is practically impossible and a new indicator is required in the prescription of appropriate exercise for them. The anaerobic threshold (AT) is a concept introduced by Wasserman (Wasserman *et al.*, 1973; Wasserman, 1984). AT is an indicator for exercise performance which is particularly effective in expressing long-term performance (Yoshida *et al.*, 1982; Hurley *et al.*, 1984). AT can be used as an indicator of level of fitness for amputees. Like maximum oxygen uptake, AT can vary with fitness level. As fitness decreases, so too does the AT. Below the AT, aerobic pathways which are quite efficient, are used to supply energy. Above the AT, aerobic pathways diminish and anaerobic pathways, which are not efficient are used with greater frequency. AT is

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a value reached by exercise load test and is measurable at relatively low load levels. However, it is of course difficult to evaluate the AT of amputees who have not yet been fitted with prostheses and the treadmill test and other incremental exercise tests which have been used in the past can be difficult for some amputees even if they are able to walk with a prosthesis. Therefore the authors used the one-leg cycling test driven by the subject's sound leg as the exercise load method to examine whether or not it is an applicable method for determining the AT of lower limb amputees. The conclusions are reported here.

Subjects

The subjects were 53 unilateral lower limb amputees, comprising 5 hip disarticulation amputees (2 male, 3 female) aged between 13 and 52 with an average age of 40.0 years, 37 trans-femoral amputees (28 male, 9 female) aged between 19 and 78 with an average age of 51.8 years and 11 trans-tibial amputees (10 male, 1 female) aged between 17 and 67 with an average age of 35.6 years. Of these, 5 were unable to walk with a prosthesis and a further 6 needed double crutches when doing so. These 11 were all trans-femoral amputees. The other 42 were able to walk without support or with one cane. The physical characteristics of the subjects are shown in Table 1.

Method

In this research a cycle ergometer was used (Lode Angio WLP-300ST, Holland) which can be used from a supine position. Informed written consent was obtained before entry into the study. The tests were conducted with the subjects seated with their upper bodies reclining at an angle of approximately 45° (Fig. 1). An incremental exercise test was begun with three minutes of unloaded pedaling with the test subjects directed to turn the pedals 60 times per minute. The exercise intensity was increased by 10 watts per minute with the increase completed at the end of each section. The exercise was at the subjects' self-assessed maximum load. The subject is driving the ergometer with his sound leg. During exercise the respiratory gas was monitored with a respiromonitor (Minato RM-300 system, Osaka, Japan) and the AT point measured. At the same time the electrocardiograph (ECG) and heart rate (HR) were monitored during exercise by Stress Test system (ML-5000, Fukuda Denshi, Tokyo, Japan) and cuff blood pressure was determined every minute with autoelectro-cardiometer (Colin STPB-780, Japan). AT was determined using the following criteria (Wasserman *et al.*, 1973) : a systematic increase in the ventilatory equivalent for O_2 ($\dot{V}_E/\dot{V}_{O_2}$ without an increase in the ventilatory equivalent for CO_2 ($\dot{V}_E/\dot{V}_{CO_2}$).

To confirm the reproducibility of AT values the one-leg cycling test was run twice on 10 of 42 subjects at an interval of several days, and the correlation between the two results was calculated. The predicted maximum oxygen uptake was calculated using the formula of Hassen *et al.* (1984) and the predicted maximum HR was calculated as (220 -age).

The data from the "self-assessed" maximum load was reported. The Pearson Product-Moment correlation technique was used in all correlation analysis. Differences were considered significant at $P < 0.05$. All values reported are means ± SE.

Results

AT and peak oxygen uptake values during exercise

Two test subjects showed severe arrhythmia on the ECG before the test and did not

Table 1. The physical characteristics of the subjects

Measures	Trans-femoral (n = 37)	Trans-tibial (n = 11)	Hip-disarticulation (n = 5)
Age (year)	51.8 ± 18.9	35.6 ± 17.0	40.0 ± 16.5
Height (cm)	162.7 ± 8.5	169.4 ± 5.9	158.5 ± 11.1
Weight (kg)	54.5 ± 9.7	60.9 ± 8.5	57.0 ± 9.9

Values are means ± SE.

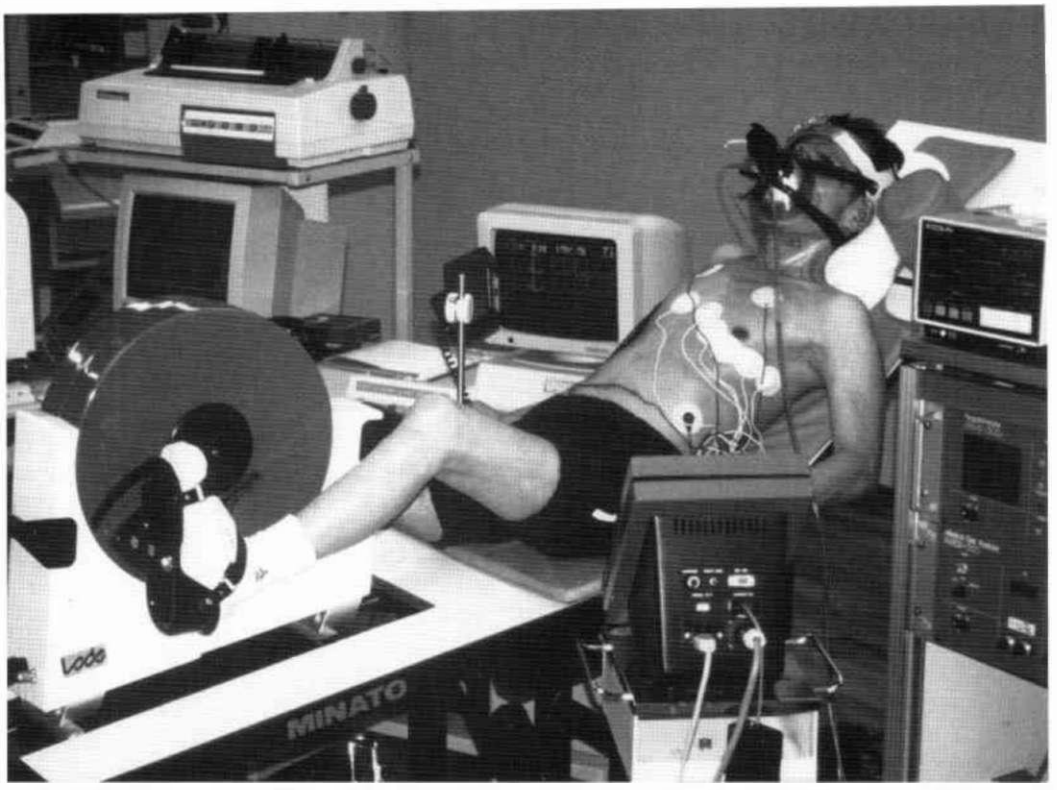

Fig. 1. One-leg cycling test. The subject is driving the ergometer with his sound leg. Under comprehensive heart monitoring, oxygen uptake and other factors are measured and their values are continuously displayed on the screen of a personal computer.

undertake the test. AT was successfully measured for 42 out of 51 subjects, an 82.3% success rate.

The remaining 9 subjects who could not be tested were all extremely weak in the muscles of their lower limbs and their tests were unavoidably suspended before completion. Among the other subjects there was no cause to suspend tests for extreme exhaustion, chest symptoms, abnormal cardiogram output or any other reason.

Among the 42 who yielded an AT value the average peak oxygen uptake was 1159.4 ± 330.2 ml/min, which corresponds to a rate per unit body weight of 20.1 ± 5.6 ml/kg/min. The average peak heart rate was 145.1 ± 22.4 beats/min. The average AT was 743.9 ± 143.7 ml/min which corresponds to a rate per unit body weight of 12.7 ± 2.2 ml/kg/min. The average heart rate on reaching AT was 117.7 ± 16.2 beats/min. The average values of predicted maximum oxygen uptake and predicted maximum heart rate were 1818.2 ± 477.7 ml/min and 177.6 ± 18.3 beats/min respectively.

The relationships between AT and the predicted maximum oxygen uptake and the peak oxygen uptake under exercise load

Among the 42 subjects who yielded an AT value the correlation coefficient for the relationship between AT and predicted maximum oxygen uptake was 0.66, indicating a significant correlation between the two ($p<0.001$) (Fig. 2). The correlation coeficient for the relationship between AT and peak oxygen uptake was 0.82, indicating a significant correlation between the two ($p<0.001$) (Fig. 3)

The reproducibility of AT found using the one-leg cycling test

The correlation coefficient between the first and second AT values found using the one-leg cycling test was 0.962, indicating a significant correlation between the two ($p<0.001$) (Fig. 4).

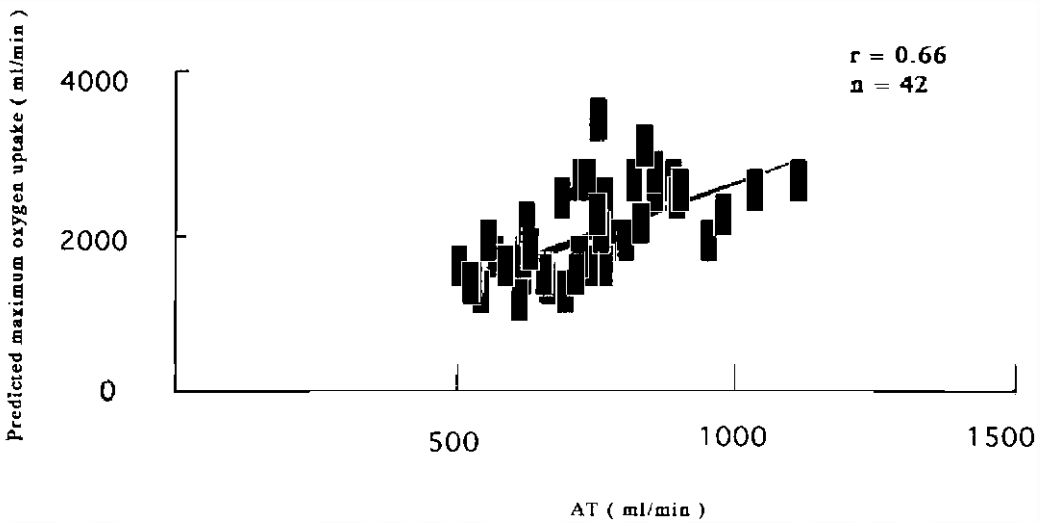

Fig. 2. The relationship between AT and the predicted maximum oxygen uptake. Among the 42 subjects for whom AT was determined a significant correlation was observed between the two factors (r = 0.66, P< 0.001).

Discussion

The purpose of the study was to demonstrate that one-leg cycling test could be used to determine AT. AT as suggested by Wasserman *et al.* yields information as to each individual's measure of fitness (Wasserman *et al.*, 1973; Wasserman, 1984) and it also has applications in exercise training as an indicator in related fields such as myocardial infarction, diabetes and obesity. It is now being adopted in clinical situations. Regrettably it has not been widely used in the field of rehabilitation of lower limb amputees suffering from numerous complications. This is because the equipment for analyzing respiratory gases is uncommon and the measurement methods are cumbersome, and also because exercise load testing of lower limb amputees by normal methods is difficult.

The authors have now taken the one-leg cycling test, using the sound limbs, as the

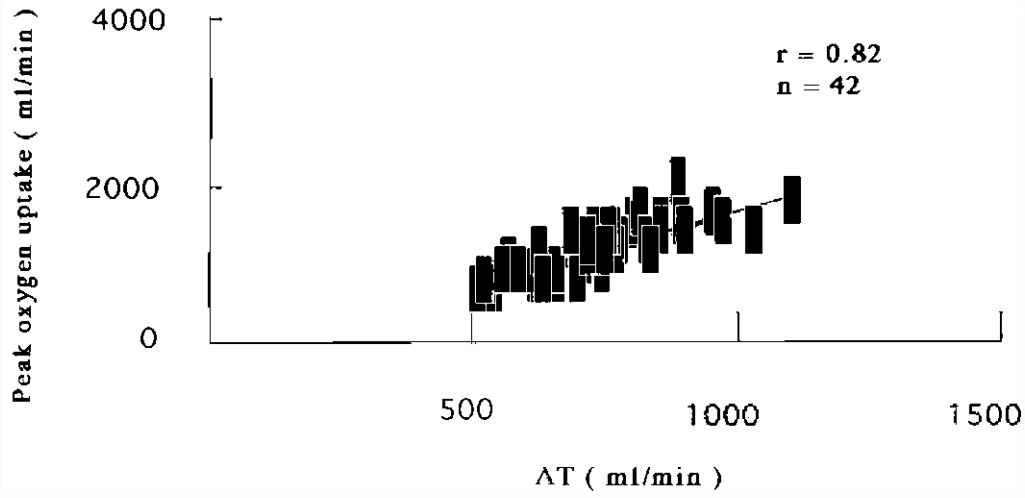

Fig. 3. The relationship between AT and peak oxygen uptake under exercise load. Among the 42 subjects for whom AT was determined a significant correlation was observed between the two factors (r = 0.82, P< 0.001).

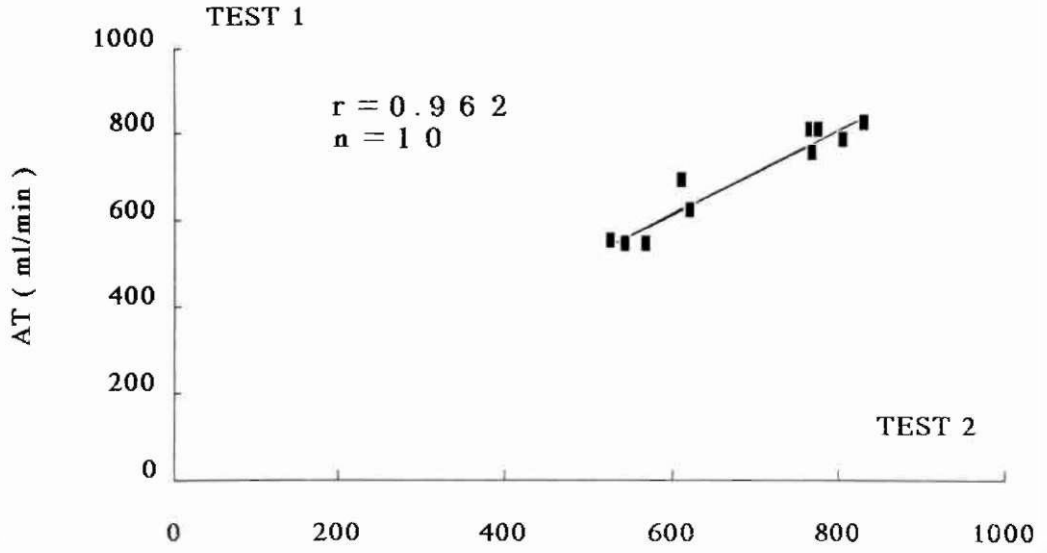

Fig. 4. Reproducibility of AT. The one-leg cycling test was repeated on ten of the amputees. A significant correlation was observed between the first and second AT levels (r = 0.962, P< 0.001).

exercise load for AT determination. On examination of its applicability it was possible to obtain the AT of 42 out of 51 patients, a high heart rate of successful detection at 82.3%. The peak heart rate was 81.6% of the predicted maximum heart rate on average. The peak heart rate also exceeded the heart rate at the AT by an average of 27.4 beats/min, which indicates that a comparatively intense exercise load above the AT level is possible. Furthermore the average AT was 40.9% of the predicted maximum oxygen uptake, which seems reasonable when compared to the reports of various other researchers (Skinner and McLellan, 1980; Davis *et al.*, 1979). Using this method the reproducibility was also good. Considering the above observations it appears that the one-leg cycling test driven by the sound limb is of use as a method for determining the AT of lower limb amputees.

Maximum oxygen uptake is closely linked to the performance of the oxygen transport system and has received wide acceptance as the indicator of level of fitness. Accordingly maximum oxygen uptake was firmly entrenched as the indicator in prescription of exercise. However, maximum load is necessary for measurement of maximum oxygen uptake which makes implementation of this test practically impossible for amputees. Prescription of exercise and evaluation of level of fitness are defined with methods and criteria suitable for able-bodied peers. The conditions in subjects with lower limb amputation are not taken into account. On the other hand this research found a significant correlation between the AT values gained from the one-leg cycling test and the predicted maximum oxygen uptake, which proved that AT is suitable as an indicator to reflect the level of fitness. In addition AT can be detected at a comparatively low exercise load which means amputees can undergo testing in safety. Thus AT can be the appropriate indicator of level of fitness for amputees. The implementation of exercise training based on appropriate indicator will improve the level of fitness of amputees, facilitating their return to life in the community as able-bodied peers.

Conclusion

The applicability of the one-leg cycling test driven by the subject's sound limb as a means of determining the AT of lower limb amputees has been proven. This study has indicated the feasibility of the clinical application of AT in exercise training for amputees.

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A review of 42 patients of 16 years and over using the ORLAU Parawalker

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Abstract

A 7 year retrospective review of 42 patients of 16 years or over using the ORLAU Parawalker has been conducted to establish the degree of long-term compliance in using the orthosis on a regular basis. Regular use was defined as putting the orthosis on at least once a week.

All subjects had been supplied with an ORLAU Parawalker via the routine supply procedures adopted in Oswestry, and were followed up at regular 6 month intervals as part of the standard treatment regime.

The records from routine follow-up were surveyed for those patients who were continuing to use their orthosis to establish age, length of time since supply of orthosis and cause of lesion. Average period of usage is calculated for those still using their orthosis, and for all patients in the study.

Of the 42 subjects, 32 were myelomeningocele patients with confirmed absence of innervation of hip extensors and abductors, the remainder being paraplegic patients with traumatic or acquired complete thoracic lesions. Compliance figures were extracted from the results, as were the minimum possible average periods of usage. The respective results were:

- of the 32 myelomeningocele patients 59.4% continued usage after an average period of 85.5 months, and
- of the 10 traumatic or acquired lesion patients 60% continued usage after an average period of 24.8 months respectively, which gave a combined compliance of 59.5% after a minimum average 71.1 months of use.

The performance of myelomeningocele patients suggests that their additional deformities do not lead to inferior compliance as adults and that a high proportion to continue to walk after adolescence.

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Introduction

The continuing interest in reciprocal walking for paraplegic patients is reflected in the growing number or orthoses which are becoming routinely available. Well established designs (Butler and Major, 1987; Douglas *et al.*,1983) have been supplemented more recently by new variants (Lissons *et al.*, 1992; Motloch, 1992; Kirtley, 1992) and further fundamental work on Functional Electrical Stimulation (FES) (Bajd *et al.*, 1989; Hermens and Baardman, 1993; Marsolais and Kobetic, 1987) and hybrid systems (Cliquet *et al.*, 1986; Nene and Jennings, 1989; Isakov *et al.*, 1992) is continuing in the search for even greater levels of ease and efficiency of walking.

The evidence of Mazur *et al.* (1989), who showed that non-walkers had five times the number of pressure sores, twice the number of bone fractures, with walkers being almost four times more likely to be independently mobile within the community as teenagers, strongly suggests that the earlier claims of therapeutic benefit and improved independence for heavily handicapped patients who ambulate regularly

(Rose, 1976; Carroll, 1974; Menelaus, 1987) are justified. Nevertheless doubts persist that the long-term patient compliance necessary to realise those benefits can be sufficiently achieved to warrant the costs of both the orthoses and the required monitoring and treatment of patients.

The market environment in which healthcare increasingly takes place demands an appropriate balance between benefits and costs. This requires that where potential benefits are identified they must be shown to occur in a significant number of patients. Prior to the development of effective reciprocal walking system lower limb orthoses were rarely used over sufficiently long periods by thoracic lesion paraplegic patients for the claimed therapeutic benefits to be realised. Many spinal injuries centres have refrained from prescribing knee-ankle-foot orthosis (KAFO) for this group because their experience suggested that less than ten per cent of such patients continue to use them for more than six months. Paraplegic patients with complete lesions between T1 and T6 were reported by Rosman and Spira (1974) not to use their orthoses at all and those with lesions at T7 to T11 for standing purposes only. A review of a range of spinal cord lesion patients of 15 years and older (Mikelberg and Reid, 1981) showed that for 12 of these with complete thoracic lesions (4 T5-T7 and 8 T12-L1) using bilateral KAFOs and 9 incomplete lesions from C7 to cauda eqina (3 of whom had one KAFO and 1 AFO) only 45% continued to use their orthoses. The follow-up period varied from one to six years but distribution of the time span was not reported. Hahn (1974) reviewed 52 patients with complete lesions at L1 or above using fixed ankle KAFOs. Of those reviewed only 35% reported they could walk 'solo' at a follow-up period of approximately two years.

A more recent study (Moore and Stallard, 1991) of adult traumatic lesion patients has shown the use of an effective reciprocal walking orthosis as part of an on-going treatment regime can produce much improved patient compliance. In a review of 50 patients with complete thoracic lesions using the Parawalker, 64% continued to use their orthoses on a regular basis at an average period since supply of 34.4 months. The higher level of compliance was attributed to the ease of walking in the device and the simplicity of putting on and taking off the orthosis. Subsequent comparative reviews of different reciprocal walking systems have confirmed this device as the most efficient and mechanically reliable of those which are currently available (Whittle and Cochrane, 1989; Banta *et al.,* 1991; Bowker *et al.,* 1992; Jefferson and Whittle, 1990; Lotta *et al.,* 1994; Bernardi *et al.,* 1995), and this suggests that the improved compliance should be consistently achieveable.

The mechanical difficulties of providing walking for children are much less pronounced (Stallard *et al.,* 1989) and as a consequence concern about compliance has largely been confined to adult patients. In most healthcare systems paediatric patients are given more regular routine attention and this further enhances the likelihood that they will persist with their walking programme. However, there is a widely acknowledged difficult transition phase at the time when paediatric patients become adolescent and are able to make their own choices. Since the majority of these will have congenital lesions with the associated increased incidence of deformities which mitigate against walking, the problems for these patients as they move into adulthood may be further exacerbated as compared with traumatic lesion adults who have independently elected to ambulate.

Routine supply of the Parawalker has continued in Oswestry since the earlier study of compliance in adult traumatic thoracic lesion patients (Moore and Stallard, 1991). During that period not have only have additional patients in that category been newly supplied, but many of the paediatric patients treated prior to the original study (who as children did not qualify for it) have passed through adolescence into adulthood. A new study incorporating freshly supplied adults and paediatric patients becoming "adult" has therefore become possible which has enabled the influence of the "adolescence transition" to be examined and more light to be shed on the potential for long-term compliance.

Patients and methods

A 7 year retrospective review of patients supplied with the ORLAU Parawalker was conducted on patients who were 16 years and over. At this age many patients lose access to

paediatric physiotherapy services and effectively become adults in terms of clinical provision.

The myelomeningocele patients in the study included all those being routinely followed up in Oswestry who were or had become adult during the 7 year retrospective review and had lesions which had caused them to have lack of control of hip abductors and extensors.

The adult traumatic or acquired lesion paraplegic patients were all over 16 years of age and had complete thoracic lesions at level L1 or above. This study includes all such subjects who had been supplied subsequent to the Moore and Stallard (1991) review.

Each of the patients had been routinely supplied with an ORLAU Parawalker (Butler and Major, 1989) at Oswestry and were following the pattern of six monthly review which is an essential part of the treatment system.

Data was collected by surveying the patients' records held in ORLAU and maintained by physiotherapists during the regular reviews. Although many factors are recorded this study is restricted to:

- age (measured at the end of the 7 year period);
- length of usage at the time of the survey, if the orthosis was still being used at least once a week. The term 'use' is here defined to mean that the subject put the orthosis on to experience some benefit as perceived by them. Thus, as in the Moore and Stallard (1991) review, this may include functional, recreational or walking benefits. Patients who did not attend for review or respond to correspondence were deemed to have discarded their orthosis and ceased complying;
- cause of lesion (traumatic (or acquired) or myelomeningocele).

Records were only available for those patients who were still using their orthosis. For this reason Average Period of Use is calculated for those still using, with a supplementary average for all patients in the survey, assuming immediate cessation after supply for those who had given up, to indicate the worst possible theoretical outcome.

Results were computed as follows:

$$\bar{P}_u = \frac{\sum t}{n_u}$$

$$\bar{P} = \frac{\sum t}{n}$$

$$\bar{A} = \frac{\sum a_u}{n_u}$$

$$C_u = \frac{n_u}{n} \times 100\%$$

where:

$\bar{P}_u$ = average period of time since supply for those still using
$\bar{P}$ = average period of usage for all subjects
$\bar{A}$ = average age of users in the study
C = Ccompliance
t = period of use for each user (assumed 0 for those no longer using)
a_u = age of each subject still using their Parawalker
n = total number of subjects in study
n_u = number of subjects still using their Parawalker

Results

Of the 42 patients in the study 10 had traumatic or acquired lesions and 32 myelomeningocele lesions. Some 25 of all these continued to use their orthosis on a regular basis.

The results are summarised in Table 1. Compliance for the traumatic or acquired lesion patients was 60% with the average time from supply being 41.3 months for those still using their orthosis (or 24.8 months assuming all non-users gave up immediately), whereas the myelomeningocele patients had a compliance of 59.4% with the average time from supply being 144 months for those still using their Parawalker (or 85.5 months assuming all non-users gave up immediately)

Combining the two groups gives an average compliance of 59.5% with an average time since supply of 119.4 months for those still using (or 71.1 months assuming all non-users gave up immediately).

The average age of all patients in the study was 25.0 years, with the paraplegics having an average of 34.8 years and the myelemeningocele patients an average of 21.9 years.

Discussion

As indicated above, those patients who failed to attend for review or answer correspondence

Table 1. Summary of results

	Traumatic subjects	Myelomeningocele subjects	Traumatic and myelomeningocele subjects
Compliance	60%	59.4%	59.5%
Average time since supply for those still using	41.3 months	144.0 months	119.4 months
*Average time of usage for all subjects	24.8 months	85.5 months	71.1 months
Average age of subjects still using	34.8 years	21.9 years	25 years

*This assumes, because information was not available, that the non-users had ceased using their orthosis immediately and therefore represents the worst case possible.

were deemed to have discarded their orthosis. Experience has shown that a small number of patients ignore calls for routine review until they perceive a problem for themselves, despite continued regular orthotic walking activity. Thus the results presented represent the most pessimistic interpretation possible and may mask a better result.

Nevertheless, results which show a compliance of 59.5% with an average period of usage of almost 6 years (assuming an absolute worst case in which those patients who had given up and for which data was not therefore available had ceased using their orthosis immediately) provide further evidence to bolster the confidence of those who consider walking for heavily handicapped patients worthwhile. Taken in the context of the Mazur *et al.* (1989) findings, in which non-ambulatory children had five times the number of pressure sores, it also suggests that there are direct economic benefits in a long-term walking programme. A full cost analysis of providing a Parawalker in ORLAU indicates that a 3 year programme requires total resources of between £2000 to £2300 per annum. This is supported by a similar analysis in a separate clinical centre (Pratt, 1992). Treatment of a pressure sore has been estimated as costing between £1,000 (El Masry, 1995) and £26,000 (McSweeney, 1994). Taking the cost of a walking programme over the 10 years duration of the Mazur *et al.* (1989) study and comparing it with that of treating the additional pressure sores of non-walkers reported by them over the same period suggests that the use of ambulation orthoses could provide a saving of between £4,000 to £8,000 per annum for each ambulatory patient. This more than covers the cost of treating the 40% of patients supplied with an orthosis who then drop out even taking no account of the additional costs of treating fractures. The misery and inconvenience caused by a pressure sore or bone fracture cannot be ascribed a financial cost, but is clearly extremely worthwhile avoiding in social terms.

Results for the paraplegic patients were strikingly similar to those reported by Moore and Stallard (1991) for a similar group. The performance of the myelomeningocele patients compares very favourably with that of the traumatic paraplegic patients within this review and that reported by Moore and Stallard (1991). This suggests that the majority do understand the benefits of a walking programme, and that despite the additional difficulties which their deformities create a significant number continue to ambulate through and beyond adolescence. The myelomeningocele patients were on average very much younger than the paraplegic

patients in both studies. Whilst no immediate significance can be read into this, it is possible that as they get older their compliance may vary more widely from that shown by the paraplegic patients.

The implied higher level of independence in later life made possible by a walking programme is potentially of great significance. In their study of 36 matched pairs of myelomeningocele patients Mazur *et al.* (1989) reported that walkers were almost four times more likely, as teenagers, to be independent in mobility within the community than non-walkers. This outcome has important financial implications for society as well as the individual patient. The burdens on social services and the patient's immediate family are likely to be considerably relieved by a post adolescent ability to take responsibility for independent general mobility. Local anecdotal evidence on adult myelomeningocele patients treated with the Parawalker from an early age suggests that their ability to achieve this is in line with that reported by Mazur *et al.* (1989) for ambulatory patients. However, additional research will be necessary for this to be positively confirmed.

Effective walking programmes demand a very significant commitment by the patient and their family. Additional problems for walking created by deformities commonly found in myelomeningocele patients are not insignificant, and these add to the difficulties of all concerned. The compliance results for the myelomeningocele patients in this study compare favourably with those shown for adult traumatic or acquired paraplegia patients, and this is particularly encouraging. The study of adult traumatic lesion patients by Moore and Stallard (1991) showed a compliance rate of 64% with average follow-up of 34.4 months. That result and its comparison with this review call into question the commonly held view that myelomeningocele patients will have widely divergent outcomes from those of traumatic lesion patients. Experience in Oswestry shows that the prescription of devices which have a high level of mechanical reliability and provide the efficiency and convenience essential to overall independence, used within the context of an integrated treatment system which includes assessment, controlled supply, appropriate orthotic rectification, training and regular routine follow-up, can enable paediatric patients to continue to receive the therapeutic and functional benefits of ambulation through adolescence into their adult lives.

The results reported in this review provide important support for the concept of walking for heavily handicapped patients. In particular it is clear that the effect of a walking programme on independent mobility within the community as patients become mature adults could have very important long-term social and economic implications. Further evidence on the reasons for abandoning ambulation might make it possible to improve the 59.5% compliance rate reported in this study. The present environment in which the delivery of healthcare takes place demands that such issues be given added prominence. It is hoped that this study will justify financial support for a more in-depth research project to establish a wider range of outcomes relative to clinical and health economics issues.

Acknowledgement

This paper would not be possible without the diligence of the therapists reviewing patients keeping the record sheets up to date, namely Penny Butler, Nicky Thompson, Pat Moore, Michelle Moore, Mary Watkins and Chris Hodnett. The assistance of Soussan Khodadadeh and Jayne Jones in collecting and verifying the data for this paper is gratefully acknowledged.

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Heel lifting as a conservative therapy for osteoarthritis of the hip: based on the rationale of Pauwels' intertrochanteric osteotomy

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Abstract

Patients with osteoarthritis of the hip were treated with a conservative therapy of heel lifting. Orthoses were applied on 35 hips in 33 subjects and the cases were followed for 23 months on average. Dramatic pain relief was reported, but the time required to reduce or completely relieve pain increased according to the stage of osteoarthritis. The radiological results were not satisfactory.

During the follow-up, only two hips showed improvement, 22 showed no change, and 11 deteriorated. The mechanism of heel lifting in relation to the hip joint was analysed, showing that pelvic obliquity was achieved and the trunk stabilized. In conclusion this simple orthosis was effective as a palliative therapy for osteoarthritis of the hip.

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Introduction

A palliative therapy is proposed for osteoarthritis of the hip. For mothers of young children and others who cannot spend time in the hospital for rehabilitation after operation, a treatment was devised to relieve hip pain without medication such as non-steroidal anti-inflammatory drugs.

Operations for osteoarthritis of the hip were developed in the latter half of the 20th century. Among them was the intertrochanteric osteotomy which was developed by Pauwels (Pauwels, 1976), and from which good results have been reported (Ohsawa, 1994). The indications for osteotomy are as follows: valgus

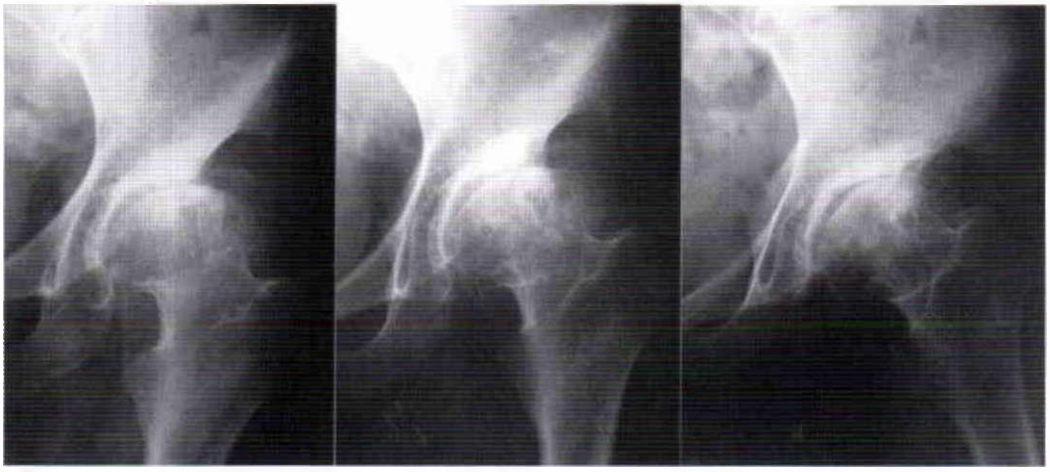

Fig. 1. Indication for a raise on the affected leg (valgus). The congruence was best in adduction of the hip. Left – adduction; middle – neutral; right – abduction.

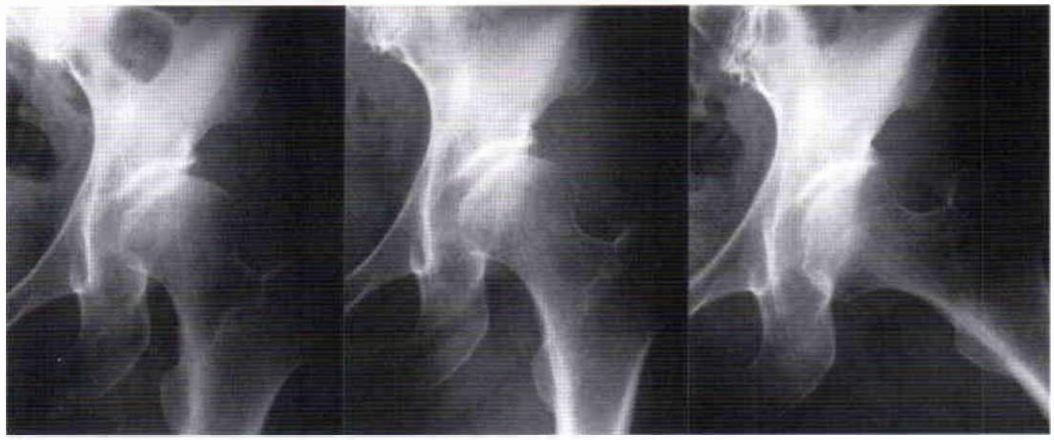

Fig. 2. Indication for a raise on the contralateral leg (varus). The congruence was best in abduction of the hip. Left – adduction; middle – neutral; right – abduction.

osteotomy is indicated when congruence improves with adduction of the hip (Fig.1). Varus osteotomy is performed when the congruence improves with abduction of the hip (Fig. 2). The authors proposed instead to use lifting to tilt the pelvis. The raised side would produce hip valgus on that side and raising of the contralateral side would produce hip varus (Fig. 3). As a result, this lifting could be as effective as intertrochanteric osteotomy. Clinical tests were performed to test this hypothesis (Ohsawa and Ueno, 1993).

Materials and methods

Indication and application

The orthosis was offered to patients who refused operation but suffered from hip pain from osteoarthritis (Table 1). The procedure was applicable to all stages of osteoarthritis. When the joint congruence improved with adduction of the hip, a valgus effect on the hip was necessary (Fig. 1). A raise was therefore applied to the affected leg. The amount of lift was the same as that of the discrepancy of the functional limb length, so that patients felt that both limbs were the same length. When the joint congruence improved with abduction of the hip, a varus effect on the hip was necessary (Fig. 2). A raise was therefore applied to the contralateral leg. The maximum amount of lift was around 1.5cm, because a higher lift caused the shoe to slip off (Fig. 4).

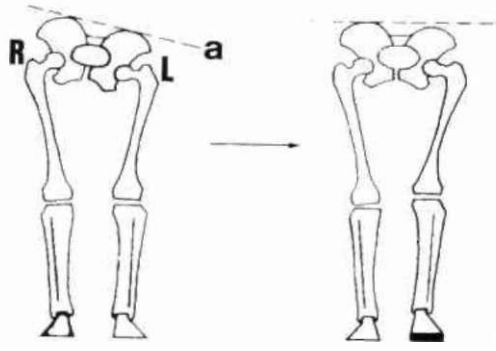

Fig. 3. Mechanism of heel lift. The pelvic line (a) was changed by the heel raise. The hip on the lifted side suffered a valgus effect (left hip, L) and the contralateral hip suffered a varus effect (right hip, R). The black square indicates a heel raise.

Table 1. Demographic data of patients

	valgus	varus
hips	24	11
average age (years)	54	51
average follow-up (months)	24	23
average hip score (points)		
before	11	12
after	12	14
stage (hips)		
early	2	4
advanced	9	7
terminal	13	0
average lift (cm)	1.4	1.5
average range of motion (degrees)		
flexing before	88	95
final	84	101
abduction before	12	29
final	12	24

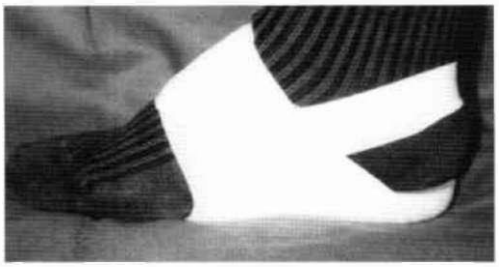

Fig. 4. Heel raise. Patient can use the raise with or without shoe to suit the Japanese life style.

Methods of analysis

Clinically, the Merle d'Aubigné hip score was used (Merle d'Aubigné and Postel, 1954), along with pelvic radiography. Patients were analysed in the standing position using a radiograph of the pelvis, and body centre analysis with and without heel lifting. Motion analysis was carried out on patients in walking.

Standing position: The angle between the pelvis and the femur was measured for all patients using a pelvic radiograph in the standing position with and without a raise (Fig. 5). The movement of the body centre of the 25 patients was studied using a body centre analyzer (San'ei, 1G06, Japan). The patients stood for 30 seconds in relaxed open-eye condition with and without the raise. The body sway was defined as the absolute value of A minus B shown in Figure 6.

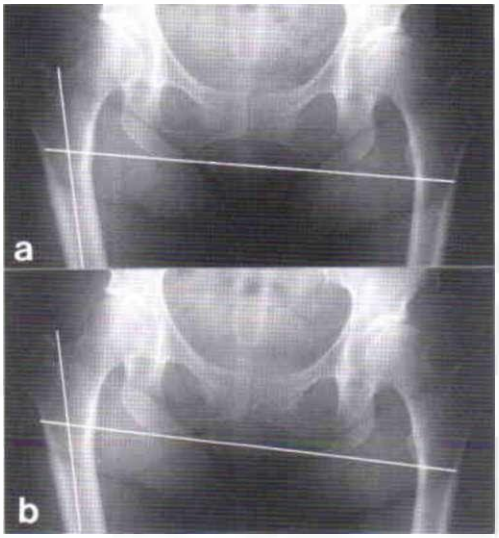

Fig. 5. Bilateral hip dysplasia with painful left hip. The angle between the pelvis and the right thigh was 79 degrees in (a) which increased to 75 degrees in (b) by lifting the right leg.

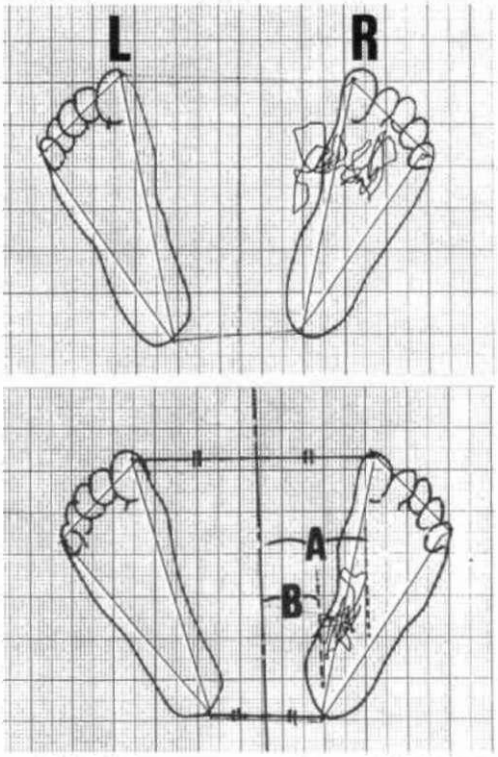

Fig. 6. Body centre analysis with (lower) and without (upper) heel raise. The distance from the midline of each foot was measured, and defined as A: largest and B: smallest. Sound side (right foot) R; right foot, L; left foot.

Gait analysis: Twenty-two patients were assessed. In measuring the motion in the frontal plane, it was assumed that the gait was similar to walking in place. A two-dimensional motion analyzer was used (EMTEK, MVA-2000, Japan). The patients had marks on both shoulders, iliac crests, patellae, and ankles and positional data was collected for five seconds (Fig. 7). The effect of heel lifting in walking was estimated by the angle between the pelvis (the line of both iliac crests) and thigh (the line of the iliac crest and the patella) in the stance phase of the test (Fig. 8). Trunk instability was measured by the relative vertical and horizontal movements of the points of the shoulder and the iliac crest.

Patients: A total of 33 cases (two men and 31 women), involving 35 hips were treated by heel lifting (Table 1). Twenty-four hips needed a valgus effect, so the affected leg was fitted with a raise. Eleven hips needed a varus effect, so the contralateral leg was fitted with a raise. Two

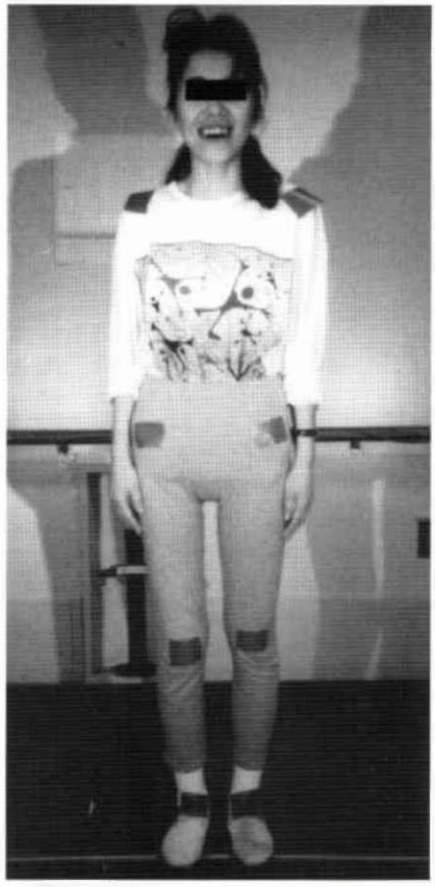

Fig. 7. Motion analysis on the frontal plane. Marking on shoulders, iliac crests, patellae, ankles and walking in place.

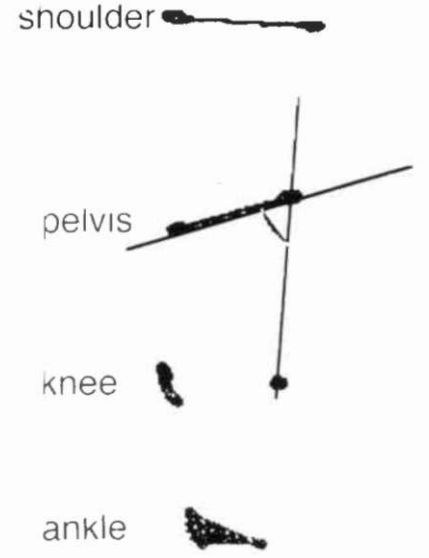

Fig. 8. Motion anaylsis in the frontal plane with the left leg in stance phase. The angle between the line of both iliac crests and the line of iliac crest and patella was measured as shown in the figure. This is the angle between the pelvis and the thigh.

cases suffered from osteoarthritis bilaterally. One hip needed a valgus effect and the other needed a varus effect. Therefore both hips were treated using a single raise. Another case was treated by a raise to produce varus effect but the hip deteriorated and the indication changed requiring a valgus effect, so the hip was treated using two methods. The average age of the patients was 51 years old. Each case was followed for 23 months on average. Statistical analysis of the t-test was carried out.

Results

Twenty-seven cases used a raise at the final follow-up. Two cases did not use one because of absence of hip pain. One stopped using the raise because contralateral pain occurred. Four hips were operated on. Two hips were treated by intertrochanteric osteotomy, and the other two by total hip arthroplasty. The time required for hip pain to decrease or disappear after the heel lifting is shown in Table 2. There is a relationship between the time it took hip pain to decrease or disappear and the stage of osteoarthritis.

Clinical hip scores increased with improvement of pain scores, but range of hip motion and gait ability did not change. Final total hip scores and pain scores significantly improved. The radiological changes of the

Table 2. Pain control by heel lifting

stage	pain decreased	pain disappeared	no change	total
early	3(1)*	3(1)	0	6
advanced	9(1.6)	5(3.6)	2	16
terminal	5(1.2)	5(13.4)	3	13
total	17	13	5	35

*average time of treatment (months) for decrease or disappearance of pain

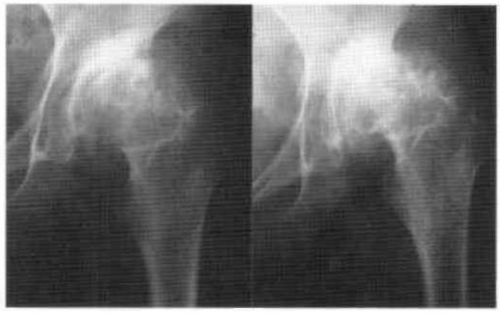

Fig. 9. Left hip radiograms. Left – before and right – 43 months after treatment. Subject was a 60 year old woman with osteoarthritis of left hip treated with a 2cm raise on the left leg. At the 43 month follow-up the hip score showed a 3 point improvement but the hip deteriorated radiographically (cf Figure 1).

osteoarthritis before lifting and at the final follow-up were estimated. Two hips in the late stage of coxarthritis improved, but others deteriorated (11 hips) or showed no change (22 hips) at the final follow-up.

Pelvic radiographs in the standing position with and without heel lifting are shown in Figure 5. The angle decreased on the raised side and a valgus effect was observed in the hip. The angle increased on the contralateral side with a varus effect at the hip. In body centre analysis, fourteen out of 26 patients regained stability by heel lifting, but the other 12 patients did not.

Motion analysis of the angle between the pelvis and the thigh was carried out as shown in Figure 8. In the patients whose hips needed a valgus effect, the angle between the pelvis and the thigh was not changed by the raise and 12 cases out of 15 did not show any effect in the motion analysis. However in the patients whose hips needed a varus effect, the angle between the pelvis and the thigh was changed by the lift; 5 cases out of 7 were affected by the lift. Some 15 patients had their trunks stabilized by lifting and seven patients did not in the motion analysis.

Discussion

The results showed that lifting did not change the natural course of the osteoarthritis. This orthosis is not considered by the authors to be the best method as a conservative therapy, but lifting of the heel is very simple and effective for relief of hip pain and can postpone operative treatment. The orthosis could be a palliative treatment for people who cannot take time from a job or child care and it obviates the need for medication, which can be a cause of gastric or duodenal ulcers. Other orthoses are available but they are large and restrict patients' activity (Hohmann, 1958; Kawamura, 1983).

The biomechanical analyses revealed that some patients were not clearly stabilized by the lift. Each method of analysis is compared with the clinical results as shown in Figure 10 where the solid bar denotes no improvement in the pain score and the empty bar improvement by one or more points. The figure showed the relationship between biomechanical effects and pain scores at the final follow-up. This figure supported the hypothesis that the heel raise affected the stability of the body and improved the hip joint position and also justified the indications according to the rationale of Pauwels' osteotomy.

The more advanced stage of the disease, the longer it took to reduce hip pain. The orthosis did not cause lumbar pain. It even provided lumbar pain relief, because the limb length discrepancy was decreased by the orthosis. This phenomenon would also support the application of a raise for patients with a hip-spine syndrome (Offierski and Macnab, 1983). The orthosis is simple and had a good effect on hip pain, but did not affect the natural course of osteoarthritis of the hip joint.

Acknowledgement

The authors are grateful to Ms. Y. Takemori for preparing the manuscript.

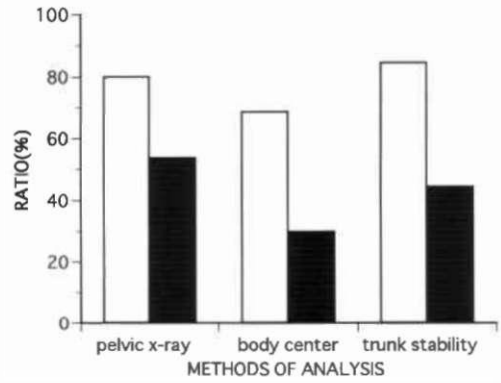

Fig. 10. The ratios of effectiveness and improvement in the pain score. The bars showed the percentage of the positive effects of lifting in each methods of analysis. Note blank bar (improvement by one or more points) were more effective than solid bar (no improvement).

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Prosthetics and Orthotics International, 1997, 21, 159–161

Clinical note

A young female patient with reflex sympathetic dystrophy of the upper limb in whom amputation became inevitable

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**Department of Rehabilitation, University Hospital Groningen, The Netherlands*
*** Department of Internal Medicine, University Hospital Groningen, The Netherlands*
**** Department of Surgery, University Hospital Groningen, The Netherlands*

Abstract

Reflex sympathetic dystrophy (RSD) is characterized mostly by: (burning) pain, restricted range of motion, oedema and autonomic disturbances. Amputations in case of RSD patients should only be performed in cases of a dysfunctional limb, life threatening conditions such as untreatable infections or in cases of unbearable pain. The authors describe a patient in whom amputation became inevitable because of threatening infections.

Introduction

RSD may develop after a variety of inciting events, but mostly after a trauma or surgery. RSD can however also arise after a minor sprain or contusion which a patient does not recall and did not consult a physician for. Most therapies are assumed to have little or only temporary success (Veldman, 1995), and vary from physical and occupational therapy to many methods to inhibit the sympathetic nervous system (Veldman, 1995). Limb amputation is the extreme choice in management of RSD patients (Erdmann *et al.*, 1992; Eyres *et al.*, 1990; Geertzen *et al.*, 1994; Ritt *et al.*, 1992; Rohrich *et al.*, 1987; Veldman 1995). It is restricted only to those cases with irreversible changes leading to loss of function or intolerable pain.

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Case report

A 26-year-old woman presented with a painful non-dominant left hand and some oedema at the proximal interphalangeal joint of the second digit. She had the complaint for half a year and did not recall an inciting event. She had already consulted a vascular surgeon and a rheumatologist. X-rays, the vascular laboratory research testing and general blood tests showed no abnormal results, no evidence of any bony, joint or nerve injury. All four fingers were lightly flexed and were cold, cyanotic and swollen and she had allodynia and hyperpathia (Fig. 1). Her social life was severely restricted (Geertzen *et al.*, 1994). She was still working as a full-time employed typist. It was concluded that this patient could have a RSD although the history and physical examination were not classical. Intensive physical and occupational treatment was started in combination with medication

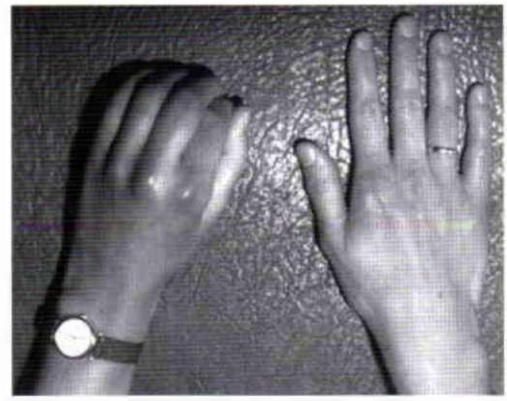

Fig. 1. The patient on presentation.

(analgesics, oxygen free radical scavengers and oral vasodilators). In two weeks the clinical symptoms deteriorated and the patient had so much pain that treatment was intensified. A psychologist was consulted and could not find any explanation or suggestion for treatment.

The clinical course was undulating and half a year later she was referred to a pain clinic. For more than two years she was treated with physical treatment, stellate blocks, morphine, oral vasodilators and a surgical thoracic sympathectomy. Success was temporary and she was re-admitted with severe pain, allodynia, hyperpathia, hyperhidrosis and changed nail-growth. The hand was clenched, swollen and could not be opened. The hand and wrist were dysfunctional (Fig. 2), and she developed cellulitis and several ulcera leading to lymphangitis and enlarged lymphatic nodes. Consultation with a rheumatologist, vascular surgeon, trauma surgeon and a specialist in rehabilitation medicine led to the decision about amputation, which was performed 10cm proximal to the wrist. Post-operatively the wound healed "per primam". The pathologist reported a dystrophic hand with microscopical minimal aspecific alterations such as atrophic muscles and perivascular infiltrations. At first she had stump pain, but one year later she only has phantom sensations when not wearing her myoelectric prosthesis. She gets an 80% disability payment.

Discussion

RSD is a very complex disease and differentiation of a normal recovery after a trauma may be very difficult. The differential diagnosis is often not easy; one should consider compartment syndrome, phlebothrombosis, infection, inflammatory conditions, rheumatologic disorders, neurologic diseases, arterial insufficiency or psychologic disorders such as automutilation or a dis(non)use of the limb.

In the case reported here all classical clinical signs of RSD became present. The patient had asked more than once for an amputation. At that time she was strongly advised against amputation. After three and a half years, with several episodes of clinical deterioration a syndrome almost similar to that of the clenched fist syndrome developed (Swift *et al.*, 1995).

Although the authors realized the possibility of the recurrence of RSD in the stump, amputation had to be performed because of the threatening infections. RSD is reported to recur after amputation in most cases (Dielissen *et al.*, 1995), although one author did not support this opinion (Stam *et al.*, 1994). Dielissen also reported that most patients were, because of this recurrence of RSD, unable to wear a prosthesis. Phantom pain was reported in 71% of the cases and phantom sensations in 85% of the cases. It was conculded that in RSD patients amputation should not be performed because of pain. Szeinberg-Arazi (1993) reported that patients with a post-RSD amputation require psychological support. In this case the patient declined support and for the time being, she is doing surprisingly well.

The main reason for amputation in this case report was increasing, ascending and poorly treatable infections and the disfunctional hand; in the patients' view it was the unbearable pain.

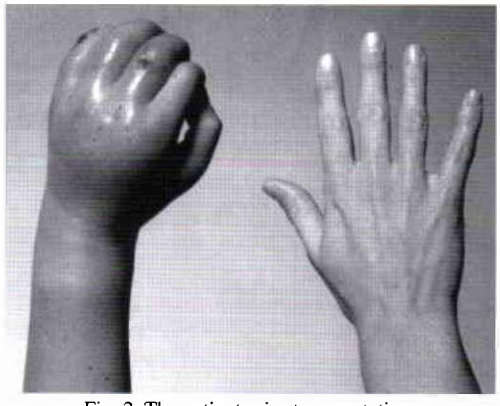

Fig. 2. The patient prior to amputation.

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Book Review

Orthopadie-technische Indikationen
(Orthopaedic Technical Indications)
Andre Bähler.
Verlag Hans Huber, Bern; Gottingen; Toronto; Seattle, 1996.
pp.592, illustrated.

This is probably the most comprehensive book in Germany on orthotics and related topics, reminiscent of Dr. Sidney Licht's book Orthotics Etc. The book includes areas beyond upper and lower limb and spinal orthotics, such as indications, techniques and medical reasons, descriptions and illustrations for stomas, e.g. colostomy, phlebology and lymphodema and their orthotics or appliance management. It also includes a short section on wheelchair seating systems.

The author is one of the most accomplished and best known orthotists/prosthetists, not only in Europe, but in the world. He has recently been named Honorary President of Interbor, the International Association of Orthotists and Prosthetists.

The introduction is by Professor Dr. med. Norbert Gschwend, an orthopaedic surgeon, who has worked with Mr. Bähler for several decades, among other areas in the development of internal prosthetic systems. Professor Gschwend feels, as I feel, this to be an important book of reference for orthopaedic surgeons, rheumatologists and paediatricians. It is not only a well written book from a technical point of view, but well organized with probably the finest illustrations I have ever seen. They are very clear with many of them computer generated and illustrative of the points the author wishes to get across. Without these fine illustrations this book would not be quite what it is in spite of the elegant, up-to-date text.

The first chapter is devoted to biomechanics, related terminology, selected laws of physics and their relationship to orthoses. In the section on lower limbs, there is a contribution by Dr. Alvin Jacob of the University Clinic in Frankfurt regarding examination techniques of the knee joint. All sections of the book are preceded by a description of the disease categories or injuries that lead to appropriate indications for various orthotic treatments.

A separate chapter deals with the descriptions of paralyses and muscle diseases, e.g. poliomyelitis, cerebral palsy, etc., including spinal cord injury. The chapter on stoma management is extraordinary as it is not found in ordinary textbooks in orthotics. The major contributor to this chapter was Bernd Hempell of Berlin.

This book is certainly the latest and most comprehensive book in the area of technical orthopaedic indications, i.e. orthotics, etc., for not only orthotists but orthopaedists, therapists and others interested in the area of orthotics rehabilitation.

H. Richard Lehneis
Research Associate Professor of Rehabilitation Medicine
New York University Medical Centre

Calendar of Events

4-6 September, 1997
5th Scientific Meeting of the Scandinavian Medical Society of Paraplegia, Reykjavik, Iceland.
Information: Sigrun Knutsdottir, Organizing Committee, Dept. of Rehabilitation and Neurology, Reykjavik, Iceland.

8-12 September, 1997
Dundee 97: International Conference on Wheelchairs and Seating, Dundee, Scotland.
Information: Conference Secretariat, Dundee 97, Meeting Makers, 50 George St., Glasgow G1 1QE, Scotland.

9-13 September, 1997
AOPA National Assembly, Charlotte, USA.
Information: Annette Suriani, AAOP, 1650 King St., Suite 500, Alexandria, VA 22314, USA.

10-14 September, 1997
Annual Meeting of the American Congress of Rehabilitation Medicine, Boston, USA.
Information: ACRM, 4700 W. Lake Ave., Glenview, IL 60025-1485, USA.

14-19 September, 1997
World Congress on Medical Physics and Biomedical Engineering, Nice, France.
Information: Nice 97, SEE-48, Rue de la Procession, F 75724 Paris, Cedex 15, France.

18-21 September, 1997
12th National Congress of FIOTO, Bari Italy.
Information: Consulta Umbria, Piazza Italia, 9-06121 Perugia, Italy.

24-27 September, 1997
Annual Meeting of the American Society of Biomechanics, South Carolina, USA.
Information: Vasanti Gharpuray, Dept. of Bioengineering, Clemson University, 501 Rhodes Research Centre, Clemson, SC 29634-0905, USA.

14-19 October, 1997
Eurospine 97, Zurich, Switzerland.
Information: Ess, J Riechert Schild, c/o Schalthess Klinik, Lengghalde 2, CH-8008, Zurich, Switzerland.

20-23 October, 1997
20th Singapore Orthopaedic Association Annual Meeting, Singapore.
Information: Dr. O. Thiagarajan, Dept. of Orthopaedic Surgery, National University Hospital, Lower Kent Ridge Rd., Singapore 119704.

22-25 October, 1997
REHA 97, Dusseldorf, Germany.
Information: Dusseldorf Trade Shows Inc., 150 N. Michigan Ave., Suite 2920, Chicago, IL 60611, USA.

22-26 October, 1997
North American Spine Society Annual Meeting, New York, USA.
Information: North American Spine Society, 6300 N. River Rd., Suite 500, Rosemont, IL 60018-4231, USA.

23-24 October, 1997
ISPO France Second National Congress, Lyon, France.
Information: Bawan Evènement, Immeuble Rive Gauche, 12 Rue Cavenne, 69007 Lyon, France.

23-25 October, 1997
25th Annual Scientific Meeting of ISPO UK National Member Society, Scotch Corner, near Darlington, England.
Information: Mrs. P. McLauchlan, Orthotic Dept., Perth Royal Infirmary, Western Ave., Perth PH1 1NX, Scotland.

29 October-2 November, 1997
19th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, Chicago, USA.
Information: Meeting Management, 2603 Main Street, Suite 690, Irvine, CA 92714, USA.

November, 1997
Hong Kong Orthopaedic Association Annual Meeting, Hong Kong.
Information: Dr. Ngai Wai Kit, Dept. of Orthopaedic Surgery, TVZN Mun Hospital, Hong Kong.

13-16 November, 1997
American Academy of Physical Medicine and Rehabilitation Annual Meeting, Atlanta, USA.
Information: AAPM&R, 1 IBM Plaza, Suite 2500, Chicago IL 60611, USA.

27-29 November, 1997
3rd International Congress on Scoliosis Research (SIRER), Lyon, France.
Information: Dr. J. C. Bernard, Congress Secretariat, Centre des Massues, 92 rue E Locard, 69005 Lyon, France.

27-29 November, 1997
4th Spanish Fair of Orthopaedics, Rehabilitation and Related Techniques, Valencia, Spain.
Information: Feria Muestrario Internacional de Valencia, Orprotec '97, Apartado Correos : 476, E-46080 Valencia, Spain.

3-6 December, 1997
9th International Conference on Biomedical Engineering, Singapore.
Information: Ontako Conference Secretariat, Block 9, Kallang Place, #04-09 Singapore 339154.

1998

28 June-3 July, 1998
9th World Congress of the International Society for Prosthetics and Orthotics, Amsterdam, The Netherlands.
Information: Congrex (Holland) BV, PO Box 302, 1000 AH Amsterdam, The Netherlands.
Tel: (+31) 20 5040 201. Fax: (+31) 20 5040 225.

10-12 September, 1998
Second Central and Eastern European ISPO Conference, Portorose, Slovenia.
Information: Ela Loparič, Institute for Rehabilitation, Linhartova 51, 1000 Ljubljana, Slovenia. Tel: (+386) 61 1376 600, Fax: (+386) 61 1376 589, E-mail: crt.marincek@ir-rs.si.