
The Journal of the International Society for Prosthetics and Orthotics

# Prosthetics and Orthotics International

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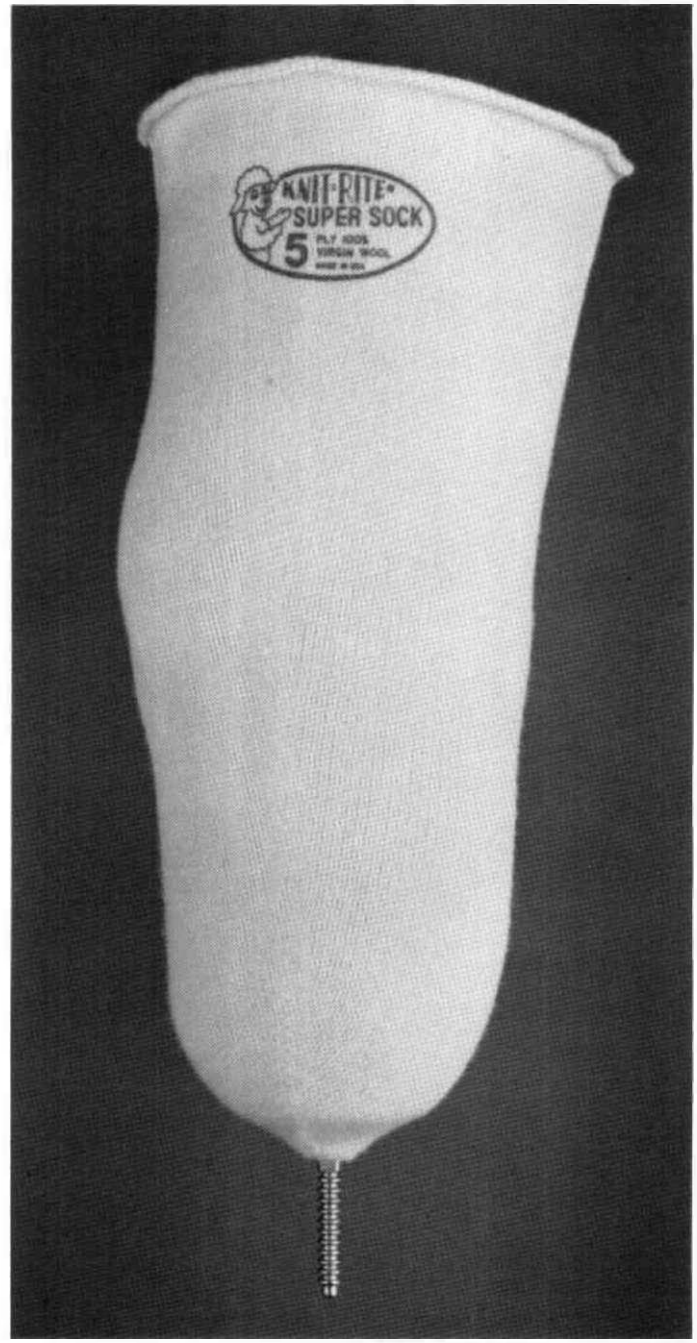


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# Editorial

ISPO is the only international, multidisciplinary society involved in prosthetics, orthotics and related topics. The strength of the Society lies in its membership. This is true not only in numerical terms, but also due to the depth of professional interests that the Society covers, the widespread geographical locations of its members and the different cultures that it represents. This diversity in membership provides the basis for a forum on the exchange of information at national, regional and international levels.

The past ten years have seen a significant growth in membership. In 1985 the Society had 1,850 members from 62 countries. By 1995 the numbers had increased to 2,724 from 79 countries. During this period the number of National Member Societies has increased from 14 to 28. As a result of this growth the influence of the Society has increased dramatically at national and regional levels.

The importance of National Member Societies cannot be over-emphasised. They are the means by which the majority of ISPO members can participate and influence the standards of prosthetics and orthotics service and practice in individual countries. National Member Societies offer a multidisciplinary forum to debate and influence local issues and, for the most part, provide a continuing focus through organising national meetings and publishing newsletters. Many act as a consultative body to national governmental agencies.

National Member Societies also have a direct influence on the way the International Society is run. They select representatives to the International Committee which is responsible for ensuring that the policies, plans and activities of the Society reflect the aims and aspirations of the membership as well as its professional, cultural and geographic variations. The International Committee has the responsibility of electing the Executive Board and evaluating and influencing the activities of the Society. In recent years the International Committee has met regularly, once at the time of the Congress and again midway through each triennium. These meetings have resulted in a positive exchange of information and ideas between the International Committee and the Executive Board. The next meeting of the International Committee is due to take place in January, 1997.

The number of new National Member Societies has greatly increased the influence of ISPO in regions hitherto untouched. In particular over recent years there has been a significant growth of activities in Central and South America and in Central Europe.

In Central and South America, National Member Societies have been formed in Argentina, Caribbean, Colombia and Panama with interest being shown by other countries in the region. The latest activity held in the region was an International Conference in Trauma organised by ISPO - Colombia in Bogota, in July 1996.

In Central Europe, National Member Societies have been formed in Hungary and Slovenia and other countries in the region have shown an interest. Earlier this year the First Central European ISPO Conference was held in Balatonfüred, organised by ISPO-Hungary and attended by participants from surrounding countries (a report of this meeting can be found on p.70 in this issue of the Journal).

Events such as these greatly enhance the influence of the Society on the region, and the contribution that it can make to the individual member. However these activities are only possible due to the strenuous efforts of individual members working in collaboration with one another. If the Society is to grow in strength and influence efforts should be made to increase membership, the number of National Member Societies and the range of activities. This can only be done through the continuing efforts of the individual member.

Norman A. Jacobs
*President-Elect*

# First Central European ISPO Conference

## 30 May - 1 June 1996
## Balatonfüred, Hungary

We have already announced in these columns the formation of ISPO Hungary in December 1994, having been fostered by Ed van Laar of ISPO Netherlands. In 1995 we organised a one day introductory event (a scientific conference and exhibition) and successfully approached the medical postgraduate authorities to make prosthetics and orthotics an integral part of the training curriculum of the relevant medical specialities (orthopaedics, traumatology and rehabilitation). In order to offer an educational option, in October 1995 ISPO Hungary organised a one week basic training course for junior doctors who were likely to become one of the above specialists. For this course the lecturers were mainly members of ISPO Hungary. The daily programme covered theoretical classes in the morning and practical demonstrations and rehabilitation practices in the afternoon sessions. Each of the five days was focused on one profile: lower limb prosthetics, upper limb prosthetics, limb orthotics, spinal orthotics and footwear, respectively. The speakers were medical doctors, prosthetists and orthotists, as well as physiotherapists. There were 19 participants, and more are expected for the course this year.

In 1996 we again organised a one day postgraduate training course, focusing on medical-prosthetic collaboration and funding in the state model of Health Care. Also in 1996 we undertook the duty of expanding ISPO influence in the region, so we organised a three-day event, namely the *First Central Eastern European ISPO Conference*. We are thankful for the enormous help and support of the ISPO Executive Board and ISPO Netherlands. The Patron of the Conference was Mr. S. Sára, film director and chairman of "Duna" Television Station, which specialises in features mostly about Hungarian culture within and around Hungary, broadcast via satellite to a large catchment area. Each of the conference days covered instructional lectures of a specific topic (lower limb prosthetics, lower limb orthotics and spinal orthotics), with speakers such as Mr. Andries de Bont, Mr. David Condie, Ms Elizabeth Condie, Ms. Bettina Grage, Mr. Fred Holtcamp, Mr. Ed van Laar, Mr. F. Lefeber, Mr Andreas Würsching, Mr B. Zinnemers as well as Hungarian speakers. For the afternoon of the second Conference day we organised a "Country profiles panel", in which invited prominent personalities of neighbouring countries brought a concise status report on prosthetics and orthotics in their homelands. The panel consisted of Prof. Crt. Marincek from Slovenia, Dr. W. Ott from Austria, Mr. L. Székely from Romania, Mr. M. Válent from Slovakia, Mr. Zivkovic from Croatia and last but not least Mr Lajos Kullmann from Hungary. To our great regret, the invited speaker from Ukraine did not arrive. There were technical demonstrations in the coffee breaks and an interesting exhibit which contained 17 sponsors (French, Italian, German, American and domestic ones). The smooth running of the Conference was assured by the Hungarian organisers "Mádai and Partners Company". The final farewell words of the closing ceremony went to Professor Marincek, who invited all 124 registered delegates to reunite again in two years time from now in Slovenia on the occasion of the *Second Central Eastern European ISPO Conference* in September, 1998.

Budapest, 20 June, 1996

G. László MD, *Chairman*
L. Ágoston, *Secretary General*

ISPO Hungary

# Award for Former Vice-President

Minister of Civil Affairs, Doji Cering, presents a Golden Ox to Sepp Heim in the presence of the German Ambassador to China.

SEPP HEIM, a former Vice-President of ISPO, was awarded a top honour of the Ministry of Civil Affairs of China on Friday 26 January 1996 in Beijing for his three years of hard work there.

Heim was presented with a Golden Ox, representing the "herd boy's willing ox" (a servant of the people) award. Doji Cering, Minister of Civil Affairs said "the honour and the prize are to praise Heim's outstanding performances as head of the China Training Centre for Orthopedic Technologists (CHICOT) in Wuhan, capital of Hubei Province, since 1992 and show him our respect."

Aside from K'uanshu Li, a Chinese-American who was given the award for his contribution to the country's disabled population in 1992, Heim is the first foreigner to be awarded the honour by the Chinese Government.

"In the last three years, Heim has devoted himself to the development of China's orthopaedic education, playing a key role in getting financial aid from the German Government for the founding of the centre, setting up a first-class artificial limbs school in China" Wu Zhongze, Director of the Ministry's Welfare Department, said.

Sepp Heim, has offered many valuable suggestions to the Chinese authorities trying to set up an orthopaedic technology education system in line with the international standards.

According to Wu the Wuhan-based CHICOT, built with funds granted by the German Government, is equipped with modern machinery and tools imported from Germany.

"It is vital for China to train more professional prosthetists and orthotists to help people with disability in the future" Wu informed the audience.

*Prosthetics and Orthotics International, 1996, 20, 72–78*

# Clinical rehabilitation of the amputee: a retrospective study

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***Rehabilitation Centre Beatrixoord, Haren, The Netherlands.*
****Department of Social Medicine, University of Groningen, The Netherlands.*

**Abstract**

The aim of this study was to determine the rehabilitation outcome of lower limb amputee patients after clinical rehabilitation. Altogether 183 amputee patients admitted for clinical rehabilitation in the years 1987-1991 were reviewed by retrospective analysis of medical record data. Three groups of amputee patients were identified by reason for amputation. The vascular group: (N = 132), mean age 67 years, mean admission time 119 days, 85% prosthetic fitting.

The oncology group (N = 15), mean age 55 years, mean admission time 77 days, 60% prosthetic fitting. The traumatic amputee group: (N = 14), mean age 41 years, mean stay 134 days and 100% prosthetic fitting. Some 22 patients were bilateral amputees and were assessed separately. The most important reasons for not fitting a prosthesis were oncological metastases, stump and wound healing problems.

After rehabilitation 86% of all patients could be discharged home. These results are more favourable than those seen in previous studies.

## Introduction

In the Netherlands in 1992, 1,551,945 hospital admissions were registered by the Central Bureau of Statistics (10.2 per 100 inhabitants) (Dutch Heart Foundation, 1994). These admissions included 2000 lower limb amputations. The total number of lower limb amputations, from sacroiliac to transmetatarsal level, has not changed in the Netherlands over the last 5 years (National Medical Register, 1993). Some 88% of all amputations are at trans-tibial and trans-femoral level. Detailed information on these amputees is not available at the moment. For survival rates and causes of amputation in the population we have to depend on population statistics from neighbouring countries (Fowkes 1988; Ebskov, 1991; Stewart and Jain, 1993). In the population of amputee patients age is increasing (National Medical Register, 1993). The population of amputee patients is predominantly over 60 years of age and amputation is mostly due to vascular problems. After amputation surgery most patients are seen by a rehabilitation medicine physician and after consultation the need for clinical rehabilitation is specified. Most amputees are transferred home or to a nursing home before prosthetic fitting.

Only a small percentage (approximately 15%) of this group are admitted to a rehabilitation centre. There they are trained in Aid to Daily Living (ADL) activities. There is a postoperative phase during which the stump is forming, prosthetic training with an interim prosthesis and appliance and training with a lower limb prosthesis. Patients receive an individual programme in a multi-disciplinary approach with daily individual and group exercises. The local government is informed of any home adjustments which are required.

There are also educational classes for the patients and their spouses and psychological support is given to cope with their emotional problems due to the loss of a part of the leg. Information about the rehabilitation outcome of this group of amputee patients is scarce. This article is mainly written to analyse the population of the amputee patients in a rehabilitation centre and to determine the rehabilitation outcome for this group. The authors have therefore carried out a first retrospective analysis of data in the north of the Netherlands.

The Rehabilitation Centre Beatrixoord in

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Haren provides clinical rehabilitation facilities for a catchment area of 1,000,000 people in the north of the Netherlands. The centre offers a total rehabilitation programme for patients with major disabilities. Amputee patients are managed in a separate ward containing 32 beds. The rehabilitation team consists of medical staff responsible for the overall rehabilitation treatment, nursing staff, physiotherapist, occupational therapist, speech therapist, social worker, clinical psychologist, a dietician and a prosthetist. Prosthetic manufacturing is carried out in the nearby workshop for posthetics and orthotics.

## Methods

Between 1987 and 1991 in the three northern provinces of the Netherlands 1037 patients had a lower limb amputation carried out at trans-femoral, knee disarticulation or trans-tibial level. (National Medical Register, 1993). Amputation surgery is performed in all hospitals in the northern region; one university and 16 general hospitals. In order to determine which group of patients were transferred to the rehabilitation centre in the above mentioned period, all admissions to the rehabilitation centre from 1987-1991 were reviewed. All patients admitted because of a lower limb amputation were selected by their International Classification of Disease (ICD) code. All medical records were studied in order to check if the patients were admitted after amputation surgery. The selection of records of lower limb amputee patients from the above mentioned five year period were studied in detail. Information was found in the discharge letter to the patient's general practitioner or information written down by the rehabilitation team members. All selected patients were admitted for prosthetic training and application of their first lower limb prosthesis, or revision of their currently used prosthetic device. Data obtained from the medical records included: age on admission, sex, referring hospital, amputation level and side, reason for amputation, need for re-amputation, date of amputation, date of admission, length of stay, use of lower limb prosthesis on discharge, date of discharge and discharge destination. All the results were made anonymous and were tabulated on a spreadsheet database for further analysis.

## Results

### *All amputees*

In the studied period from 1987-1991, 183 lower limb amputee patients were admitted to the rehabilitation centre.

The age distribution of all 183 amputees is illustrated in Figure 1, giving an overview of

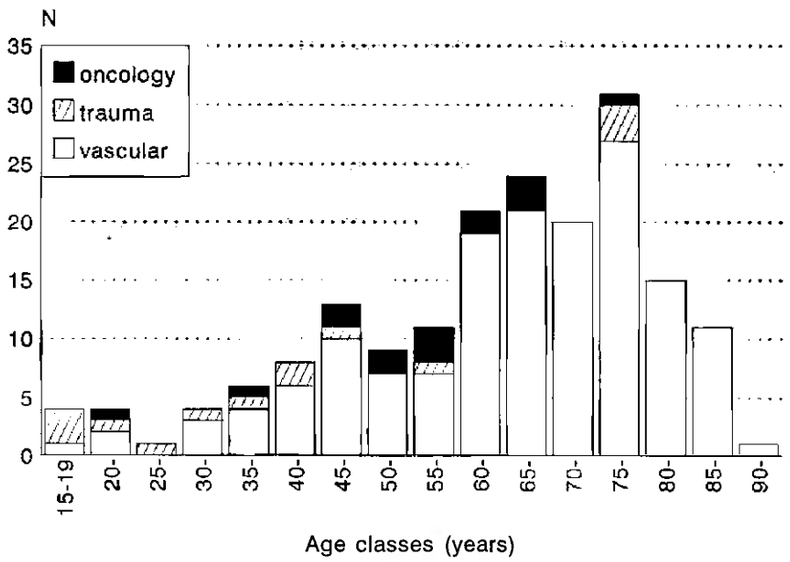


Fig. 1. Amputees admitted in 1987-1991 (N = 183). Distribution by cause of amputation and age.

Table 1. Amputation level of unilateral amputees (N = 161) with the related reason for amputation, age and sex

| Amputation level | N | Reason for amputation (N) | Age (years) mean | Age (years) SD | Male (N) | Female (N) |
|---|---|---|---|---|---|---|
| Hemipelvectomy (HP) | 3 | oncology: 3 | 49 | 10 | 3 | 0 |
| Hip disarticulation (HD) | 4 | oncology:3<br>vascular: 1 | 40.7<br>66 | 21<br>0 | 1<br>1 | 2<br>0 |
| Trans-femoral (TF) | 60 | oncology: 7<br>trauma: 4<br>vascular: 49 | 59.1<br>52.2<br>67.2 | 8<br>17<br>14 | 6<br>4<br>34 | 1<br>0<br>15 |
| Knee disarticulation (KD) | 4 | trauma: 2<br>vascular: 2 | 25.5<br>54.5 | 8<br>12 | 2<br>2 | 0<br>0 |
| Trans-tibial (TT) | 89 | oncology: 2<br>trauma: 8<br>vascular: 79 | 68<br>39.6<br>66.6 | 13<br>25<br>14 | 2<br>4<br>41 | 0<br>4<br>38 |
| Ankle disarticulation | 1 | vascular: 1 | 59 | 0 | 1 | 0 |
| Total | 161 | | 63.3 | 16 | 101 | 60 |

the amputee population related to cause of amputation. The vast majority (82%) are due to vascular problems. The male : female ratio is 1.6:1. The mean age is 62.9 years (range 15-90), 68% of the patients are over 60 years of age.

The average admission time for all patients was 123 days (median 108 days; range 2-524 days).

Of these 183 patients 22 had a bilateral amputation. In the studied period 7 of these 22 were previously admitted as unilateral amputees.

### *Unilateral amputees*

Table 1 gives an overview of all 161 unilateral amputees. It illustrates all amputation levels from hemipelvectomy to ankle disarticulation level. Altogether 154 amputations (95.7%) were performed at trans-femoral, knee disarticulation or trans-tibial level. The high level of hemipelvectomy and hip disarticulation amputation was in all but one case, due to oncological pathology. The vascular case was a re-amputation of a trans-femoral amputation due to wound complications. The vascular group includes most cases: 132/161 = 82.0% of all unilateral amputees. Their mean age was 66.6 years and the mean admission time in the centre was 119 days. In this group 85% of the amputees could successfully be fitted with a lower limb prosthesis.

The admitted group of amputees after an oncological lower limb amputation was rather small: 15/161 = 9.3%. Their mean age of 55.2 years was less than that of the vascular group. The mean admission time in this group was 77 days. Sixty percent of these amputees were successfully fitted with a prosthesis.

The trauma group includes 14 patients (14/161 = 8.7%) and was youngest with a mean age of 41.2 years. Their mean period of admission of 134 days, was slightly less than that of the vascular group. In the trauma group all amputees were successfully fitted with a prosthesis.

Of all 161 unilateral amputee patients admitted 83.9% were successfully fitted with a prosthesis on discharge. As shown in Table 2; 26 of these unilateral amputee patients were discharged without a prosthesis. Twenty (20) of them were vascular patients and 6 of them were amputated due to oncological problems. Metastases, stump problems, and wound healing problems were reported in 57.7% of the cases. Five (5) of these 26 amputees were able to use a prosthesis after continuation of their training in outpatient rehabilitation treatment. After the clinical rehabilitation period and outpatient treatment 139/161 = 86.3% of the unilateral amputee patients could use their prosthesis in their own environment.

After clinical rehabilitation 151 of the 183 patients could be discharged home. Five (5) of the 12 patients living in a home for the elderly, were discharged back to their previous environment. Eight (8) patients were discharged to a nursing home. Eleven (11) amputees were

Table 2. The unilateral amputees discharged without a lower limb prosthesis (N = 26). Indication problems preventing use of a prosthesis with related average age, average admission time and discharge destination. (HP = Hemipelvectomy; HD = Hip disarticulation; TF = Trans-femoral; TT = Trans-tibial

| Reason for not using a prosthesis | N | Age (years) (mean) | Level | Admission time (days) (mean) | Discharge destination | Prosthesis after discharge yes/no |
|---|---|---|---|---|---|---|
| metastases, oncology | 5 | 61.6 | HD 1 | 42.2 | hospital | no |
| | | | HD 1 | | home | yes in out-treatment |
| | | | TT 3 | | hospital: 3 | no |
| Stump problems | 5 | 63.8 | TF 4 | 112.2 | hospital: 2 | no |
| | | | | | h. elderly: 1 | no |
| | | | | | home: 2 | yes 1 in out-treatment |
| | | | TT1 | | hospital: 1 | no |
| Wound healing problems | 5 | 57.6 | HP 1 | 78.6 | hospital | no |
| | | | HD 1 | | home | yes in out-treatment |
| | | | TF 1 | | home: 1 | no |
| | | | TT 2 | | hospital: 2 | no |
| Fractures | 2 | 58 | TT 2 | 31.5 | h. elderly: 1 | no |
| | | | | | home: 1 | no |
| Mental disorders | 2 | 74.5 | TF 1 | 148.5 | nurs. hom-psych | no |
| | | | TT 1 | | nurs. home | no |
| Fear of walking | 1 | 37 | HP 1 | 72 | home | no |
| Internal problems | 1 | 60 | TT 1 | 79 | hospital | no |
| ADL problems | 2 | 77.5 | TF 1 | 37 | home | no |
| | | | TT 1 | | home with family | no |
| Conditional problems | 1 | 66 | HD 1 | 91 | home | yes in out-treatment |
| Dead | 1 | 78 | TT 1 | 32 | home | no |
| Unknown | 1 | 31 | TT 1 | 28 | home | yes in out-treatment |
| Total | 26 | 61.8 | | 73.1 (SD 61.88) | | |

discharged to a hospital due to complications after surgery or because of internal and oncological complications. After the rehabilitation treatment 86% of all patients could be discharged to their own environment.

***The bilateral amputees***

The 22 bilateral amputees were studied separately. This is because a bilateral amputee has specific problems during the rehabilitation process. Not only does the patient have to train with two prostheses but also the increased energy requirement needs special attention.

The 22 bilateral amputees, mean age: 59.5 years (range 17-83), were studied by amputation level.

All patients were admitted for prosthetic training. The average admission time was 170 days (range 15-524 days).

Table 3 illustrates the amputation level on

Table 3. Bilateral amputees (N = 22) with individual amputation level, age, sex, time between unilateral and bilateral amputation, and admission time (TF = Trans-femoral; KD = Knee disarticulation; TT = Trans-tibial).

| Amputation level | | N | Reason for amputation | Age (years) | | Sex | | Time between amputations (days) | Admission time (days) |
|---|---|---|---|---|---|---|---|---|---|
| L | R | | | mean | range | m | f | | |
| TF | TF | 3 | Vascular | 67.5 | 64-70 | 1 | 2 | 528 | 121.6 |
| TF | TT | 1 | Vascular | 79 | | 1 | 0 | 2992 | 190 |
| KD | TF | 1 | Vascular | 42 | | 0 | 1 | 103 | 209 |
| KD | KD | 1 | Vascular | 17 | | 0 | 1 | 0 | 316 |
| KD | TT | 1 | Vascular | 68 | | 1 | 0 | 2807 | 107 |
| TT | TT | 11 | Vascular | 67.5 | 32-83 | 8 | 3 | 462 | 155.5 |
| TT | TT | 4 | Trauma | 39.5 | 22-46 | 1 | 3 | 0 | 221.7 |
| Total/mean | | | | 59.5 | 17-83 | 12 | 10 | 571 | 170 |

both sides of all bilateral amputees. Some 18.2% were due to traumatic causes, 81.8% of the bilateral amputations were caused by vascular problems.

The average time between first and second limb amputation was 571 days. All traumatic amputees underwent amputation on both sides in the same surgical session. The discharge destination of the bilateral amputees after clinical rehabilitation was as follows:–

Home: 18 (81.8%); Home for elderly: 2 (9.1%); Nursing home: 1 (4.5%); Hospital: 1 (4.5%). After rehabilitation 15/22 (68.2%) of the bilateral amputees were fitted with prostheses on both sides. The 4 bilateral trans-tibial traumatic amputees were able to walk without walking aids. They were all discharged home. Of the 18 vascular amputees 11 received a prosthesis on both sides; 7 patients were discharged without a lower limb prosthesis (Table 4). They were all vascular amputees with multiple medical problems. In most cases stump problems and general condition were the most important reason for not using a prosthesis.

## Discussion

The group investigated is a selection of the total amputee population in the 3 northern provinces of the Netherlands. Of all unilateral amputees 85% were fitted with a lower limb prosthesis, which they used frequently. In this group the results are more favourable compared with other authors (Beekman and Axtel, 1987; Houghton *et al.*, 1992). Helm *et al.* (1986) found comparable results with the findings reported. In the rehabilitation group admitted to the rehabilitation centre 68% of the amputees were over 60 years of age. The reason for amputation in 82% of the cases was due to vascular problems. This correlates well with the results of Stewart and Jain (1993) in Dundee.

The proportion of oncology amputees in this study is larger than in other series (Ebskov, 1991). This may be due to the presence of the university hospital in the area, where there is access to special facilities for cancer treatment. Oncology amputations are more often proximal at hemipelvectomy and hip disarticulation level, and need special attention to prosthetic fitting. Using a prosthesis with this high level of amputation demands extra energy (Fowkes, 1988; Jaegers, 1993). The oncology group (mean age 55.27 years) was relatively young and 60% of these amputees were successfully fitted with a prosthesis. Amputation level selection in this field of amputation surgery depends primarily on the location and nature of the tumour.

The average admission time in this small group was 77.73 days. The traumatic group was the youngest (mean age 41.21 years). Amputation was mostly due to traffic accidents and accidents with agricultural farming machines. All traumatic amputees were fitted with a prosthesis. This group is in good general condition and is able to deliver the extra energy required for walking with a prosthesis. Six (6) patients had training for their first prosthesis. Others were admitted in order to revise their existing prosthesis. The relatively long admission time (134.15 days) can be explained by the need for stump revision and the prosthetic problems that had to be solved for adequate function. The patients admitted for renewal of their prosthesis had an admission time of 93 days. The age of vascular amputee

Table 4. Bilateral amputees discharged without a prosthesis (N = 7). Amputation level, age, reason for not fitting a prosthesis and discharge destination are displayed. (TF = Trans-femoral; KD = Knee disarticulation; TT = Trans-tibial).

| Amputation level | | | Reason for not fitting | | Discharge |
|---|---|---|---|---|---|
| L | R | Age | a prosthesis | Sex | destination |
| TF | TF | 69 | general condition | F | Home |
| TF | TF | 64 | stump problems, takes wheelchair | M | Home |
| TF | TF | 70 | general condition | F | Hospital |
| KD | TF | 42 | hand problems, dialysis | F | Home |
| TT | TT | 75 | stump problems, depression | M | Home for elderly |
| TT | TT | 65 | prefers wheelchair | F | Home with family care, advice nursing home |
| TT | TT | 66 | stump problems, in 1993 mobilisation with prosthesis | F | Home |

patients is predominantly greater (mean age 66.77 years).

This group is comparable with other studies in this area (Boontje, 1980). The average admission time was 119 days which is shorter than for the traumatic amputee group. The vascular group consists of vascular and diabetic patients. In a lot of cases vascular surgery was performed prior to amputation as is advised by the European Consensus Group on Critical Limb Ischemia (Deerochanawong *et al.*, 1992).

The vascular amputees receive a personal training programme. Stair climbing and bicycle riding training will be provided if necessary. Even the amputees over 90 years of age can successfully be fitted with a prosthesis if their general condition is good. The overall quality of life of amputees is likely to be enhanced by focusing rehabilitation efforts on improving mobility (Pell *et al.*, 1993). The authors realise that approximately 15% of all lower limb amputees in the region are admitted to a rehabilitation centre to obtain a prosthetic device if possible. Little is known about the provision of lower limb prostheses in nursing homes and homes for the elderly. More study is needed in this field. Buyk and Stephan (1988) showed that most patients who were provided with a prosthetic device used it a long time after discharge from hospital. After clinical rehabilitation 86% of patients could be discharged home (5 of the 13 patients discharged to a home for the elderly lived there already before the amputation). This result is consistent with those of Ebskov (1991) who stated that 84% of patients returned to their homes. This only applied to trans-tibial amputees. It is not clear however if these patients were fitted with a prosthesis. Amputees were discharged to a hospital either to undergo a contralateral amputation or because of life threatening internal or oncological problems.

Long admission time is not only caused by poor rehabilitation progress but also by the delay in transfer to the discharge destination. Special attention needs to be given to those people who are discharged without a prosthesis. Their average admission time was 81 days. Even if a patient is not fitted due to stump, fitting or conditional problems he is trained in using walking aids, wheelchairs and making transfers and using bath and toilet facilities, in order to live independently at home. A small group of patients with mental disorders due to vascular problems were thought unsuitable for prosthetic training (Lavan, 1991; Pohjolainen and Alaranta, 1991; Pinzur *et al.*, 1992). In the group studied 2 patients could not sufficiently be trained in using a prosthesis. However, this is highly individual among patients with moderate mental disturbances (Stewart and Jain, 1993). In the authors' opinion in this group prosthetic training can always be very useful for living independently at home.

Bilateral amputees are a special group with specific problems. The average admission time was 170 days. All but four were vascular patients. The young traumatic amputees in the group were able to use their prosthesis without walking aids. Fifteen (15) patients, out of the group of 22, were successfully fitted with a prosthesis. According to Bodily and Burgess (1983) it is known that, after the first amputation, in 50% of the cases a contralateral amputation is necessary within 24 months. In the current study the mean time between unilateral and bilateral amputation was 19 months. Because of the growing group of older age amputees the bilateral lower limb amputee group will increase in the future.

## Conclusion

In this article 183 lower limb amputees referred for clinical rehabilitation were evaluated. After clinical rehabilitation 86% of all amputee patients were successfully fitted with a lower limb prosthesis.

There were three different groups:

1. the elderly vascular group where 85% of the amputees were successfully fitted with a prosthesis.
2. the small oncology group with high amputation levels where 60% were fitted with a prosthesis.
3. the trauma group with the youngest amputees, where all unilateral and bilateral amputees were fitted with a prosthesis.

The average admission time for all groups was 119 days. Eighty-five percent (85%) of the patients could be discharged home, most of them functioning adequately with a prosthesis. The main reason for some amputees not being fitted with a prosthesis were oncology complications, poor wound healing, and problems with their general condition. Stump problems and fitting difficulties were also

encountered. It was found that after clinical rehabilitation even without a prosthesis the amputee was trained in using walking aids, wheelchairs and making transfers and using toilet and bath facilities, enabling them to live independently at home. A successful prosthetic fitting offers the amputee mobility and often the possibility to return home after a lower limb amputation.

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# Functional outcome of lower-limb amputees: a prospective descriptive study in a general hospital

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**Abstract**

This paper describes functional outcome of a population of lower limb amputees five months after amputation compared with their pre-operative functional abilities and studies the relationship between potential determinants and functional outcome.

Twenty out of 26 patients who underwent a lower limb amputation between December 1993 and August 1994 in a general hospital in Amsterdam were included in the study. Their functional abilities before amputation were retrospectively assessed using an ICIDH (International Classification of Impairments, Disabilities and Handicaps)-based questionnaire. Functional outcome was assessed after amputation of the lower limb using ICIDH-based and SIP (Sickness Impact Profile)-questionnaires.

The mean SIP scores were high (referring to a low functional outcome). Disabilities were spread over the five disability fields of the ICIDH. The functional outcome of the diabetic versus the non-diabetic group was lower on the physical, activities of daily living (ADL), psychological and communicative categories of the ICIDH. In most patients, functional outcome decreased. The diabetic patients compared to the non-diabetics showed more diversity in functional outcome, compared with their preoperative functional abilities. Increasing age is significantly associated with a low functional outcome on the SIP scores. Diabetes is age-related for this sample. Co-morbidity and motivation are strongly age-related for this sample.

It was concluded that lower-limb amputees appear quite disabled in all disability categories of the ICIDH and as assessed by the SIP scores. In most patients, functional abilities decrease after lower limb amputation. Age seems to be a significant factor related to functional outcome.

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## Introduction

Several determinants seem to play an important role in the rehabilitation process of the lower limb amputee and influence functional outcome. It is important to learn more about these prognostic determinants and their relationship to functional outcome. With this knowledge, a prognosis of functional outcome can be formulated and a better policy for rehabilitation developed.

## Review of potential determinants of functional outcome

*Age*

Age seems to be a significant factor related to functional outcome of the lower limb amputee. Though it is not clear whether the determinant is age itself or whether increasing morbidity and diminished physical condition with age influence functional outcome (Steinberg *et al.*, 1985; Poulssen, 1988; Lavan, 1991).

*Smoking, sex and body mass*

In a study concerning predictive factors of functional ability after lower limb amputation, Pohjolainen and Alaranta (1991) found that smoking had an unfavourable association with walking distance, ability to walk outdoors and walking time in the male group of vascular trans-tibial (TT) amputees. Neither the sex of the amputees nor their body mass index showed any association with their walking capacity. Steinberg *et al.* (1985), Helm *et al.* (1986) and

Zijp *et al.* (1992) found no association between sex and functionality.

*Length of the stump, type of amputation and bilaterality*

In the study of Pohjolainen and Alaranta (1991) the length of the stump, in the group of TT amputees, had a significant favourable relationship with walking distance and in the group of trans-femoral (TF) amputees a mild positive association. However, they did not precisely describe their measurements of stump length and did not explain the relationship. Several authors (Pohjolainen *et al.*, 1990; Pohjolainen and Alaranta, 1991; Helm *et al.*, 1986; Narang *et al.*, 1984) have described a coincidence of a worse prognosis for functional performance with higher levels of amputation or bilaterality.

*Co-morbidity*

The presence of co-morbidity in the lower limb amputee probably will have consequences on functional performance postoperatively (Nissen and Newman, 1992; Steinberg *et al.*, 1985), however this could not always be found (Helm *et al.*, 1986; Zijp *et al.*, 1992). The presence of diabetes mellitus in one study was considered as a probable negative factor on functional outcome in the elderly amputee because of possible multi-organ damage and a higher incidence of re-amputations (Akkerman, 1983), but most studies found that the presence of diabetes mellitus made no difference in regard to functional outcome (Steinberg *et al.*, 1985; Helm *et al.*, 1986; Zijp *et al.*, 1992). In the population studied by Steinberg *et al.* (1985) more than half of the group of bilateral amputees consisted of diabetic patients. Kerstein *et al.* (1975) found that rehabilitation of bilateral lower limb amputees with diabetes mellitus was impaired severely only when visual complications or significant renal disease and hypertension were present. The presence of a heart disease is described to be a negative factor in the rehabilitation process (Kerstein *et al.*, 1975; Thornhill *et al.*, 1986), as also is chronic pulmonary disease, local stump problems (Kerstein *et al.*, 1975), stroke (Thornhill *et al.*, 1986; Pohjolainen and Alaranta, 1991) and contractures (Thornhill *et al.*, 1986). Pohjolainen and Alaranta (1991) could find no significant association between heart disease and physical function of the lower limb amputee. Three studies investigated the effect of mental problems on functional outcome. These studies all describe mental disorders as a negative factor influencing functional performance after amputation of the lower limb (Hanspal and Fisher, 1991; Kerstein *et al.*, 1975; Thornhill *et al.*, 1986).

*Motivation*

Motivation is influenced by physical and environmental factors. According to Zijp *et al.* (1992), a strong positive relation was found between motivation of the patient and the functional outcome. In their study, involving lower limb amputees in a nursing home, they looked retrospectively into the relationship between the results of rehabilitation in terms of functional performance and eight psychological and physical factors.

*Social situation*

Being employed preoperatively had favourable associations with all the ambulation functions after TF or TT amputation (Pohjolainen and Alaranta, 1991). However, employment is closely related to age and displayed similar patterns with respect to functional ambulation. Thornhill *et al.* (1986) reported on five bilateral TT amputees, who were employed at the time of second amputation and returned afterwards to work using prostheses. Independence from social provisions preoperatively (Helm *et al.*, 1986) showed favourable relationships with functional capacity and postoperative dependence after lower limb amputation. Comparison of pre- and postoperative social dependence revealed that only 6% of the patients became less dependent after operation, whereas in 36% of the patients the degree of dependence remained unchanged and in 58% it increased. No significant associations were found between functional ability and social dependence on the one hand and cohabitation on the other.

*Time lag between surgery and prosthetic fitting*

Time lag certainly is a dependent factor, related to wound healing, co-morbidity etc. The time lag between amputation and prosthetic supply displayed an unfavourable association with prosthetic usage (Pohjolainen and Alaranta, 1991). Knahr and Menschik (1991)

noticed in their group of 80 lower limb amputees a 100% social dependency in the group of patients without a prosthesis versus 17% social dependency in the group of patients with a prosthesis. The patients who were rehabilitated without a prosthesis had a diminished physical condition and duration of clinical rehabilitation was shorter.

*Stump pain or phantom pain of the stump*

Postoperative pain of the amputation stump can be disabling and can be related to co-morbidity. This pain is a dependent factor, but can be considered as an independent factor influencing functional performance. Postoperative pain in the stump or phantom pain was reported in the studies of Helm *et al.* (1986) and Pohjolainen and Alaranta (1991) and associated with decrease of functionality, particularly walking distance.

The aim of this exploratory study is firstly to describe functional outcome of a population of lower limb amputees five months after amputation compared with their functional abilities before the amputation and secondly to study the relationship between potential prognostic determinants and functional outcome.

## Patients, methods and statistics

This study concerned all 26 patients who underwent a trans-tibial amputation (TTA), knee disarticulation (KD), trans-femoral amputation (TFA) or underwent a rotation osteotomy (Rot. Ost.) between December 1993 and August 1994 in the Onze Lieve Vrouwe Gasthuis, a general hospital in Amsterdam. Six patients were excluded from the study because five (1 KD, 4 TF amputees) died and one did not want to cooperate at follow-up. The population consisted of unilateral amputees: 11 TTA, 4 patients after Rot. Ost., 4KD, 1 TFA. The median age of the population was 64 years, range 17-92 years.

The outcome variables (Table 1): Patients were interviewed for the first time during admission at the hospital for the amputation. Preoperative function (three months before surgery) was assessed by using a questionnaire, based on the International Classification of Impairments, Disabilities and Handicaps (ICIDH) (Jiwa-Boerrigter *et al.*, 1990; van den Berg and Lankhorst, 1990). Assessments are made on a 4-point scale as follows:

0. The individual is able to perform activities without difficulty on his/her own, with or without the use of aids and appliances.
1. The person is able to perform activities with some difficulty on his/her own, with or without the use of aids and appliances.
2. The person is able to perform activities with much difficulty on his/her own, or with the help of others.
3. The person cannot perform activities even with aid.

Van den Berg and Lankhorst (1990) have shown that reliability of this instrument is

Table 1. List of outcome variables, consisting of 28 items in five disability fields (Jiwa-Boerrigter *et al.*, 1990)

| | |
|---|---|
| *Physical* | 14. preparing meal |
| 1. transfer from lying, sitting | 15. household |
| 2. transfer from sitting, standing | 16. employment |
| 3. walking (inside) | 17. recreation |
| 4. traversing (outside) | 18. family role |
| 5. climbing stairs | 19. social integration |
| 6. reaching, retrieval, lifting | *Psychological* |
| 7. manual activities | 20. orientation |
| 8. endurance | 21. memory, attention |
| *ADL* | 22. behaviour, mood |
| 9. eating, drinking | 23. learning abilities |
| 10. using the lavatory | *Communicative* |
| 11. bathing | 24. understanding speech |
| 12. clothing | 25. talking |
| *Social* | 26. hearing |
| 13. transportation | 27. seeing |
| | 28. writing |

Table 2. Potential determinants

| | |
|---|---|
| age | <65 or ≥65 years |
| sex | man or woman |
| smoking | yes or no |
| type of amputation | TTA/Rot. Ost. or KD/TFA |
| co-morbidity: | |
| diabetes mellitus | yes or no |
| number of diagnoses | <2 or >2 diagnoses |
| duration preoperative admission | <39 or >39 days |
| motivation | good or impaired |
| social partner | yes or no |
| time lag amputation-prosthetic supply | <15 or ≥15 weeks |
| stump pain | yes or no |

TTA: Trans-tibial amputation
Rot. Ost.: Rotation osteotomy
KD: Knee disarticulation
TFA: Trans-femoral amputation

satisfactory.

Function outcome five months after the amputation was assessed by the same ICIDH-based questionnaire as well as a Sickness Impact Profile (SIP)-questionnaire (de Melker *et al.*, 1990; Jacobs *et al.*, 1990). The SIP questionnaire has a standardised list of 136 statements aimed at measuring changes of conduct in everyday activities due to sickness, and may be used as a measure of outcome in clinical studies.

The potential determinants of outcome are listed in Table 2: They were expressed as dichotomies. Data were obtained by analysis of the medical records.

To study the relationship between the determinants and functional outcome two and three factor Analysis of Variance was applied. The chosen level of significance was $p<0.05$.

## Results

The mean SIP total score of the population was 20.7%, the mean SIP physical score was 26.9%. Disabilities were spread over the five disability fields of the ICIDH. All patients showed problems in traversing (outside) at the time of follow-up. The functional outcome of the diabetic versus the non-diabetic group was lower on the physical, ADL, psychological and communicative categories of the ICIDH (Fig. 1). The functional outcome, as assessed by the ICIDH scores decreased in most patients: 13 patients functionally regressed, especially on the physical, ADL and social fields. The ambulatory activities were particularly compromised. Seven patients progressed functionally in all disability fields (Fig. 2). Four patients had a job in the year previous to amputation but none of these patients had returned to work five months after surgery. The diabetic patients compared to the non-diabetics demonstrated more diversity in functional abilities, compared with their functional abilities before the amputation, with relatively

Table 3. Relation of potential determinants and functional outcome

| | outcome | number of patients in subgroup | mean SIP total score (%) | mean SIP physical score (%) |
|---|---|---|---|---|
| age | ≥65 years | 9 | 26.8 | 35.7 |
| | <65 years | 11 | 15.7 | 19.7 |
| sex | man | 12 | 20.4 | 25.3 |
| | woman | 8 | 21.2 | 29.4 |
| smoking | | | | |
| (with arteriosclerosis | yes | 5 | 22.3 | 27.0 |
| or diabetes) | no | 8 | 22.6 | 31.0 |
| type of amputation | TTA/Rot. Ost. | 15 | 21.5 | 28.8 |
| | KD/TFA | 5 | 18.2 | 21.2 |
| diabetes mellitus | yes | 10 | 22.7 | 30.9 |
| | no | 10 | 18.7 | 22.9 |
| number of diagnoses | >2 | 7 | 24.4 | 31.5 |
| | <2 | 8 | 18.0 | 22.9 |
| duration preoperative | >39 days | 9 | 22.6 | 29.4 |
| admission | <39 days | 9 | 20.4 | 25.9 |
| motivation | impaired | 6 | 28.9 | 38.4 |
| | good | 14 | 17.2 | 22.0 |
| social partner | yes | 11 | 21.7 | 28.8 |
| | no | 9 | 19.6 | 24.6 |
| time lag amputation- | ≥15 weeks | 8 | 22.2 | 28.2 |
| prosthetic supply | <15 weeks | 7 | 17.7 | 23.5 |
| stump pain | yes | 8 | 20.4 | 23.7 |
| | no | 12 | 20.9 | 29.1 |

SIP: Sickness impact profile
TTA: Trans-tibial amputation
Rot. Ost.: Rotation osteotomy
KD: Knee disarticulation

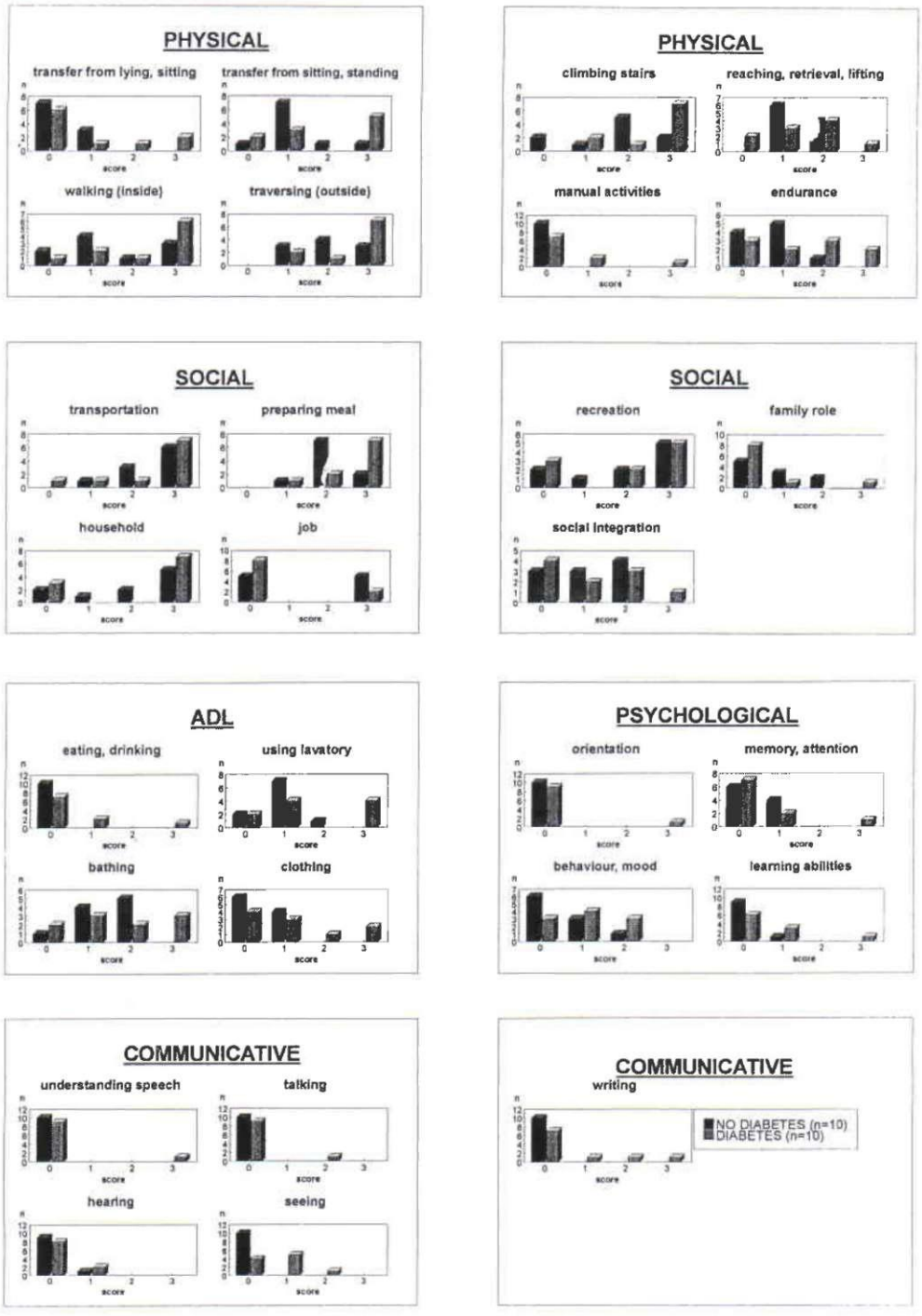



Fig 1. Functional outcome of 20 patients in five disability fields as assessed by the ICIDH scores fives months after the amputation.

Disabilities are spread over the five disability fields of the ICIDH. The diabetic patients are compared to the non-diabetics. The functional outcome of the diabetic group is lower on the physical, ADL, psychological and communicative fields.

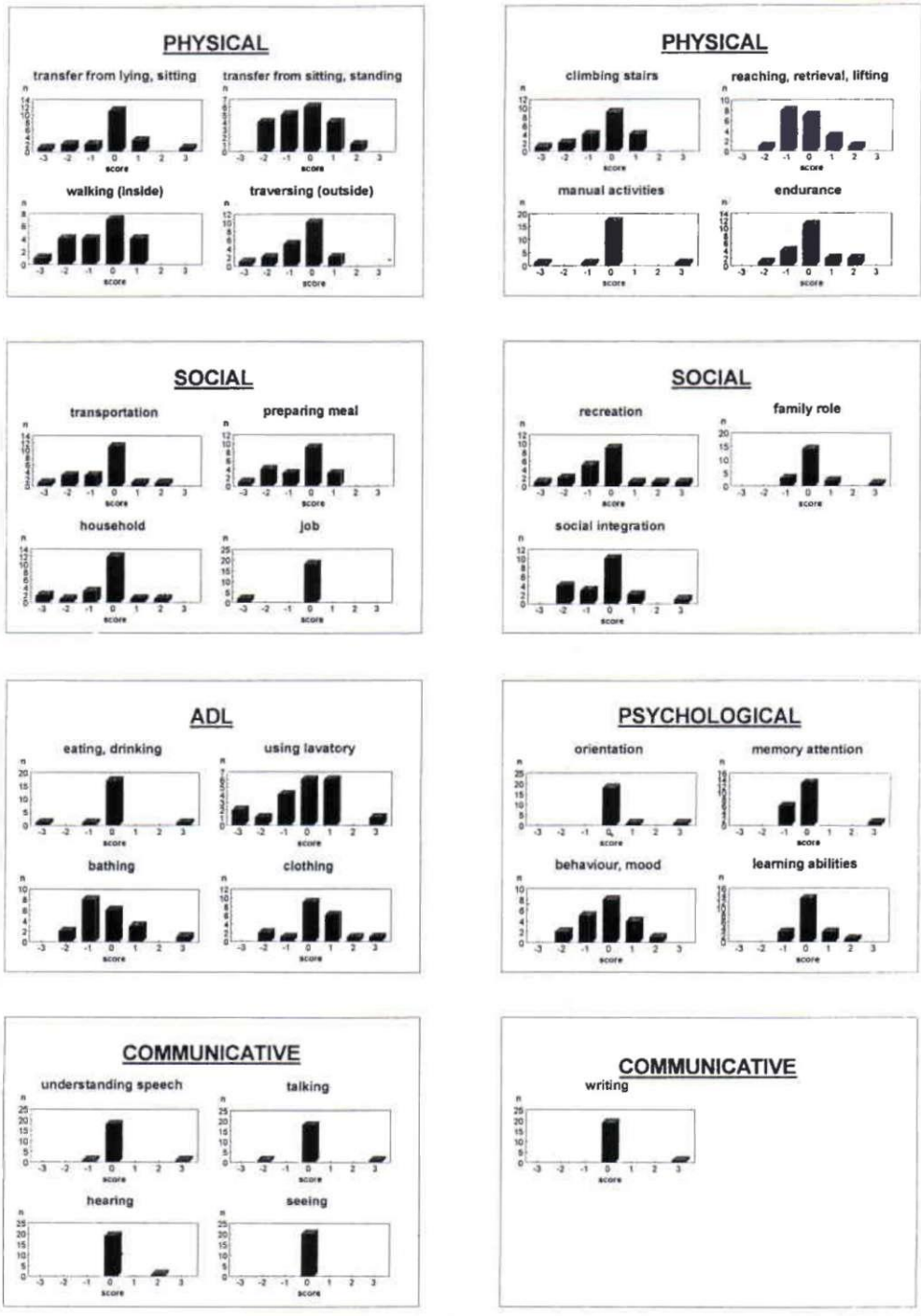



Fig 2. Functional outcome of 20 patients in five disability fields of the ICIDH five months after the amputation compared to preoperative functioning. Postoperative ICIDH acores are subtracted from the preoperative ICIDH scores. Especially on the physical, social and ADL fields a decrease in functional outcome is noted, particularly compromising the ambulatory activities. An increase in functional outcome is seen in all disability fields.

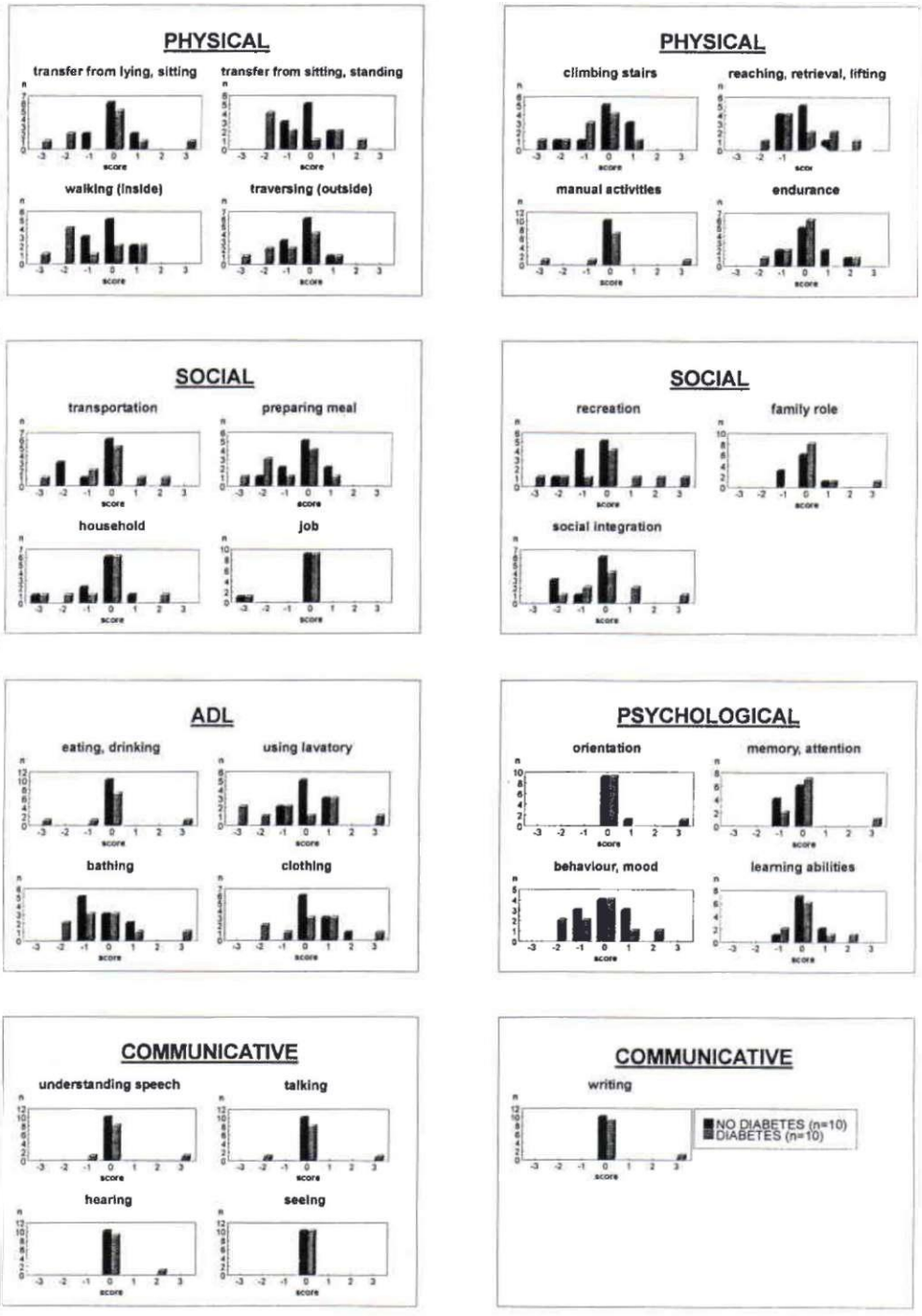
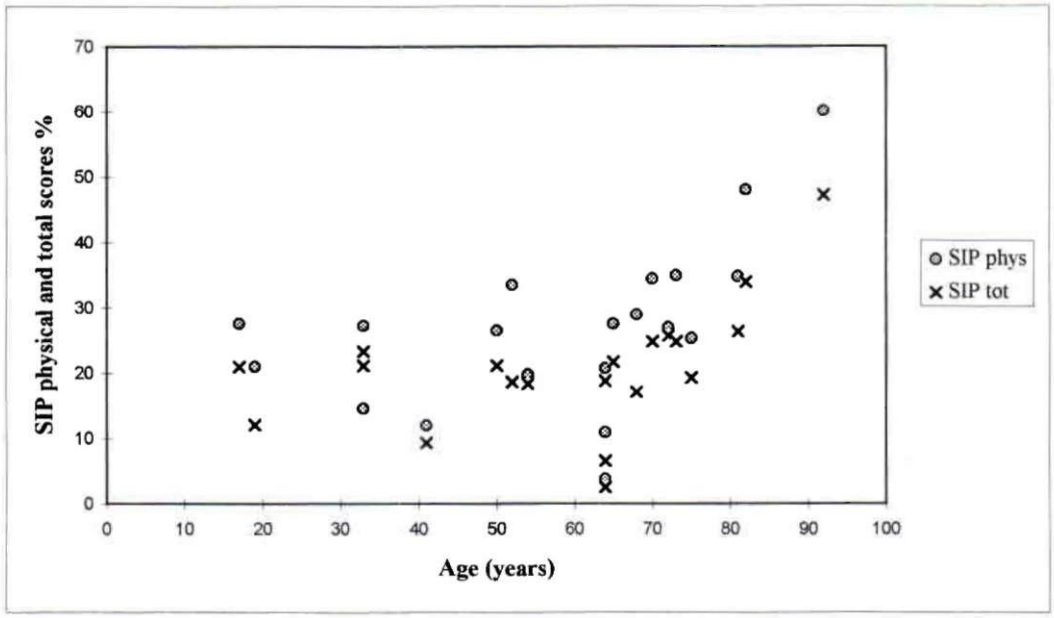



Fig. 3 . Functional outcome of 20 patients in five disability fields of the ICIDH five months after the amputation compared to preoperative functioning. Postoperative ICIDH scores are subtracted from the preoperative ICIDH scores. The diabetic patients are compared to the non-diabetics. The diabetics show more diversity in functional outcome with relatively more decline in physical and ADL fields.




SIP: Sickness Impact Profile

Fig. 4. The relation between age (years) and functional outcome as assessed by the SIP physical and total scores (%). The relation shows a zero graduated slope till the age of 65 years and an increasing slope afterwards.

more decline in the physical and ADL fields of the ICIDH (Fig. 3). No significant relationship could be found between diabetes and functional outcome on the SIP physical score and total score (Table 3). The presence of diabetes is age related for this sample. Increasing age is significantly associated with low functional outcome on the SIP physical and total scores ($p<0.05$). The relationship is shown in Figure 4. There was no correlation found between sex, smoking, type of amputation, social situation, time lag between surgery and prosthetic supply, stump pain, duration of hospital admission (during the year previous to amputation) and functional performance measured by the SIP scores. Co-morbidity and postoperative motivation are strongly age dependent for this sample. Diagnoses which were included in co-morbidity were: cardiovascular and chronic pulmonary disease, visual and hearing problems, mental disorders, problems of the musculo-skeletal system and ulcers of the lower limb. In the five postoperative months three patients underwent an ipsilateral higher level reamputation, one patient underwent an ipsilateral higher level reamputation and a contralateral lower limb amputation, one patient underwent a contralateral lower limb amputation, two patients underwent partial foot amputations. Five out of seven reamputated patients had diabetes mellitus.

## Discussion

The finding of reduced functional performance of the patients five months after amputation confirms the conclusions reported in the literature, as do most of the other data. The findings of a lower functional outcome of the diabetics and more diversity in functional outcome in the different fields of the ICIDH of the diabetic versus the non-diabetic amputees could possibly be explained by the multi-organ effects of diabetes mellitus or by an age effect. It is necessary to study a larger sample of patients to be able to find a significant difference due to the presence of diabetes mellitus in regard to functional outcome. In this sample most of the reamputations took place in the patients with diabetes mellitus. So far, no other investigators have found a correlation between diabetes mellitus and functional outcome. The relation between age and functional outcome for this sample, shown in Figure 4, shows a zero graduated slope till the

age of 65 years and an increasing slope afterwards. A larger sample of patients is necessary to map out this relationship. Though it is still not clear whether age itself is a significant factor or whether increasing morbidity, diminished physical condition and maybe diminished motivation with age influence functional outcome. Nothing can be said about the correlation between amputation level and functional abilities in this study because of the unhomogeneity of the population.

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# Statistical analysis of amputations and trends in Korea

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**Abstract**

Epidemiological research on amputees is being continued extensively world wide, but there are different epidemiologic reports from country to country. This study undertakes an epidemiologic report of the medical records of amputees in Korea which has developed very rapidly, when compared with other countries. This study included 4258 amputees who either had an amputation and/or received prosthetic training at Yonsei University College of Medicine, Severance Hospital from January 1970 to June 1994. The most common cause of amputation was trauma (66.7%), and the second most common cause was peripheral vascular disease. While amputations due to infection or trauma were the most common in the 1950's, amputations due to peripheral vascular disease have gradually increased until they now make up 23.5% of all amputations in the 1990's. Lower limb amputation, more common than upper limb amputation, accounted for 68.7% of all amputations. Multiple amputation accounted for 9.3% of all amputations, and the occurrence rate of multiple amputation was relatively higher in cases of burn injuries, train accidents, frostbite, and Buerger's disease than in cases brought about by other causes. The various amputation causes change according to the circumstances of the times, as can be seen in this study.

## Introduction

Amputation is one of the most ancient of all surgical treatments, and the history of amputations is as old as that of the human race. Early in the 16th century Ambroise Paré was the first to use ligatures to control bleeding after amputation and also designed relatively sophisticated prostheses. Later, with the development of tourniquet's aseptic surgical procedures and anaesthesia, amputation technique was developed radically by people such as Lisfranc and Syme as Tooms (1992) has described.

The increase in amputees today is due to the improvement of transportation methods, as the development of mechanical civilisation, the prolongation of life, etc. Epidemiological research on amputees is being continued extensively in many countries all over the world. After studying the cause of amputation of young men from Illinois in the USA, Lambert and Sciora (1959) found that trauma was the most common cause, accounting for 52% of all amputations, Warren and Kihn (1968) have reported that 76% of 1964 amputees who received treatment at the Veterans Administration Hospital were amputation cases caused by vascular insufficiency. Later, Hansson (1964) reported that as a result of studying 586 amputees who received amputation surgery in Sweden from 1947 to 1963, 85% of lower limb amputations were because of peripheral vascular disease, and that the rate of amputations due to peripheral vascular disease yearly. Pohjolainen and Alaranta (1988) reported that of the 750 lower limb amputees studied, 43% were because of Diabetes Mellitus. Still later, Stewart and Jain (1993) reported that the majority of amputations in Scotland were caused by peripheral vascular disease, especially arteriosclerosis.

Putting these reports together, it is possible to

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conclude that in the industrial world most amputation cases are caused by peripheral vascular disease, but in Asia, where the number of developing countries is large, there are different findings.

According to reports by Chan *et al.* (1984), a retrospective study undertaken on 1821 amputees who were treated over a period of 24 years at the Prosthetic and Orthotic Unit of Kowloon Rehabilitation Centre in Hong Kong showed that the most common cause of upper limb amputation was trauma, and that in cases of lower limb amputation the leading cause was disease. Hla Pe (1988) of Burma has made a somewhat different report, saying that trauma was the most common cause in both upper and lower limb amputations, accounting for 87% and 47% of amputations respectively. Later, Al-Turaiki and Al-Falahi (1993) performed a retrospective study on 3210 amputees who received treatment at the Riyadh Medical Rehabilitation Centre from 1977 to 1990, and found that 86.9% of upper limb amputations were due to trauma, and that 52.9% of lower limb amputations were also due to trauma, making trauma the leading cause in both cases.

As Korea differs from many other countries in having experienced the devastating 1950 Korean War; the dispatch of troops to Vietnam during the Vietnam War; rapid industrial development and improvement of transportation methods, it seems that the epidemiological statistics of Korea would be different.

Therefore, the object of this study is to investigate the basic available statistics in Korea which include the cause of amputation; the site of amputation; the distribution of age groups, and to examine the aspects that differentiate Korea from other countries, and to compare their parameters according to the changing times.

## Methods

*Subjects*

A retrospective study has been carried out of patients who have received an amputation or went through the process of prosthetic prescription, fitting and training at Yonsei University College of Medicine, Severance Hospital from January 1970 to June 1994, a span of 24 years and 6 months. Among these patients are excluded those for whom it was not possible to provide precise data because of insufficient or lost medical records, and also those who received rotation plasty because definite classification was not possible. Therefore, 4665 cases including multiple amputations from 4258 amputees, were the subject of this study.

*Data collection*

The data that was thus selected was reviewed through medical and prosthetic records. These records include the demographic factors and general characteristics including age and sex of the amputee, the amputation cause, the operation date, bilaterality, etc. The records were reviewed by medical doctors for the purposes of this study.

*Amputation age and cause*

The amputation age was determined on the basis of the last operation date, and the point of congenital limb deficiency at the time of birth. The amputation cause was classified according to the primary cause only. The large categories of amputation causes were classified into trauma, peripheral vascular disease, infection, malignancy, and congenital anomaly. The category of trauma included amputation caused by machinery, industrial injury, traffic accidents, explosions, train accidents, and burn injuries. The category of peripheral vascular disease was subdivided into diabetes, arteriosclerosis, ischaemic disease, Buerger's disease, and frostbite. The category of congenital anomaly included congenital limb deficiency and polydactyly or syndactyly. The amputation site was determined by classification provided by ICD-9 (1978), and the amputation and disarticulation of fingers and thumbs was included in the hand category. The amputation of toes and feet was included in the foot category.

*Data analysis*

All data was analysed through descriptive statistics of the SPSS statistic analysis programme.

## Results

*Amputation age and sex*

The amputation age ranged from 1 month to 83 years, and the most common amputation age group was the twenties, a time of vigorous social activity; then the teens; and then the

Table 1. Age and sex distribution of amputee

| Age (years) | Male | Female | Total | M/F Ratio* |
|---|---|---|---|---|
| 0 – 9 | 411 ( 8.1) | 243 (18.1) | 654 (15.4) | 1.7 |
| 10 – 19 | 662 (19.8) | 165 (23.3) | 827 (19.4) | 4.0 |
| 20 – 29 | 944 (28.4) | 146 (21.2) | 1090 (25.6) | 6.5 |
| 30 – 39 | 722 (21.9) | 83 (12.5) | 805 (18.9) | 8.7 |
| 40 – 49 | 373 (11.3) | 70 (10.8) | 443 (10.4) | 5.3 |
| 50 – 59 | 209 ( 6.3) | 46 ( 7.1) | 255 ( 6.0) | 4.5 |
| 60 – | 139 ( 4.2) | 45 ( 7.0) | 184 ( 4.3) | 3.1 |
| Total % | 3460 (81.3) | 798 (19.7) | 4258 (100.0) | 4.3 |

Values are given as number (%)
*M/F ratio of men to women.

thirties. These younger age groups accounted for 63.9% of all amputees. There were more male amputees than female with the overall ratio of men to women being 4.3 to 1. In the age group under 9, little difference existed between genders, but in the thirties age group male amputees vastly exceeded female amputees at a rate of 8.7 to 1 (Table 1).

*Amputation cause*

In this study, trauma was the most common amputation cause, accounting for 72.3% of all cases. The next most common cause of amputation was peripheral vascular disease. Among the acquired amputations excluding congenital anomalies, amputation due to trauma made up the largest share of all amputation cases (85.9%) in the 1950's, but have gradually decreased to 58.3% in the 1990's. Amputation due to peripheral vascular disease has gradually increased until it accounts for 23.5% of all amputations in the 1990's. Infection was the

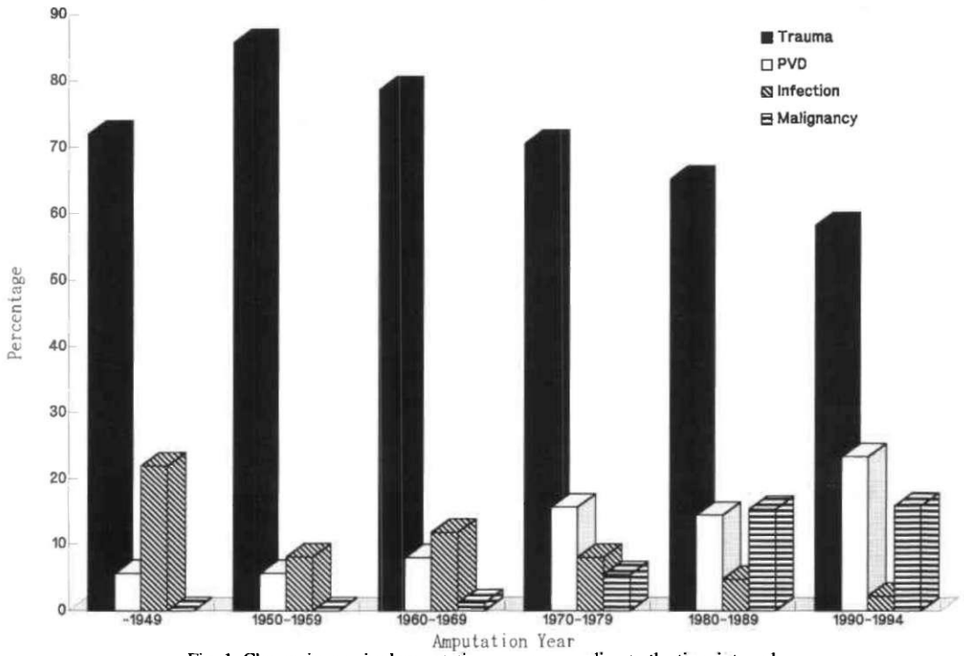


Fig. 1. Change in acquired amputation causes according to the time interval.
Amputations due to trauma have decreased gradually but cases of PVD have increased. Amputations due to congenital anomaly were excluded.

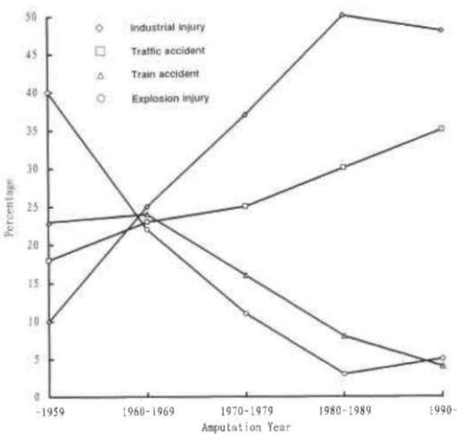


Fig. 2. Distribution of trauma subtypes according to time interval.
Amputations due to industrial injury, traffic accident have increased, but cases of explosion have abruptly decreased.

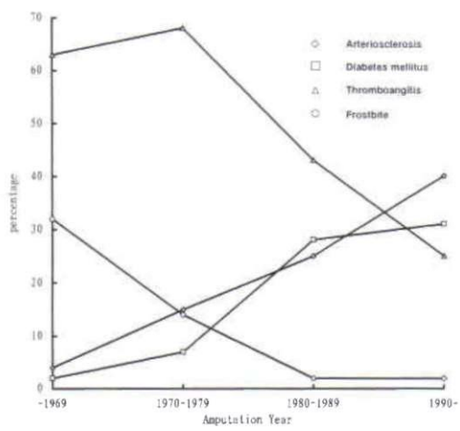


Fig. 3. Distribution of peripheral vascular disease subtypes according to time interval.
Amputations due to diabetes or arteriosclerosis have steadily increased, but cases of Buerger's disease have gradually decreased.

second leading cause of amputation in the 1950's, but has dwindled until it has become the least common cause since the 1980's (Fig. 1).

Cases of amputation caused by trauma due to industrial injury have increased since the 1950's until the rate stabilised in the 1980's (50%) and 1990's (49%); while causes due to traffic accidents have increased steadily until they made up 35% of all amputation cases in the 1990's. Amputation caused by explosives has decreased abruptly since the 1950's (Fig. 2).

Cases of amputations caused by peripheral vascular disease due to Buerger's disease have decreased gradually from the high point of 63% in the 1960's. Cases due to frostbite have also decreased gradually from the point of 32% in the 1960's, but cases due to diabetes or anteriosclerosis have gradually increased until they respectively made up 31% and 42% of all cases in the 1990's (Fig. 3).

The leading cause of amputation was trauma in all age groups except those in the sixties. In each of the younger groups from the teens to the thirties, amputation due to trauma accounted for more than 70% of all amputations. Cases of amputations caused by peripheral vascular disease tended to increase along with the increase in age. In the group over 60, peripheral vascular disease (51.5%) was a more common cause of amputation than trauma (30.4%). Of the 269 amputees who received amputation surgery due to malignancy, 72 were in their

Table 2. Distribution of amputation causes according to age

| Amputation age (years) | Trauma | PVD | Infection | Malignancy | Congenital | Total |
|---|---|---|---|---|---|---|
| 0 – 9 | 343 (52.4) | 5 ( 0.8) | 21 ( 3.2) | 14 ( 2.2) | 271 (41.4) | 654 (100.0) |
| 10 – 19 | 643 (77.8) | 23 ( 2.8) | 63 ( 7.6) | 72 ( 8.7) | 26 ( 3.1) | 827 (100.0) |
| 20 – 29 | 859 (78.7) | 78 ( 7.2) | 91 ( 8.3) | 40 ( 3.7) | 22 ( 2.1) | 1090 (100.0) |
| 30 – 39 | 571 (70.9) | 134 (16.6) | 53 ( 6.6) | 44 ( 5.5) | 3 ( 0.4) | 805 (100.0) |
| 40 – 49 | 253 (57.1) | 98 (22.1) | 47 (10.6) | 43 ( 9.7) | 2 ( 0.5) | 443 (100.0) |
| 50 – 59 | 117 (45.9) | 72 (28.2) | 29 (11.4) | 36 (14.1) | 1 ( 0.4) | 255 (100.0) |
| 60 – | 56 (30.4) | 94 (51.5) | 13 ( 7.1) | 20 (10.9) | 1 ( 0.6) | 184 (100.0) |
| Total | 2842 (66.7) | 504 (11.9) | 317 ( 7.4) | 269 ( 6.3) | 326 ( 7.7) | 4258 (100.0) |

Values are given as number (%)
PVD: peripheral vascular disease.

Table 3. Distribution of amputation causes according to amputation level

| Amputation age (years) | Trauma | PVD | Infection | Malignancy | Congenital | Total |
|---|---|---|---|---|---|---|
| Shoulder | 22 ( 0.7) | 0 ( 0.0) | 1 ( 0.3) | 15 ( 5.6) | 0 ( 0.0) | 38 ( 0.8) |
| TH | 343 (11.1) | 3 ( 0.5) | 5 ( 1.5) | 8 ( 3.0) | 0 ( 0.0) | 359 ( 7.7) |
| TR | 439 (14.2) | 12 ( 1.9) | 13 ( 3.8) | 3 ( 1.1) | 11 ( 3.2) | 478 (10.2) |
| Hand | 341 (11.0) | 24 ( 3.9) | 8 ( 2.3) | 12 ( 4.4) | 199 (58.7) | 584 (12.5) |
| Hip | 45 ( 1.5) | 3 ( 0.5) | 8 ( 2.3) | 61 (22.6) | 1 ( 0.3) | 118 ( 2.5) |
| TF | 823 (26.6) | 151 (24.3) | 150 (43.8) | 137 (50.7) | 9 ( 2.7) | 1270 (27.3) |
| TT | 1023 (33.1) | 357 (57.5) | 151 (44.2) | 24 ( 8.9) | 15 ( 4.4) | 1570 (33.7) |
| Foot | 57 ( 1.8) | 71 (11.4) | 6 ( 1.8) | 10 ( 3.7) | 104 (30.7) | 248 ( 5.3) |
| Total | 3093 (100.0) | 621 (100.0) | 342 (100.0) | 270 (100.0) | 339 (100.0) | 4665 (100.0) |

Values are given as number (%)
PVD: peripheral vascular disease; TH: trans-humeral and elbow disarticulation; TR: trans-radial and wrist disarticulation; Hand: hand and finger; TF: trans-femoral and knee disarticulation; TT: trans-tibial and Syme; Foot: foot and toe.

teens, making this group the majority in this category. Most of the cases due to congenital anomaly received treatment or amputation before the age of 9 (Table 2).

*Amputation site*

With the exception of hip disarticulation and foot amputation; for all amputation levels, trauma was the most common cause of amputation. Malignancy was the most common cause of hip disarticulation (61 out of 118 cases). Peripheral vascular disease was the leading cause of amputation in cases of trans-tibial amputation, and, in cases due to congenital deformity, hand or foot amputation was the most common procedure performed (Table 3).

There were more cases of lower limb amputation (68.7%) than upper limb amputation. Peripheral vascular disease, infection, and malignancy occurred more often in the lower limbs than in the upper limbs, and only congenital deformity was more common in the upper limbs (Table 4).

*Multiple amputation*

Multiple amputation accounted for 9.3% of all amputations. The occurrence rate of multiple amputation was relatively higher in cases caused by burn injuries, train accidents, frostbite, or Buerger's disease than in cases due to other causes (Table 5).

## Discussion

Statistics on amputations differ because of factors such as the population of the subject group, the method of research, and the social conditions of the country. This is a hospital based, retrospective study in Korea, which has experienced dramatic social changes including rapid industrial development literally from the ashes of the Korean War in 1950.

In this study the younger age groups from the teens to the thirties accounted for 63.9%, thus making up the leading amputation age group. This grouping is different from other reports. Kerstein *et al.* (1974) reported that the average age of amputation was 56.8, and Warren and Kihn (1968) also reported that the majority

Table 4. Comparison of upper and lower limb amputation causes

| | Trauma | PVD | Infection | Malignancy | Congenital | Total |
|---|---|---|---|---|---|---|
| ULA | 1145 (37.0) | 39 ( 6.3) | 27 ( 7.9) | 38 (14.1) | 210 (61.9) | 1459 (31.3) |
| LLA | 1948 (63.0) | 582 (93.7) | 315 (92.1) | 232 (85.9) | 129 (38.1) | 3206 (68.7) |
| Total | 3093 (100.0) | 621 (100.0) | 342 (100.0) | 270 (100.0) | 339 (100.0) | 4665 (100.0) |

Values are given as number (%)
PVD: peripheral vascular disease; ULA: upper limb amputation; LLA: lower limb amputation

Table 5. Comparison of multiple amputation rates according to causes

| Cause | No. of MA/AA | Rate (%) |
|---|---|---|
| Trauma | 287/2842 | 10.1 |
| Burn | 37/ 149 | 24.8 |
| Train accidents | 112/ 447 | 25.1 |
| Others | 138/2246 | 6.1 |
| PVD | 94/ 504 | 18.7 |
| Frostbite | 32/ 67 | 47.8 |
| Buerger's disease | 52/ 252 | 20.6 |
| Other | 10/ 185 | 5.4 |
| Infection | 6/ 317 | 1.9 |
| Malignancy | 3/ 269 | 1.1 |
| Congenital | 6/ 326 | 1.8 |
| Total | 396/4258 | 9.3 |

Values are given as number (%)
No. of MA/AA: number of multiple amputation per all amputation; PVD: peripheral vascular disease.

were over 60. On the other hand, Seiler and Richardson (1986) who only investigated traumatic amputees reported the average amputation age as 31.5, and Al-Turaiki and Al-Falahi (1993) reported that their hospital based study showed the younger age group took the majority. The age of the amputee in this report is younger than in others due to the fact that most amputations were due to trauma, which occurs frequently to younger people who lead active lives. There is also, the fact that the age group from the teens to the thirties makes up about 60% of the whole population of Korea.

Reports on the causes of amputation differs from country to country. Stewart and Jain (1993) reported that peripheral vascular disease was the leading cause (80%), and that amongst those cases arteriosclerosis accounted for 60%. Kerstein *et al.* (1974) reported as the results of an exclusive study on lower limb amputees that 85% of the cases were due to peripheral vascular disease. But Al-Turaiki and Al-Falahi (1993) reported in their study in Saudi Arabia that trauma accounted for 60%, and disease for 30% of all cases. Hla Pe (1988) in Burma reported that the leading cause of amputation is trauma. In this study amputation caused by trauma accounted for 66.7% of all cases, while peripheral vascular disease accounted for 11.9% and infection for 7.4%, making trauma undoubtedly the primary cause of amputation. The leading causes of amputation in different countries is influenced by the degree of industrialisation, the transportation system, and the medical care available in each country. In the case of Korea, it seems that trauma is the leading cause of amputation because of factors such as the 1950 Korean War; explosion injuries from the Vietnam War where Korea sent troops in the 1960's; the rapid development of an industrial and mechanical civilisation since the 1960's, and the increase of the volume of traffic, along with increased speeds. This is similar to Banerjee's report (1982), which said that trauma, infection, and Buerger's disease were the three leading causes of amputation in developing countries, and except in times of war, peripheral vascular diseases such as diabetes and arteriosclerosis were the leading cause of amputation in developed countries.

The leading cause of amputation has changed with time. Amputation due to trauma decreased to 58.5% in the 1990's from 85.9% in the 1950's, and amputation due to infection has decreased until it has become the least common cause of amputation. This appears to be a result of the improvement in treatment techniques brought about by the development of medical science and antibiotics. But cases of amputation due to peripheral vascular disease increased greatly from 5.7% in the 1950's to 23.5% in the 1990's. This fact may be related to a surprising gradual increase in the intake of a high protein, high fat diet, and the fact that the population of senior citizens over the age of 65 is increasing 5.4% every year. If one looks at reports on the change of peripheral vascular disease with time. Hansson (1964) says that while 2% of all people using prostheses in 1926 received amputation surgery because of peripheral vascular disease, that percentage increased to 57% in 1955. Hierton and James (1973) reported the rate of amputation due to peripheral vascular disease increased from 69% in 1947 to 73% in 1957, and to 93% in 1967-1969, and that this was because of the increase in the number of advanced age citizens in the population. But Buchanan and Mandel (1986) reported as the result of a study on amputation cases performed through institutions offering prosthetic services throughout Canada that the rate of amputation due to trauma increased from 33.6% in 1960 to 43.1% in 1980. They gave the contradictory opinion that this could be because amputation

cases due to disease decreased thanks to the improvement of treatment methods of internal diseases while the rate of amputations due to trauma increased correlatively.

Looking at the cause of amputation according to age, Kerstein *et al.* (1974) reported that peripheral vascular disease occurred mainly in the senior age group and that it especially occurred frequently in the fifties age group. These reports are consistent with this study, and show that cases where amputation was performed because of accidents occur relatively more often in the younger age groups while cases due to peripheral disease are more frequent than cases due to accidents in the sixties and older age group and that the occurrence rate of these peripheral vascular disease related cases increases with age. Cases of amputation due to malignancy were most common in the teen age group compared to the other age groups. This seems to be because of the speciality of osteogenic sarcoma which is a malignancy that occurs mainly in the teens and twenties age groups as Tebbi *et al.* (1985) have reported.

Lower limb amputation accounts for 68.7%, which is a larger percentage than that of upper limb amputations. This is a similar result to the report of Tooms (1972) saying that upper limb amputations make up about 15-20% of all amputations, and the report of Al-Turaiki and Al-Falahi (1993) saying that lower limb amputations account for about 65%

Looking at the various sites of amputation according to the cause, cases due to accidents or peripheral vascular disease were more common in lower limbs than in upper limbs, and among those lower limbs amputations the most frequent site of amputations was trans-tibial amputation including Syme amputation. In amputation cases due to malignancy, however, trans-femoral amputation occurred most frequently. This is because the distal portion of limbs are generally more likely to be injured, and surgeons tend to amputate as distally as possible to enhance functional activity.

As for the frequency of multiple amputations, Hansson (1964) reported 16.9%, Stewart and Jain (1993) 18%, Al-Turaiki and Al-Farahi (1993) approximately 5%, and Kerstein *et al.* (1974) 3.4%. The rate of multiple amputation in this study is 9.3%, which is similar to other reports. In cases where the cause of amputation was electric burns, train accidents, frostbite, or Buerger's disease, the occurrence rate of multiple amputation was relatively high. This is because in cases of electric burns, there is an entrance and exit area as in gunshot wounds as Baxter (1970) has reported. In train accident cases, the reason seems to be that train accidents tend to be more severe than other traffic accidents. As Buerger's disease is not a localised disease involving only one limb, the recurrence rate is high as Ohta and Schionoya (1988) have reported. Therefore, it seems that in these cases there would be a higher multiple amputation relative risk compared with other causes. La Borde and Meier (1978) reported that the multiple amputation rate of electrical burn injury patients is 50%, and Kegel *et al.* (1978) have reported that most multiple amputations were caused by peripheral vascular disease.

## Conclusion

Putting the results of this study together, we arrived at statistics similar to those of other Asian countries. Unlike developed countries, trauma was the leading cause at present. However, the amputation cause according to the circumstances of the times changed dramatically as can be seen in this study as the amputation cases due to peripheral vascular disease have increased gradually and the cases due to trauma have decreased gradually.

It is hoped that this study on amputees in Korea will help in the efforts to provide more comprehensive rehabilitation treatment by predicting the course of future amputation patterns and being used as basic data in setting up plans in the developing countries.

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# Stump length as related to atrophy and strength of the thigh muscles in trans-tibial amputees

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**Abstract**

Stump length and the thigh muscles strength of the amputated limb are among the major factors influencing outcome of prosthetic rehabilitation of trans-tibial amputees. In the present study the authors evaluated and compared the strength of quadriceps and hamstrings muscles of both limbs in trans-tibial amputees, as measured by means of an electrical dynamometer. The obtained results showed that the thigh muscles of the sound limbs are significantly stronger than those of the amputated limbs ($p<0.01$). The results obtained for amputees with shorter stumps were compared to those with longer stumps. In the group of amputees (n=9) with a stump shorter than 15.1 cm, values of peak torque (in isokinetic contraction) and maximal average torque (in isometric contraction) were significantly ($p<0.5$) weaker when compared to those (n=9) with a stump longer than 15.1 cm. The results obtained for amputees with a higher rate of thigh muscle atrophy were compared to those with lesser atrophy. In the group of amputees where muscle atrophy was accompanied by decrease in thigh girth of over 5.9 cm, muscles strength did not significantly decrease ($p<0.5$) as compared to amputees where thigh girth decrease was less than 5.9 cm. It is concluded that atrophy of the thigh muscles of trans-tibial amputees is accompanied with a significant decrease in strength. In amputees with a short stump, the short lever action provided by the stump interferes with the ability of the thigh muscles to control the prosthesis efficiently during daily activities such as standing and walking.

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## Introduction

A short stump might interfere with success in prosthetic rehabilitation of trans-tibial amputees. An adequately long stump provides the amputee with a good proprioceptive feedback (Guerts and Mulder, 1992) resulting from both a large contact surface and good stability of the stump-socket unit. These factors enable good standing and walking in these patients (Isakov *et al.*, 1985; Isakov *et al.*, 1994; Scliktar *et al.*, 1980). When the stump of a trans-tibial amputee is short and the transverse and longitudinal dimensions are almost similar, the stump acquires a round shape and may become unstable inside the prosthetic socket (Seliktar *et al.*, 1980; Nissan, 1977). As a result, instability of the stump-prosthesis complex creates shear forces with resulting pain and/or blisters or friction sores which prevent prosthesis usage for a long period of time. Researchers have studied the forces acting on a short trans-tibial stump (Sanders *et al.*, 1992; Lilja *et al.*, 1993) and various prosthetic solutions have been suggested for improving stability of the stump-socket unit and the quality of standing and walking (Isakov *et al.*, 1992; Pritham, 1979).

It is suggested that the stump of trans-tibial amputees is less active in the daily functions of standing and walking. In fact, atrophy of the thigh muscles of the amputated limb is often observed among such amputees, a finding evident from quadriceps muscle biopsies (Renstrom *et al.*, 1983). Evaluation of standing balance activity of both limbs in trans-tibial amputees showed that the foot-ground reactive forces generated by the amputated limb are

Table 1. Variables of trans-tibial amputees subjects (TT: tibial tuberosity, PTS: patellar tendon supracondylar, PTB: patellar tendon bearing)

| | | | |
|---|---|---|---|
| n=18 | | | |
| Sex | (F/M) | 6/12 | |
| Age | (yr) | 45.7±14.7 | (27-74) |
| **Amputation** | | | |
| - side | (R/L) | 6/12 | |
| - duration | (yr) | 13.4±14.4 | (1-46) |
| **Stump** | | | |
| - length | (cm) | 15.1±3.2 | (8.5-21) |
| - thighs girth diff.: | (cm) | | |
| | - 18 cm above TT | 5.9±2.5 | (2-12.5) |
| | - 30 cm above TT | 4.0±2.9 | (0-10.5) |
| **Walking** | | | |
| - distance | (km) | 2.5±1.4 | (0.5-10) |
| - prosthesis | | | |
| | PTS | 4 | |
| | PTB+belt | 12 | |
| | PTB+corset | 2 | |
| - aids | none | 17 | |
| | cane/crutch | 3 | |

smaller (Isakov *et al.*, 1992; Geurts and Mulder, 1992). Analysis of gait in trans-tibial amputees showed a shorter body weight bearing stance phase on the amputated limb as well as asymmetry in other gait phases (Baker and Hewison, 1990). It can therefore be assumed that reduced involvement and activation of the amputated limb muscles in activities such as standing and walking results in disuse atrophy of the thigh muscles of the affected limb. The present study aimed at evaluating whether the stump length and thigh muscles atrophy are related to the strength of the thigh muscles in trans-tibial amputees.

## Subjects

Eighteen volunteers, 6 female and 12 male, with trans-tibial amputation were assessed (Table 1). Their average age was 45.7±14.7 years (range, 27 to 74 years). The mean time from the amputation to the present study was 13.4±14.4 years (range, 1 to 46 years). Causes for amputation were; traffic injury 6, work accident 5, war injury 1, diabetes mellitus 2, peripheral arterial disease 1, tumour 1, infection 2.

Subjects' mean stump length was 15.1±3.2 cm (range, 8.5 to 21 cm). The thigh girth at 18 cm above the tibial tuberosity of the amputated limb was smaller compared with the contralateral limb (5.9±2.5; range: 2-12.5 cm). At 30 cm above the tibial tuberosity the difference was 4.0±2.9 cm (range: 0-10.5 cm). All subjects were fitted with a patellar tendon-bearing prosthesis. They were using their prostheses for the whole day and were able to walk continuously a mean distance of 2.5±1.4 km. No subject had any knee joint contracture or complaints of knee or stump pain during the test.

## Methods

### *Measurements*

In each subject, measurements of the thigh girth were taken at an equal distance from the tibial tuberosities of both legs, proximally 18 and 30 cm. Measurements were taken by one of the team members using a standard tape measure, a method found reliable for measuring circumference of the lower limb (Renstrom *et al.*, 1983). Measurements were performed in the lying position with knees extended and thigh muscles relaxed.

Stump length was measured while the knee was in 90 degrees of flexion. The upper reference point was the medial femoro-tibial inter-condylar line and the lower point the stump tip.

### *Instrumentation*

The isokinetic concentric and eccentric and isometric muscle torques of the knee extensors

and flexors were measured on-line with Biodex Model B-2000 (Biodex, NY, USA) (Wilk and Johnson, 1988; Thompson *et al.*, 1993). A special lever was constructed to meet with the special needs of the tested stump. For each subject the lever arm was adjusted to rest at the same distance from the lower pole of the patella in both limbs. In order to eliminate the effect of gravity, the torque due to the mass of the limb was determined and then added to measured torque values when working against gravity and subtracted from the torque values when working with gravity. The dynamometer was calibrated according to the manufacturer's recommendations once a week.

*Measured parameters:* Muscular efficiency was evaluated by measuring the quadriceps and hamstrings isokinetic concentric and eccentric and isometric strength. Peak torque (Nm) is the highest value of torque developed throughout the range of motion curve. Maximal average torque (Nm) is the greatest average torque produced for a repetition within a set.

*Procedure*

*Positioning:* The patients were seated with hip joints angle of 90-100 degrees. Stabilisation straps were placed across the trunk, around the waist, and mid-thigh of the limb to be tested. The anatomical axis of the knee joint was visually aligned with the axis of rotation of the dynamometer.

*Learning phase:* Prior to each test, subjects were instructed and allowed to try the dynamometer. The differences between concentric, eccentric and isometric were explained. The subjects then performed submaximal contractions in order to be acquainted with the exercise.

*Test sequence:* The test procedure consisted of five consecutive isokinetic concentric contractions for the quadriceps and hamstrings muscle groups, followed by five eccentric contractions with five minutes rest. The constant angular velocity was established at 60 degrees per second. Following 5 minutes rest, each subject performed five consecutive isometric contractions of 5 seconds each. The quadriceps contraction was evalued at 45 degrees of flexion and the hamstrings contraction at 60 degrees of flexion. Subjects were asked to perform their maximal effort and verbal encouragement was provided to each subject throughout the test session. Statistical analysis of the significance of the results was performed using the paired t-test and the level of significance was set at $p<0.05$.

## Results

The means and standard deviations of values (Nm) obtained for the quadriceps and hamstrings of both limbs are detailed in Table 2. Results relate to the three different strength measurements during isokinetic concentric, isokinetic eccentric (peak torque) and isometric contractions (maximal average torque). Values obtained in the muscles of the amputated limb were significantly smaller compared with the sound limb muscles ($p<0.01$). The test subjects were divided into two groups. The first group of nine amputees included those with stump length less than 15.1 cm, the second group of nine amputees had a stump length of more than 15.1 cm. Values of torques obtained in these two groups are detailed in Table 3. In the group with the shorter stump, quadriceps (eccentric contraction) and hamstrings (concentric contraction) strength was significantly smaller when compared to the group with a longer stump ($p<0.05$). All other values in the group

Table 2. Means and standard deviations (Nm) of muscles peak torque (isokinetic contraction) and maximal average torque (isometric contraction)

| Muscle | Type of muscle contraction | Amputated limb | Sound limb | p |
|---|---|---|---|---|
| **Quadriceps** | Isokinetic concentric | 40.4±20.5 | 76.7±31.0 | .001 |
| | Isokinetic eccentric | 112.0±47.2 | 171.2±45.4 | .002 |
| | Isometric | 46.0±26.4 | 93.0±34.0 | .001 |
| **Hamstrings** | Isokinetic concentric | 29.9±20.0 | 74.6±34.8 | .001 |
| | Isokinetic eccentric | 64.5±37.3 | 128.2±45.1 | .001 |
| | Isometric | 30.3±20.4 | 52.6±24.3 | .007 |

Table 3. Values of muscles peak torque (isokinetic contraction) and maximal average torque (isometric contraction). Means and standard deviations (Nm) are related to stump length

| Muscle | Type of muscle contraction | **Stump length:** <15.1 cm (n=9) | <15.1 cm (n=9) | p |
|---|---|---|---|---|
| **Quadriceps** | Isokinetic concentric | 32.6±16.6 | 48.3±21.9 | .113 |
| | Isokinetic eccentric | 87.0±27.6 | 137.1±50.6 | .016 |
| | Isometric | 35.4±17.3 | 60.4±31.7 | .112 |
| **Hamstrings** | Isokinetic concentric | 19.1±6.0 | 40.6±23.6 | .038 |
| | Isokinetic eccentric | 47.8±10.5 | 81.3±47 | .075 |
| | Isometric | 22.1±12.6 | 38.3±24.2 | .087 |

with the shorter stump were also smaller however less significantly ($p<0.3$).

Atrophy of the amputated thigh muscles as related to thigh girth differences were considered also. The subjects were divided into two groups according to girth differences obtained 18 cm above the tibial tuberosity. In the first group of nine amputees, thigh girth mean differences were greater than 5.9 cm and in the second group thigh girth mean differences were less than 5.9 cm. The mean of muscle torque in both groups were calculated and compared (Table 4) showing the differences in muscles strength between these groups to be insignificant ($p<0.5$).

## Discussion

Although amputation surgery is a constantly improving process, there are still cases where it is not possible to construct a sufficiently long trans-tibial stump. In the present study, the amputees' average length of the stump was 15.1 cm (range: 8.5-21 cm) while others report an average length of 16 cm with a range between 10 and 25 cm (Persson and Liedberg, 1983). Stump length is the major factor determining stability of the BK stump inside the prosthetic socket and consequently the quality of standing and walking (Seliktar *et al.*, 1980; Nissan, 1977; Isakov *et al.*, 1992). Other important factors which influence the rehabilitation outcome of amputees are an optimal prosthesis fitting and strength of the quadriceps and hamstrings muscles controlling the knee of the affected limb (Klingenstierna *et al.*, 1990).

The authors evaluated the isokinetic and isometric strength of the amputated limb thigh muscles by means of a dynamometer. The advantages of this method when compared to manual muscle testing (Nicholas *et al.*, 1978; Iddlings *et al.*, 1961) are that it provides an objectively reproducible and reliable test (Steiner *et al.*, 1993; Wilke and Johnson, 1988; Thompson *et al.*, 1993). The obtained results for peak torque in isokinetic concentric and eccentric and for maximal average torque in isometric contraction were significantly smaller in the amputated limbs when compared to the sound limbs. The strength of thigh muscles of amputees with a short stump was compared with those with a longer stump. The obtained results indicate clearly a decrease in strength of the amputated limb thigh muscles in trans-tibial amputees with a short stump as evaluated under the described conditions. It is therefore assumed that the inefficient lever action provided by the

Table 4. Values of muscles peak torque (isokinetic contraction) and maximal average torque (isometric contraction). Means and standard deviations (Nm) are related to thigh girth. All comparisons are insignificant (p>.5)

| Muscle | Type of muscle contraction | **Thighs girth differences:** <5.9 cm (n=9) | <5.9 cm (n=9) |
|---|---|---|---|
| **Quadriceps** | Isokinetic concentric | 35.9±13.8 | 44.9±25.6 |
| | Isokinetic eccentric | 108.5±47.9 | 115.6±49.1 |
| | Isometric | 45.7±18.7 | 50.1±36.0 |
| **Hamstrings** | Isokinetic concentric | 30.0±20.8 | 29.8±20.5 |
| | Isokinetic eccentric | 62.1±27.3 | 66.9±46.8 |
| | Isometric | 24.1±11.6 | 35.8±26.3 |

short trans-tibial stump, compromises the performance ability and level of the knee muscles activity during prosthesis usage.

Thigh muscle atrophy was determined by measurements of thigh girth. The measuring level was chosen where differences between thigh girth were greater, 18 cm proximal to the tibial tuberosity. Two groups of amputees were compared; those where thigh girth differences were greater than 5.9 cm and those with a difference smaller than 5.9 cm. Comparing values of muscles strength measured in these two groups, a general decrease in strength was noticed among those with a greater thigh muscles atrophy nevertheless, differences were not significant.

In conclusion, although most trans-tibial amputees manage their activities of daily living and conduct a fairly active life, strength of the amputated limb thigh muscles was found significantly reduced, especially in amputees with a short stump. In so far as the ultimate goal in the rehabilitation of amputees is to return the patient to an acceptable level of function, it is recommended that the amputee should be trained and encouraged in self strengthening exercises for the amputated limb thigh muscle. Stronger muscles will improve standing balance and quality of gait, especially among those with a short stump.

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# Use of an instrumented treadmill for real-time gait symmetry evaluation and feedback in normal and trans-tibial amputee subjects

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**Abstract**

The purpose of this research was to evaluate a newly developed system for assessing and providing feedback of gait symmetry information in real time to subjects walking on a motorised treadmill (the CCF Treadmill). The advantages of the system are that it allows the rapid collection and comparison of temporal and kinetic parameters of gait for multiple successive strides, at a constant known speed, without forcing subjects to target their footsteps. Gait asymmetries of six normal (mean age 42.7 years) and six unilateral trans-tibial amputee subjects (mean age 41.7, and average 6.0 years using a prosthesis) were quantified. The amputee group was the re-evaluated after receiving five minutes of training with each of three different types of real-time visual feedback (RTVF). Asymmetries in the measured parameters before feedback were 4.6 times greater in the amputee population than in the normal group, and were consistent with the finding of previous authors. Significant decreases in gait asymmetry were demonstrated for all forms of feedback after amputees received feedback training. Results, however, indicate that gait asymmetries for different variables are not necessarily related, and that more work needs to be done to identify those variables for which attaining a more symmetrical gait pattern is most beneficial. Further work also needs to be done to determine the long term effects of such RTVF training. The CCF Treadmill and RTVF were shown to be potentially useful tools both for defining rehabilitation targets and for quantifying patients' progress towards those goals.

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**Background and significance**

Nearly sixty thousand major lower limb amputations are performed in the United States each year (DHHS Publication No. PHS 92-1774, 1992) and more than half of those are trans-tibial (TT) amputations (Wilson, 1989). The majority of amputees are elderly patients with peripheral vascular disease (PVD), usually related to diabetes mellitus (DeLuccia *et al.*, 1992; Harris *et al.*, 1991), and have poor general prognosis. As many as 45% of these patients are unable to master the use of a prosthesis (Moore *et al.*, 1989) and often become candidates for long term institutional care. The cost of caring for elderly amputee patients in the United States is expected to grow to as much as three billion dollars a year by the year 2000 (Cherner, 1993). Developing ways of improving rehabilitation outcomes could go a long way towards reducing these costs and providing a better quality of life for these patients.

In human gait, symmetry between left and right limbs can be measured for anthropometric, temporal, kinetic, kinematic, or electromyographic (EMG) data. Several authors have reported small, but consistent asymmetries in the timing (Rosenrot *et al.*, 1980; Hirokawa, 1989; Herzog et al., 1989), ground reaction force profiles (Herzog *et al.*, 1989), and kinematics (Gunderson *et al.*, 1989) of normal subjects. Herzog *et al.*, (1989) defined asymmetry as the ratio of the difference between the left and right values to the average

of the two values times 100%, where perfect symmetry was given by SI = 0%. Asymmetries for 12 variables, extracted from the vertical force curves of 62 men and women were all within ±4%, with standard deviations ranging from ±2.0% to ±41.3% (Herzog *et al.*, 1989). Variables derived form the anteroposterior and mediolateral force curves exhibited greater asymmetries. These data suggest that there are slight asymmetries inherent even in normal gait. However, the magnitudes of asymmetry reported differed for different variables, suggesting that asymmetry is not a universal quality of gait, but is dependent upon the particular variable being measured.

Symmetry is an issue in the gait of amputees because of the unnatural asymmetry imposed on the biomechanical system by the prosthesis (Winter and Sienko, 1988). The most prominent asymmetries found in amputee gait have involved shortened stance times (Breakey, 1976; Cheung *et al.*, 1983; Skinner and Effeney, 1985; Seliktar and Mizrahi, 1986; Baker and Hewison, 1990) and decreased ground reaction forces (Skinner and Effeney, 1985; Seliktar and Mizrahi, 1986; Baker and Hewison, 1990) for the prosthetic limb compared to the natural limb. Most studies in the literature have focused on the qualitative description of gait asymmetries (Skinnner and Effeney, 1985), or quantitative measures based on raw differences (Breakey, 1976; Skinner and Effeney, 1985; Cheung *et al.*, 1983; Baker and Hewison, 1990), or ratios (Seliktar and Mizrahi, 1986) of values recorded for each limb. Lack of plantar flexion, and normal ankle motion has been described as the primary cause of most amputee gait deviations, including asymmetrical gait timing, knee joint motions, and increased muscle activities in both amputated and contralateral limbs (Breakey, 1976; Winter and Sienko, 1988). Loss of normal neuromuscular control and proprioceptive feedback functions have been cited as the major causes of the increased variability in gait timing between normal and amputee subjects (Zahedi *et al.*, 1987).

Cheung *et al.* (1983), reported that raw differences in total support times for four TT amputee patients decreased from 5.7% to 3.5% of the total stride time after six weeks of gait training. Similar results were reported by Baker and Hewison (1990) for asymmetries in single support times of twenty unilateral amputee subjects, indicating that these inherent gait asymmetries can be reduced with training. Bach *et al.* (1994) tested a computer simulation which adjusted inertial loading and mass distributions in the prostheses of five trans-femoral amputee patients in order to maximise swing phase symmetry. Significantly greater swing phase symmetry, reduced oxygen consumption, and increased subjective ratings were found for subjects wearing the symmetry optimised prostheses. These results support the idea that improved gait symmetry, at least for certain variables, is related to reduced energy expenditure, and is therefore an appropriate goal in rehabilitation.

Biofeedback techniques have been used in a variety of areas involving gait rehabilitation. Systems have been built which provide quantitative feedback of temporal (Hirokawa and Matsumura, 1989), kinematic (Fernie *et al.*, 1978), kinetic (Gapsis *et al.*, 1982), or EMG information (Colborne and Olney, 1990), or a combination of these. Such feedback is usually auditory (Fernie *et al.*, 1978) or visual in nature, or both (Hirokawa and Matsumura, 1989; Colborne and Olney, 1990). Gapsis *et al.* (1982) reported the use of a device (the Limb Load Monitor, or LLM) designed to provide auditory feedback of weight bearing information. The authors studied the rehabilitation outcomes of ten subjects with different gait disabilities using the LLM device compared with ten subjects matched for age and diagnosis who did not. The group of patients who used the LLM reached their goals in a significantly shorter period of time than did the control group (7.3±3.0 days versus 13.6±5.8 days, $p<0.001$) (Gapsis *et al.*, 1982). The same device was later used by Gauthier-Gagnon *et al.* (1986) to assist a group of TT amputees in early balance training. These studies demonstrate that biofeedback can be used to improve rehabilitation outcome.

The purpose of the research reported here was to evaluate a newly developed system for assessing and providing feedback of gait symmetry information in real time to subjects walking on a motorised treadmill. The system involved the use of a specially designed device (the "CCF Treadmill") with two force plates mounted under the treadmill belt (Dingwell and Davis, 1995). The CCF Treadmill was used to compare various parameters of gait symmetry between two groups of normal and TT amputee

subjects, and to evaluate the effectiveness of Real-Time Visual Feedback (RTVF) training at reducing gait asymmetries for the TT amputee subjects. Amputee patients are currently evaluated in a subjective manner by trained prosthetists (Kapp and Cummings, 1992), and quantitative gait analyses are usually not performed. There are often difficulties associated with standard gait analysis techniques that require subjects to perform multiple trials walking over ground and placing ther feet on one or more force plates. The data collection effort can be quite costly and time consuming and thus unsuitable for providing "instantaneous" feedback to patients learning to walk in a rehabilitation setting. This study describes the use of a device that was built to address these issues and to improve and expand upon current gait analysis and rehabilitation techniques. It was anticipated that such a device could be an especially effective aid in a rehabilitative context where patients could receive RTVF during the gait retraining process.

## Methods

The current research was performed using a specially designed treadmill/force plate device (the "CCF Treadmill") built at the Cleveland Clinic Foundation. The CCF Treadmill was based on previous treadmill/force plate designs (Kram and Powell, 1989; Davis *et al.*, 1991) and is described in more detail elsewhere (Dingwell and Davis, 1995). The primary advantage of this system is that it allows the collection and comparison of temporal and kinetic parameters of gait for both limbs for multiple, successive strides in real time. These comparisons can be made at a constant known speed, without forcing subjects to target their footsteps. With the use of this treadmill system, large amounts of data can be collected and analysed in a very short period of time. Ten to fifteen complete, consecutive strides of data can be collected in as little as 20 to 25 seconds. Under standard gait laboratory conditions, the collection of such data (for non-consecutive strides) would typically take over an hour to complete. This is an especially significant advantage when dealing with amputee subjects or other rehabilitation patients who cannot physically tolerate walking for more than a few minutes at a time.

Two AMTI force plates (Advanced Mechanical Technologies, Inc., Newton, Massachusetts), model OR6-1, were bolted to an aluminium mounting platform inside the treadmill such that the top surface of the force plates was directly underneath the treadmill belt. The mounting of the force plates has been described by Dingwell and Davis (1995). The treadmill was mounted such that the surface of the treadmill belt was even with the laboratory floor and was further modified to include two adjustable hand rails, mounted on either side of the treadmill and a monitor stand bolted in front of the treadmill to provide subjects with RTVF displays. The setup for the CCF Treadmill is shown in Figure 1. Data were collected and analysed on a Gateway 2000 IBM compatible 486 DX2 computer (© Gateway 2000, Inc.).

Software was written to continuously collect, process, and display gait symmetry information in real time at collection frequencies of up to 100 Hz. Three different displays of RTVF were developed. Each feedback routine was chosen to represent a different type of gait information. The Centre Of Pressure (COP) display was designed to draw the centres of pressure calculated for the left and right feet as the subject walked on the treadmill. Figure 2 (top) shows a representation of this output display, with the superimposed centre of pressure paths for three full strides of gait for both limbs. The divided rectangle shown on the screen represented the top view of the two forces plates in the treadmill. This display traced the path of the subjects' feet as they walked on the treadmill, allowing them to see any differences in the length of stride for either foot, or if the either foot was leading or lagging behind the

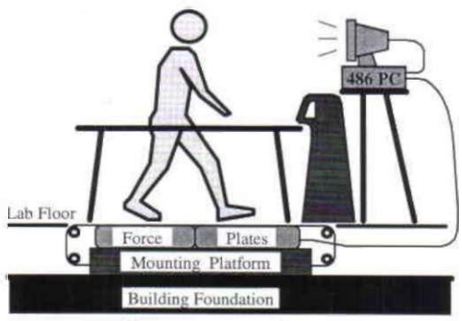


Fig. 1. CCF Treadmill set-up for real-time gait analysis and visual feedback.

other (e.g. the right foot in Figure 2 appears to be slightly leading the left foot at heel strike).

The Percent Stance Time (%ST) display was based on the algorithm's ability to determine times of occurrence of heel strike and toe off. Times of heel strike and toe off were extracted from the time derivative of the smoothed mediolateral position of the centre of pressure curve (d/dt(Dx)). The d/dt (Dx) curve was approximately zero during mid stance, and showed distinct positive and negative peaks when weight was shifted from each foot to the other (Dingwell and Davis, 1995). Stance times were computed from heel strike and toe off times as a percentage of the total stride time by the following equation:

$$\%\ \text{Stance Time} = \frac{(\text{Toe Off} - \text{Previous Heel Strike})}{(\text{Heel Strike} - \text{Previous Heel Strike})} \quad (1)$$

and were displayed graphically for both left and right feet, as shown in Figure 2 (middle). A typical ratio of 60% is indicated by a horizontal dashed line, and the patient's actual data are given numerically and represented as X's on the vertical bar graphs.

Push Off Force (POF) was calculated based on the maximum force recorded on the rear force plate for each foot. An Index of Symmetry (SI) was calculated based on an equation modified from Herzog *et al.* (1989):

$$SI = \frac{(X_{Right} - X_{Left})}{(X_{Right} + X_{Left})} \times 100\% \quad (2)$$

This equation produces an SI with a continuous linear range of values from -100% to +100% with perfect symmetry being equivalent to SI = 0%. The calculated SI value was displayed graphically on a horizontal bar graph (indicated by an 'X'), with the associated numerical value of SI also displayed. A representation of this display is shown in Figure 2 (bottom).

Six normal healthy subjects, with no previous history of lower limb injury, were selected to participate in the study. Subjects were selected whose ages were approximately in the age range of the amputee subjects, though no specific attempts were made to match normal and amputee subjects by age or sex. Mean age of normal subjects was 42.7 years (range 33 to 54 years). All subjects responded that they were right leg dominant when questioned. Six trans-tibial amputee subjects were selected from the patient data base from the Department of Orthotics and Prosthetics and the Cleveland Clinic. All unilateral TT amputee patients who were in good general health, and were judged to be "established walkers," capable of tolerating twenty minutes of treadmill walking, were eligible to participate in the study. The six subjects chosen had a mean age of 41.7 years (range 31 to 69 years), and had been wearing their prostheses on average of 6.0 years (range 6 months to 21 years). Cause of amputation was traumatic in three cases, related to cancer or other illness in two cases, and peripheral

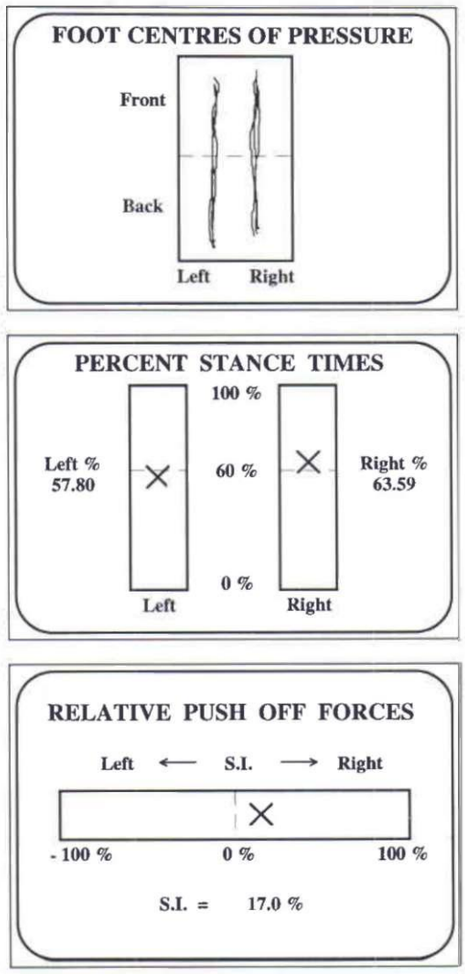


Fig. 2. Three different output displays for real-time visual feeback: COP display (top), %ST display (middle), and POF display (bottom).

vascular disease in one case. All subjects participated on a volunteer basis, and signed appropriate informed consent forms.

Normal and amputee subjects were given four minutes of walking time to acclimatise to the treadmill, and initial asymmetry data was collected. All normal subjects walked at a pace of 2.5 km/hr (0.69 m/s) and amputee subjects walked at self-selected speeds with an average of 2.0 km/hr (0.56 m/s) (range 1.5 to 2.5 km/hr). No instructions were given to either normal or amputee subjects other than to walk normally and no subjects received any visual feedback at any time during treadmill acclimatisation or initial data collection. The TT amputee subjects were then given four minutes of RTVF training with each of the three feedback routines and data were again collected for that routine while the subject was receiving feedback. Subjects were allowed to rest briefly between feedback sessions as needed. While receiving feedback of symmetry information, subjects were instructed to adjust their gait patterns to achieve the most symmetrical gait possible for the particular feedback being given. Subjects received RTVF training in a pseudo-random fashion such that no two subjects were shown the three modes of feedback in the same order.

Three asymmetry variables were quantified for differences between left and right limbs from the RTVF data: foot placement at heel strike (anterior position of the centre of pressure, $SI_{COP}$), percent stance times ($SI_{\%ST}$), and maximum push off forces ($SI_{POF}$). The $SI_{COP}$ variable was calculated to determine if amputee subjects tended to place either foot ahead of the other at heel strike (i.e. if they tended to "lunge" with either their sound limb or their prosthesis). $SI_{COP}$, $SI_{\%ST}$, $SI_{POF}$ were evaluated before and after each respective feedback was given. In addition, a fourth asymmetry variable representing single support times ($SI_{SST}$) was quantified for all four conditions (before feedback, and after each of the three feedback sessions). This variable was examined to determine the effects of the different feedback modes on a variable not directly associated with the feedback parameter itself. $SI_{SST}$ differs from $SI_{\%ST}$ in that the calculations for $SI_{\%ST}$ include the duration of double support, while $SI_{\%ST}$ does not (see Appendix). It was anticipated that double support times would be longer for amputee subjects than for normals, and would also be longer for amputees shifting weight from their sound limb to their prosthesis than vice versa. SI values for each variable were quantified for normal and amputee subjects using the following equations, modified from Herzog *et al.* (1989).

$$SI_{Normal} = \frac{(X_d - X_{nd})}{(X_d + X_{nd})} \times 100\% \qquad (3)$$

$$SI_{Amputee} = \frac{(X_n - X_p)}{(X_n + X_p)} \times 100\% \qquad (4)$$

Where "X" was the measured variable, "d" and "nd" represented dominant and non-dominant limbs, respectively, and "n" and "p" represented natural and prosthetic limbs, respectively.

Twenty five seconds of data, representing 10 to 15 complete strides of walking, were collected for all subjects for each of the specified conditions: normal subjects without feedback, and TT amputee subjects before and after receiving each type of visual feedback training. A variety of statistical tests was performed to compare asymmetries and variability in gait asymmetries between both groups of subjects, and to evaluate the effects of RTVF training. To compare average SI results to the condition of perfect symmetry (SI = 0), data for each subject was averaged, and one sample, two-tailed T-tests for means were performed for all four variables for both groups of subjects (n = 6 subjects per group). Single factor analysis of variance (ANOVA) tests with repeated measures were performed to compare asymmetry data from normals to that for amputees. To determine if amputees showed greater variability in gait asymmetry, standard deviations for ten strides of gait for each subject were computed and compared using a one-tailed T-test for samples of equal variance. To determine if symmetry between different variables of gait were related, the symmetry values for ten strides of gait were averaged for each subject for each variable and correlations between the average values were computed for both groups. The effects of RTVF on the symmetry of TT amputee gait were analysed using a two factor ANOVA with repeated measures to determine differences before and after feedback training for the three RTVF

Table 1. Symmetry indices for normal and amputee subjects. (Mean (average std. dev.) for n = 6 subjects x 10 strides per subject)
(* = significantly different to SI = 10 at p<0.01)

| Variable | Normals | Amputees | p - value |
|---|---|---|---|
| $SI_{COP}$ | –0.18% (1.56%) | –1.54% (1.31%) | 0.380 |
| $SI_{\%ST}$ | +1.84% (1.31%) | +6.98% (3.25%)* | **0.015** |
| $SI_{POF}$ | –1.36% (2.08%) | +2.56% (2.97%) | **0.015** |
| $SI_{SST}$ | +2.47% (2.39%) | +10.57% (4.96%)* | **0.014** |

variables quantified ($SI_{COP}$, $SI_{\%ST}$, and $SI_{POF}$). Data for single support times were collected for all four treatments and a two factor ANOVA with repeated measures was performed to analyse differences between treatments. Individual differences were then compared using the method of least significant differences.

## Results

All of the asymmetries quantified for normal subjects were less than 2.5%. Comparison of these data to the condition of perfect symmetry showed that none of these asymmetries were statistically different to a value of SI = 0 (p >0.10). Data for TT amputee subjects demonstrated significant non-zero asymmetries for $SI_{\%ST}$ and $SI_{SST}$ (p<0.01) each). Asymmetries for $SI_{COP}$ and $SI_{POF}$ did not quite achieve the 0.05 level of significance (p = 0.07 each).

Table 1 shows mean SI values (n = 6 subjects) and ANOVA results comparing asymmetries of normal and amputee subjects. Means (n = 6 subjects) of standard deviations in SI (n = 10 strides per subject) are shown in

Table 2. Correlations between average symmetry variables
(* = significant at p<0.05, ** = significant at p<0.01))

| Normal Subjects (n = 6) | | | |
|---|---|---|---|
| Variable | $SI_{COP}$ | $SI_{\%ST}$ | $SI_{POF}$ |
| $SI_{\%ST}$ | **0.830*** | – | |
| $SI_{POF}$ | –0.361 | –0.569 | – |
| $SI_{SST}$ | **0.862*** | **0.996**** | –0.598 |
| **TT Amputee Subjects (n = 6)** | | | |
| Variable | $SI_{COP}$ | $SI_{\%ST}$ | $SI_{POF}$ |
| $SI_{\%ST}$ | 0.338 | – | |
| $SI_{POF}$ | 0.554 | **0.926**** | – |
| $SI_{SST}$ | 0.279 | **0.986**** | **0.869*** |

Table 3. Average symmetry index values before and after feedback training
(Means for n = 6 subjects x 11 strides per subject)

| Variable | Before | After | p - value |
|---|---|---|---|
| $SI_{COP}$ | –1.58% | –0.56% | **0.002** |
| $SI_{\%ST}$ | +7.03% | +5.18% | **0.007** |
| $SI_{POF}$ | +2.47% | +1.38% | **0.033** |

parentheses. Amputees demonstrated significantly greater asymmetries than normal subjects for three of the four measured variables; $SI_{\%ST}$, $SI_{POF}$, and $SI_{SST}$ and asymmetries for all variables were an average of 4.6 times greater for amputees than for normal subjects. T-tests comparing variabilities of both groups of subjects showed that TT amputee subjects demonstrated greater average variability in three of the four variables quantified; however, this difference was only significant for $SI_{\%ST}$ data (p = 0.03). Increases in variability of push off force and single support time for asymmetries ($SI_{POF}$ and $SI_{SST}$) for TT amputees were not quite significant (p = 0.11 and 0.06 respectively).

Results of correlations between the average values of the four variables quantified are shown in Table 2. Significant correlations were computed between asymmetry values for $SI_{COP}$, $SI_{\%ST}$, $SI_{SST}$ for normal subjects, and between $SI_{\%ST}$, $SI_{SST}$, and $SI_{POF}$ for TT amputees. The remaining six correlations were not significant.

Results of ANOVA analyses comparing changes in asymmetries for TT amputee subjects before and after RTVF training are shown in Table 3, and Figure 3. Significant decreases in the degree of assymetry were demonstrated for all three variables after amputees were shown visual feedback of the data.

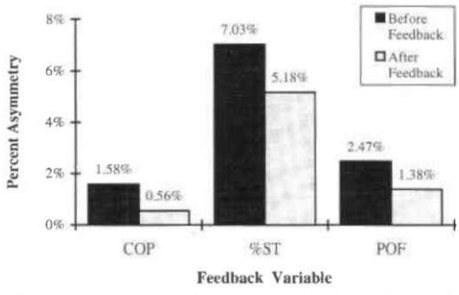


Fig. 3. Improvement in symmetry indices after real-time visual feedback training.

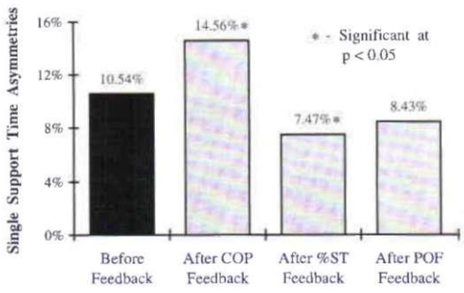


Fig. 4. Changes in single support time asymmetry (SISST) After real-time visual feedback training. (* – significantly different to before feedback at $p < 0.05$).

ANOVA results comparing asymmetries in single support times ($SI_{SST}$) between the four tested conditions indicated significant differences before and after RTVF training and the method of least significant differences was used to determine which feedback routines significantly affected $SI_{SST}$ asymmetries. These results are shown in Figure 4. $SI_{SST}$ data measured before feedback training showed a significant increase after COP feedback training ($p<0.05$), a significant decrease after %ST feedback training ($p>0.05$), and no significant change after POF feedback training.

## Discussions

T-tests comparing average asymmetries of six normal and six TT amputee subjects to a perfect symmetry value of SI = 0 showed that only two of the eight comparisons were statistically significant. However, tests on larger groups of subjects might reveal these asymmetries to be significant. The results of the current study agree qualitatively with the findings of other authors regarding asymmetries in temporal gait patterns (Rosenrot *et al.*, 1980; Hirokawa, 1989) and ground reaction forces (Herzog *et al.*, 1989) of normal subjects. The asymmetries quantified in the current study were slightly greater than those reported by Herzog *et al.*, (1989). Possible reasons for this include the fact that Herzog's data were obtained on a larger sample of subjects than was used in this study. Additionally, subjects in this study were asked to walk at a pace of 2.5 km/hr (0.96 m/s) in order to obtain data more easily comparable to that collected for the TT amputee subjects. Gait patterns of normal subjects have been shown to be more consistent at preferred walking velocities (Rosenrot *et al.*, 1980). Therefore, and increase in the asymmetries of normal subjects' gait patterns may have resulted from asking them to walk at a pace slower than their normal velocity.

Asymmetry values from Table 1 demonstrate that TT amputees spent a significantly reduced time in total stance ($SI_{\%ST} = +6.98\%$) and single stance ($SI_{SST} = +10.57\%$) on their prosthetic limb compared to their natural limb. Although TT amputees also generated less force at terminal stance on their prosthetic limbs ($SI_{POF} = +2.57\%$), this difference was not significant. These increases in temporal asymmetries ($SI_{\%ST}$ and $SI_{SST}$) and peak force magnitudes ($SI_{POF}$) agree with the findings of previous researchers regarding the timing and force profiles of amputee gait patterns (Skinner and Effeney, 1985; Breakey, 1976; Cheung *et al.*, 1983; Seliktar and Mizrahi, 1986; Baker and Hewison, 1990). The variability in stride to stride asymmetries of amputees was greater than that of normals for three of the four variables quantified, although this difference was significant only for percent stance time asymmetry. This increase in variability could de due to the loss of normal neuromuscular control in the amputated limb (Zahedi *et al.*, 1987), to an imperfect socket fit resulting in motion occurring between the stump and the prosthesis, or a combination of these factors. Further investigation should be conducted to confirm these results.

Significant correlations were found between percent stance time and single support time asymmetries for both the normal and amputee subjects and also both percent stance time and single support time asymmetries, and push off force asymmetry for the TT amputee subjects. These positive correlations lend support to the theory that asymmetries in gait cycle timing are directly influenced by a loss of normal push off force in the gait of TT amputees (Breakey, 1976; Winter and Sienko, 1988). However, none of the six remaining correlations was significant, and three were in fact negative, suggesting that while asymmetries of certain variables might be related to each other, the asymmetries of other variables, in general, are not. This idea was supported by the ANOVA results examining the effects of visual feedback training on single support time asymmetry ($SI_{SST}$) which demonstrated that decreases in asymmetry for those variables being displayed

were not necessarily reflected in improved symmetry for other parameters of gait. The TT amputee subjects may have in fact altered their gait patterns to decrease one form of asymmetry by increasing other asymmetries. No known study to date has adequately addressed the questions of how asymmetries measured for different variables are related, or what magnitudes of asymmetry for any of these parameters are necessary to adversely affect gait. In this respect, for future studies trying to use such RTVF to improve rehabilitation outcome, it would be advantageous to identify those specific variables for which attaining a more symmetrical gait would have the greatest consequence for long term benefit.

The results in Table 3 show that asymmetries in the gait patterns for all three feedback variables were significantly reduced after subjects trained with RTVF. These results demonstrate that subjects have the ability to manipulate their walking patterns based on the visual feedback information being given. The subjects of Cheung *et al.* (1983) showed a reduction in asymmetry of percent stance time from +3.7% to +2.4% after six weeks of gait training, a 35% change in asymmetry. Subjects from the current study showed percent times of +7.03% and +5.18% before and after visual feedback training, and effective reduction in asymmetry of 26% in only five minutes. These results, although encouraging, must be interpreted with caution for two primary reasons; it is not yet known whether subjects would be able to maintain these decreased asymmetries in the absence of visual feedback, and it is also not yet clear that these controlled changes in gait symmetry would result in long-term learning of a more symmetrical gait pattern. Since the primary goal of this project was to determine if amputee subjects could respond positively to RTVF, and since this was shown to be the case, the questions of long term gait training effects are left to future research.

## Conclusions

The primary objectives of the current study were to evaluate the gait asymmetry characteristics of a group of normal subjects compared to a group of TT amputee subjects, and to evaluate the effectiveness of giving TT amputee patients RTVF training of gait symmetry information. The normal subjects studied did not demonstrate significantly asymmetrical gait patterns though small asymmetries were recorded. The asymmetries measured in TT amputee gait patterns were greater than those of normal subjects, and agreed qualitatively with asymmetries previously reported in the literature (Breakey, 1976; Cheung *et al.*, 1983; Seliktar and Mizrahi, 1986; Baker and Hewison, 1990). These increases in gait asymmetry are most likely due to the mechanical asymmetries imposed by the prosthesis (Winter and Sienko, 1988) and the loss of normal neuromuscular control and proprioceptive feedback in the amputated limb (Zahedi *et al.*, 1987). RTVF training was shown to be an effective means of producing significant short term reductions in the gait asymmetries of these amputee subjects.

Quantifying asymmetries in amputee gait patterns as they relate to normal subjects is the first step in trying to define what degree of asymmetry is acceptable, or desirable in patients' gait patterns during the rehabilitation process. Devices such as the CCF Treadmill can be useful tools both for defining and quantifying rehabilitation targets, and for measuring patients' progress towards those targets over a period of time. The results of this study should be taken as encouraging, but further study of gait asymmetries needs to be conducted in two areas; first, to identify and define the relationships between asymmetries measured for different variables and their functional relationship to the process of ambulation, and second, to determine the long term rehabilitation benefits of gait retraining with RTVF.

## Appendix

Single support time asymmetries ($SI_{SST}$) were quantified to determine the effects of different feedback modes on a gait parameter not directly associated with the feedback. Although $SI_{SST}$ and $SI_{\%ST}$ were shown to be strongly related, they were different variables in that $SI_{SST}$ omitted the duration of double support. If the gait cycle for a given subject was 1 second long, with 0.62 and 0.58 seconds spent in total support on the sound and prosthetic limbs, respectively, and if the duration of double support were 0.1 seconds, the calculations of $SI_{\%ST}$ and $SI_{SST}$ (from equation 4) would produce the following results:

$$SI_{\%ST} = \frac{(X_n - X_p)}{(X_n + X_p)} \times 100\%$$

$$= \frac{(62\% - 58\%)}{(62\% + 58\%)} \times 100\% = 3.33\% \quad (5)$$

$$SI_{SST} = \frac{(X_n - X_p)}{(X_n + X_p)} \times 100\%$$

$$= \frac{(0.52s - 0.48s)}{(0.52s + 0.48s)} \times 100\% = 4.00\% \quad (6)$$

If, however, the duration of double support were increased to 0.15 seconds, $SI_{\%ST}$ would remain unchanged, but the calculation of $SI_{SST}$ would become:

$$SI_{SST} = \frac{(X_n - X_p)}{(X_n + X_p)} \times 100\%$$

$$= \frac{(0.47s - 0.43s)}{(0.47s + 0.43s)} \times 100\% = 4.44\% \quad (7)$$

Thus for the same percent stance times, $SI_{SST}$ will be greater than $SI_{\%ST}$, and $SI_{SST}$ will increse as the duration of double support increases. This can be seen in the data from Table 1 where $SI_{SST}$ data were greater than $SI_{\%ST}$ data, and $SI_{SST}$ data were also higher for TT amputees than for normal subjects. These data support the notion that amputees spend a greater portion of time in the double support phase of gait, probably to ensure better stability during locomotion.

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# Outcome of fitting an ICEROSS prosthesis: views of trans-tibial amputees

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**Abstract**

A report of the outcome of fitting ICEROSS prostheses to trans-tibial amputees from a sub-regional amputee rehabilitation centre is presented. This work has mainly concentrated on obtaining patients' own views to judge advantages and disadvantages of ICEROSS compared to their previous patellar-tendon-bearing (PTB) prostheses. Sixty-nine patients were entered for this study, but the results of the study are based on 54 patients who responded. Fifteen patients (27.7%) had rejected their ICEROSS prosthesis at the time of the study. Provision of ICEROSS prostheses did not improve indoor and outdoor walking abilities in terms of distance or use of other walking aids, nor were they more comfortable to wear. An increase in sweating in the first 3 months of wearing ICEROSS was significant, but settled afterwards. The amputees considered that the rate of stump skin breakdown with ICEROSS compared to their PTB prostheses was significantly less. Walking up and down stairs was more comfortable and in a general overall rating of ICEROSS prostheses they were scored significantly higher by the amputees themselves. It is concluded that appropriate patient selection is vital and in certain cases ICEROSS will provide considerable benefits to the amputees.

## Introduction

The use of the Icelandic roll on silicone socket (ICEROSS) as a prefabricated socket for lower limb amputees began in 1986, though custom made silicone sockets began several years earlier (Kristinsson, 1993; Fillauer *et al.*, 1989). It is suggested that the main advantage of the ICEROSS is improved suspension and that it also considerably improves the weight bearing capabilities of the prosthesis and the interface between the stump and the prosthesis (Kristinsson, 1993).

Published reports of clinical experience and outcome of fitting ICEROSS prostheses by independent researchers are limited. The experience of using the ICEROSS system for trans-tibial amputees by a team from the Netherlands has been encouraging (Cluitmans *et al.*, 1994). There has also been a report of an audit of 89 trans-tibial amputees fitted with ICEROSS sockets in Birmingham, England (Panagamuwa *et al.*, 1994).

Following appropriate training of the prosthetists, ICEROSS for trans-tibial amputees was formally introduced in April 1993 in a sub-regional rehabilitation centre in Sheffield. The additional financial cost of hardware , as well as the extra time required of the prosthetic staff, necessitates convincing evidence of both short and long term benefits for the amputees. Apart from the professionals' own experience of use of the new system, it was felt equally important to consider the users, i.e. the amputees' own views and opinions when outcomes are considered. In this project the main concentration has been on the trans-tibial amputees' views of the ICEROSS systems. They were all previously using PTB prostheses, thus allowing a comparison between PTB and ICEROSS prostheses.

## Materials and method

Between April 1993 and October 1994, 69 unilateral trans-tibial amputees were provided

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with ICEROSS sockets on Endolite* prostheses with multiflex ankle joints. An ICEROSS roll on silicone sleeve was used in all cases in conjunction with a polypropylene outer socket and they were connected to each other by means of a "shuttle lock" mechanism. In all cases a thin stump sock was worn between the silicone sleeve and the outer hard socket. All these patients were previously using Endolite prostheses with mutliflex ankle joints and polypropylene PTB sockets with inner Pelite liners. The decision to change to an ICEROSS for these patients was taken in the prosthetic clinics by the rehabilitation physician in conjunction with the prosthetist, with full discussion with the amputee. In some cases the amputee had enquired about the ICEROSS system, having seen a commercial advertisement. An ICEROSS system was not provided to amputees with poor hand function or to patients who could not reach their stump with both hands due to major restriction of joint movements or other reasons.

A composite questionnaire was devised after an initial pilot excercise to ascertain users' views. The three page questionnaire included direct questions e.g. "how many hours per day do you wear your prosthesis on average", some 2 or 3 point response closed questions, e.g. Yes/No or Same/More/Less and some questions with response on a digital score of 0-5 (0 = very poor, 5 = very good). The questionnaire also included 3 open questions inviting comments on users' own perceived advantages and disadvantages and their own ideas for possible areas of improvement of the ICEROSS system. In all but the 3 open questions, patients' response for both ICEROSS and PTB were specifically required so that a comparison could be made.

The main indications for changing over to the ICEROSS system were, problems with suspension, skin problems e.g. skin grafts or very scarred stumps vulnerable to frequent breakdown. In 13 patients ICEROSS was prescribed for young active amputees where it was felt that improved weight bearing tolerance of the ICEROSS and possible reduction of the shear forces to the skin of the stump could be beneficial. Statistical analyses of the responses to questionnaires were done by using *t*-tests for parametric data. McNemar test and Wilcoxon matched pairs signed ranks test were used for statistical analysis of non-parametric data.

## Results

Out of 69 amputees who were sent the questionnaire, 54 returned their questionnaires giving a response rate of 78.26%. Out of these 54 patients amputation had been carried out due to trauma in 27, vascular disease and/or diabetes mellitus in 11, congenital limb deficiency in 6 and other miscellaneous causes in 10 patients. The PTB prostheses for these 54 patients prior to the supply of ICEROSS were suspended by leather cuff suspension in 31, self-suspending supracondylar sockets in 21 and by elasticated sleeve suspension in 2 patients.

All analyses of results are from these 54 returned questionnaires. The questionnaires were incomplete in some instances and these were taken into account in presenting the results and analyses.

The average age of the amputees was 48.35 years (range 22-80 years). At the time of the questionnaire, 15 out of 54 were not using the ICEROSS and reverted back to their old PTB prostheses. Of these 15 patients, 10 had stopped using the ICEROSS due to the development of skin problems e.g. marked skin rash, blisters, sometimes associated with excessive sweating, 4 patients had stopped due to pain and discomfort at the distal end of the stump and 1 patient felt insecure with the ICEROSS system as he missed the mediolateral knee joint support of a supracondylar PTB socket.

The respondents had worn their ICEROSS systems for between 2 and 104 weeks (mean 21.22 weeks). A comparison was made between PTB and ICEROSS of the number of hours the limb was said to be worn per day. An average use of 12.26 hours per day for PTB and 10.42 hours per day for ICEROSS was not significantly different (*t*-test, $p = 0.074$). Similarly a comparison between the two types of prostheses of distance said to be walked per day was not significant (*t*-test, $p = 0.776$).

Trans-tibial amputees who were provided with ICEROSS prostheses did not wear them longer, did not walk longer distances, did not find walking on rough ground any easier and did not use walking aids any less compared to their PTB prostheses. Sweating of stump was significantly increased in the first three weeks of using ICEROSS, but settled after 3 weeks. Skin breakdown tended to be less and walking

*Trade name of Blatchford modular, carbon fibre endoskeletal prosthesis.

Table 1. Analysis of response to questions requiring direct Yes/No answers

| Information from questionnaire | Number of respondents | With PTB | With ICEROSS | p value | Statistical significance |
|---|---|---|---|---|---|
| Use of walking aids, indoors | 49 | 11 | 6 | 0.2168# | NS* |
| Use of walking aids, outdoors | 51 | 25 | 26 | 0.6875# | NS* |
| Use of walking aids on rough ground and bad weather | 45 | 24 | 22 | 1.0000# | NS* |
| Presence of pain in the stump | 46 | 10 | 8 | 0.8145# | NS* |
| Presence of skin breakdown of the stump | 47 | 32 | 28 | 0.1153# | NS* |

*NS – Not significant # – McNemar test

up and down stairs was significantly improved compared to the PTB prosthesis. The amputees rated ICEROSS significantly higher than the PTB prosthesis in the overall rating.

Responses and detailed analyses of the questions are presented on Tables 1, 2 and 3.

The response to the open ended questions could not be analysed statistically and many did not comment on these questions. Some mentioned, as expected, more than one advantage, disadvantage or suggestion. These responses have been collated to individual grouping and are presented in Tables 4 and 5.

## Discussions

This study is based on the amputees' own experience of using the ICEROSS system. As all patients in this study were established PTB prosthesis users, this provides an opportunity to make a comparison of amputees' subjective opinion between the two types of prosthesis. Though the responses are generally subjective in nature, they appear to coincide with clinical observation, as the patients were regularly and routinely reviewed in the clinic by the same team. It is therefore felt that the responses and comments given by the amputees are a generally accurate reflection of their perceptions.

Following the introduction of the ICEROSS system in the clinic, all members of the team are on a learning curve in respect of identifying

Table 2. Analysis of response to questions with 3 response choices

| Information from questionnaire | Number of respondents | Same | Less | more | p value | Statistical significance |
|---|---|---|---|---|---|---|
| Rate of skin breakdown with ICEROSS compared to PTB | 49 | 10 | 26 | 13 | 0.0376# | Some significance |
| Rate of sweating in first 3 weeks with ICEROSS compared to PTB | 52 | 8 | 4 | 40 | 0.0003# | Strong significance |
| Rate of sweating with ICEROSS after 3 weeks compared to PTB | 49 | 14 | 15 | 20 | 0.3954# | No significance |

# – McNemar test

Table 3. Analysis of information derived from questions using a digital scale (range 0-5; 0=Very poor, 5=Very good)

| Information from questionnaire | ICEROSS | | PTB | | p value | Statistical significance |
|---|---|---|---|---|---|---|
| | Number of respondents | Mean score | Number of respondents | Mean more | | |
| Comfort of wearing | 52 | 3.50 | 51 | 2.82 | 0.679# | NS* |
| Comfort of walking for long distances | 51 | 3.06 | 50 | 2.66 | 0.2976# | NS* |
| Ease of donning and doffing | 52 | 3.63 | 51 | 3.25 | 0.2399# | NS* |
| Comfort of walking over rough terrain | 51 | 3.00 | 50 | 2.52 | 0.1157# | NS* |
| Ease of maintenance of prosthesis | 50 | 3.86 | 50 | 3.36 | 0.0546# | NS* |
| Comfort of climbing stairs | 51 | 3.55 | 50 | 2.82 | 0.0230# | Some significance |
| Comfort of coming down stairs | 51 | 3.43 | 50 | 2.90 | 0.0390# | Some significance |
| Overall rating | 49 | 3.82 | 49 | 3.12 | 0.0182# | Some significance |

*NS - Not Significant # – Wilcoxon matched pairs signed ranks test

indications, and contra-indications, casting methods and fabrication. The "success rate" may therefore improve by more appropriate patient selection and the application of improved technical expertise.

The rejection of the ICEROSS by 15 patients is of some concern. In 10 of these 15 patients, the ICEROSS was rejected because of skin rash, blisters, irritation and marked sweating, singularly or in combination. The above problems continued for a prolonged period and the decision to reject was taken after an adequate trial in all cases. Two of the 15 found the tightness of the sleeve at the distal end of the stump was too painful to be able to continue wearing their ICEROSS.

In some cases this problem could now be resolved as a greater range of silicone sleeve

Table 4. Responses to open questions regarding advantages of ICEROSS prosthesis

| Advantages mentioned | Number of respondents |
|---|---|
| Better comfort | 14 |
| Better donning and doffing | 12 |
| Less friction/less skin problems | 8 |
| Secure/better suspension | 8 |
| Better movement of knee | 5 |
| Better cosmesis | 2 |
| Better control | 2 |
| No more cuffs and straps leading to less wear and tear of clothes | 2 |
| Feels part of human body | 1 |
| ICEROSS is the only prosthesis which is good | 1 |

Table 5. Responses to open questions regarding disadvantages of ICEROSS prosthesis

| Disadvantages mentioned | Number of respondents |
|---|---|
| Skin ulceration/itching/rash/sweating | 20 |
| Difficulty in donning and doffing | 7 |
| Connection at bottom of stump | 4 |
| Discomfort/feels the socket is hard | 3 |
| Needs washing frequently | 2 |
| Prosthesis feels insecure/rotates | 2 |
| Wool sock gets stuck in lock | 1 |
| Sleeve gets damaged | 1 |
| Prosthesis feels heavier | 1 |
| Constriction at top of sleeve | 1 |

sizes is now available where previously some patients fell between two sizes. The team believes that for patients who have marked tenderness and hypersensitivity, especially at the distal end of the stump, the ICEROSS system is likely to fail. One blind patient, who is an insulin dependent diabetic and who felt very comfortable with the ICEROSS and liked its improved suspension, developed a deep ulceration over the head of the fibula. It is believed that this is because of a marked increase in the use of her prosthesis because of its advantages. This patient, however, has diabetic neuropathy and a non-sensate stump and was unable to feel or see the results of increased pressure over the head of the fibula – before the ulceration developed.

The finding of significant increase in sweating for about the first 3 weeks of use of the ICEROSS and the difference in sweating after this period compared to the PTB prostheses becomes non-significant, corroborating clinical experience.

The observation of some significant reduction of skin breakdown is worthy of note. The tendency had been to provide the ICEROSS to patients who were troubled with vulnerable stump skin, e.g. split skin grafts, adherent scarring resulting in frequent stump breakdowns from PTB prostheses. So, it is possible that the rate of skin breakdown might have been even lower if patients had not been pre-selected for these reasons. The case of major skin breakdown in the diabetic patient, reported above is something which the professionals and amputees must be aware and vigilant.

It is believed that comfort in climbing and descending stairs is due to the improved suspension of the ICEROSS system and both these activities were significantly easier according to the amputees. Patients are concerned about increased sweating when using the ICEROSS. They should be informed that for most patients this increased sweating settles after the first 3 weeks. Twenty patients felt that sweating, itching and skin rash were the main disadvantages of ICEROSS system. Some patients commented that it was easier to wash the silicone liner and wipe it dry and that the ability to wear the limb immediately was an advantage over the traditional Pelite liner for the PTB sockets. The increased weight of the ICEROSS prrostheses compared to the PTB prostheses did not appear to cause any difficulties for any of the patients. It would appear that the improved suspension of the ICEROSS negated the theoretical problem of the increased weight of the prostheses.

Improved suspension, reduction of skin breakdown and better overall rating of the ICEROSS compared to the PTB by amputees are important observations to be taken into account when selecting the type of prosthesis for the trans-tibial amputee. From the writers' experience and from the literature, there is no convincing clinical evidence to suggest that any significant gains could be achieved by considering ICEROSS as a "standard" prosthesis for all trans-tibial amputees. Rejection of ICEROSS by 15 amputees in this study suggests that ICEROSS sockets may not be suitable as standard prostheses for all amputees. Panagamuwa *et al.* (1994) also reported that 36% of patients provided with ICEROSS prostheses did not have a satisfactory outcome. Improving technique in fitting, improved availability of sizes and types of silicone sleeves and appropriate patient selection should decrease failure or rejection rate. In the light of this current limited and short term experience the authors reserve ICEROSS sockets for trans-tibial amputees who are having, or are likely to have, significant problems with suspension and stump skin breakdown. Appropriate selection of the type of prosthesis for an individual amputee can only be determined by a thorough and complete assessment of the patient, combined with knowledge, expertise and the availability of appropriate prosthetic technology.

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# A comparison of energy expenditure by a high level trans-femoral amputee using the Intelligent Prosthesis and conventionally damped prosthetic limbs

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**Abstract**

Comparisons were made between the Intelligent Prosthesis (IP), Mauch and pneumatic swing phase control damping systems on the same prosthesis worn by a high level trans-femoral amputee. Speeds self selected by corridor walking (4.4 – 5.5 $kmh^{-1}$) proved not to be sustainable for treadmill walking. Comfortable speeds were attained when the subject walked on a treadmill at 2.0, 2.6 and 3.2 $kmh^{-1}$ in two tests for each prosthesis type. Oxygen uptake ($\dot{V}O_2$), cadence and heart rate were measured over 5 minute walks interspersed with rest periods.

Spearman's correlation was used to test for differences between prosthesis types at each speed. At the two slower speeds no significant difference was found, but at the higher speed of 3.2 $kmh^{-1}$, the IP was associated with a significantly lower $\dot{V}O_2$ ($p<0.05$). A two way analysis of variance with replication (ANOVA) demonstrated a significant difference between $\dot{V}O_2$ for different limb types ($p=0.015$). A square law function was fitted to the mean $\dot{V}O_2$ for each prosthesis type by the method of least squares regression. ANOVA demonstrated a significant difference between velocity coefficients for the different prosthesis types ($p<0.05$). Cadence was almost constant during the period of each walk, varying by 1 step $min^{-1}$ at most. However the test-retest differences in cadence were considerable.

It is concluded that there was little difference in energy expenditure between prosthesis types at slower speeds, but at higher speeds (=>3.2km $h^{-1}$) the IP gave a lower oxygen uptake by about 10%.

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## Introduction

Conventionally damped prosthetic limbs use a pneumatic or hydraulic damping cylinder which is adjusted by the prosthetist to provide optimum gait parameters at the subject's customary walking speed (CWS). If the amputee walks at a different speed, he or she must compensate for the pendulum action of the prosthesis in order to alter stride length or step rate by tilting the pelvis to delay extension or by " throwing the leg through", in order to ensure that the foot is in the right place for the next heel strike. This not only leads to an abnormal gait, but requires extra physical effort. In 1993, an "Intelligent Prosthesis" (IP) was introduced (Chas. A Blatchford & Sons Ltd) featuring a microprocessor controlled knee extension damper. The IP uses a proximity switch to detect the step time and automatically alters the level of knee extension damping to suit, using a motor driven needle valve on a pneumatic cylinder. Thus the knee should extend at a rate appropriate to the actual walking speed, removing the need to compensate and reducing effort.

Initial measurements and the results of a 100 subject survey of IP users were reported by Zahedi (1993). This early report suggested that the IP could reduce the physiological cost of walking by as much as 10%, that gait deviations are reduced and that optimum walking speed and range of speeds are increased. It is not clear

whether optimum walking speed refers to CWS or to most metabolically efficient walking speed, not necessarily identical (Jaegers *et al.*, 1993). Amputees reported that the IP was not as tiring as conventionally damped prostheses, and that manoeuvring around obstacles was easier. Observers reported that the amputees walked more naturally and smoothly.

This study aimed to compare the relative energy expenditure necessary to walk at different speeds using the IP with conventional pneumatic and hydraulic prostheses. Energy expenditure was determined by measurement of rate of oxygen consumption ($\dot{V}O_2$) (Astrand and Rodahl, 1977) at three different speeds of treadmill walking. Early results were reported by Clark (1994).

## Methods

*Subject*

The subject (one of the authors) was an active 33 year old male, an established amputee with a high level amputation (due to trauma) at the proximal quarter femur level.

The subject was taking antihypertensive drugs so heart rate could not be used as an indicator of energy expenditure. It was not expected that antihypertensive drugs would affect oxygen consumption.

*Assessment of walking speeds*

The manufacturer recommends that the IP is programmed for the subjects customary, fast and slow walking speeds. The IP parameters are programmed with the subject wearing the IP and walking in a straight line at self selected speed. It was planned to use these speeds for treadmill walking. These 3 self selected speeds were measured by timing the central 5 metres of 3 corridor walks of 10 metres at each speed. Slow, normal and fast walking speeds were 4.4, 5.1 and 5.5 $kmh^{-1}$ respectively.

The subject was then introduced to treadmill walking and practised at increasing speeds in order to acclimatise to the unfamiliar walking technique. It became apparent that the subjects self selected speeds could not be sustained on the treadmill for periods long enough for energy expenditure to be reliably measured.

A second set of slow, comfortable and fast speeds was then determined by the subject for treadmill walking. These were measured at 2.0, 2.6 and 3.2 $kmh^{-1}$ respectively.

The IP was reprogrammed for these new speeds which were felt by the subject to be more representative of, for example, walking in the street rather than his customary walk between rooms at his place of work. They are also close to the CWS for traumatic transfemoral amputees (3.1 $kmh^{-1}$) found by Waters and Yakura (1989).

*Prostheses tested*

Four prosthesis types were tested:

- intelligent prosthesis (IP on)
- intelligent prosthesis programmed for constant damping (IP off)
- Mauch SNS hydraulic swing phase controller (MAUCH)
- Endolite pneumatic swing phase controller (PSPC).

All prostheses used had Endolite StanceFlex knees and the same quadrilateral socket, rigid pelvic belt and Seattle foot. Socket alignment was preserved by splitting the knee joint at the StanceFlex pivot pin leaving the alignment coupling attached to the socket.

The IP was programmed for the treadmill speeds 2 indicated previously (IP on). The case of constant damping (IP off) was included as it has been used to simulate the PSPC (Zahedi, 1993).

The conventionally damped prostheses were adjusted according to the manufacturers instructions and all prostheses included a foam cosmesis. In order to retain clinical validity, no attempt was made to equalise the weights of the prostheses. The subject had acquired at least 5 weeks experience with walking on each limb.

*Oxygen uptake*

$\dot{V}O_2$ and rate of carbon dioxide production ($\dot{V}CO_2$) (ml $kg^{-1}$ $min^{-1}$) were measured using an Oxycon Gamma gas analyser fitted with a paramagnetic oxygen analyser and infra-red carbon dioxide analyser. The subject walked on the treadmill at the three identified walking speeds. It was expected that these speeds would represent exercise below the subjects anaerobic threshold (Waters and Yakura, 1989). Heart rate was continuously monitored.

*Treadmill testing*

The test limb was worn for at least 4 days before the treadmill walk to enable acclimatisation. Tests were carried out at 3pm on Fridays in order to minimise on confounding

Table 1. Oxygen uptake ($\dot{V}O_2$) for two trials of each limb type at the three test speeds

| | Speed ($kmh^{-1}$) | 2 | 2.6 | 3.2 |
|---|---|---|---|---|
| Oxygen Uptake $\dot{V}O_2$ ($ml\ kg^{-1}\ min^{-1}$) | PSPC | 12.27<br>13.82 | 14.55<br>16.16 | 17.83<br>19.18 |
| | MAUCH | 13.09<br>12.70 | 14.48<br>14.45 | 16.87<br>16.84 |
| | IP on | 12.84<br>12.58 | 14.56<br>13.66 | 15.45<br>16.40 |
| | IP off | 12.59<br>11.64 | 14.03<br>13.31 | 17.74<br>17.17 |

factors due to variation in daily work patterns, eating, Circadian rhythms etc. Ambient temperatures were between 19.1 and 22.6° C. The subject's weight varied from 85.8 to 87.2 kg during the period of the tests.

Tests were carried out in order of increasing speed with 5 minutes walking with a 15 minute rest after the first test and a 30 minute rest after the second. $\dot{V}O_2$ and $\dot{V}CO_2$ were measured at rest and throughout the walk. Heart rate was recorded at rest and at 30 second intervals. Cadence was averaged over a 30 second interval at the beginning and end of each walk. Two separate test series were carried out on each prosthesis type, in the following sequence: PSPC, MAUCH, IP on, IP off, PSPC, MAUCH, IP on, IP off.

## Results

$\dot{V}O_2$ and heart rate were calculated by averaging the measurements for the final 3 minutes of each 5 minute walk.

*Oxygen uptake*

Table 1 shows $\dot{V}O_2$ measured for each prosthesis type at the three test speeds. Spearman's correlation was used to test for differences between prosthesis types at each speed. At the two slower speeds no significant difference was found, but at the higher speed of 3.2 $kmh^{-1}$ the IP was associated with a significantly lower $\dot{V}O_2$ ($p<0.05$). This result is similar to those of Molen (1973) and James (1973) who found no significant differences in $\dot{V}O_2$ between leg amputees and non-amputees until speeds of 3.6 $kmh^{-1}$ (trans-tibial) and 3.9 $kmh^{-1}$ (trans-femoral) were reached. A two way analysis of variance with replication (ANOVA) shows a significant difference between $\dot{V}O_2$ for different limb types ($p=0.015$).

Several workers have reported a linear correlation between the square of walking speed ($v^2$) and energy expenditure (Molen, 1973, James, 1973). The function $\dot{V}O_2 = 1+kv^2$ (where $1$ and $k$ are constants) was fitted to the mean of the $\dot{V}O_2$ points for each limb and speed by the

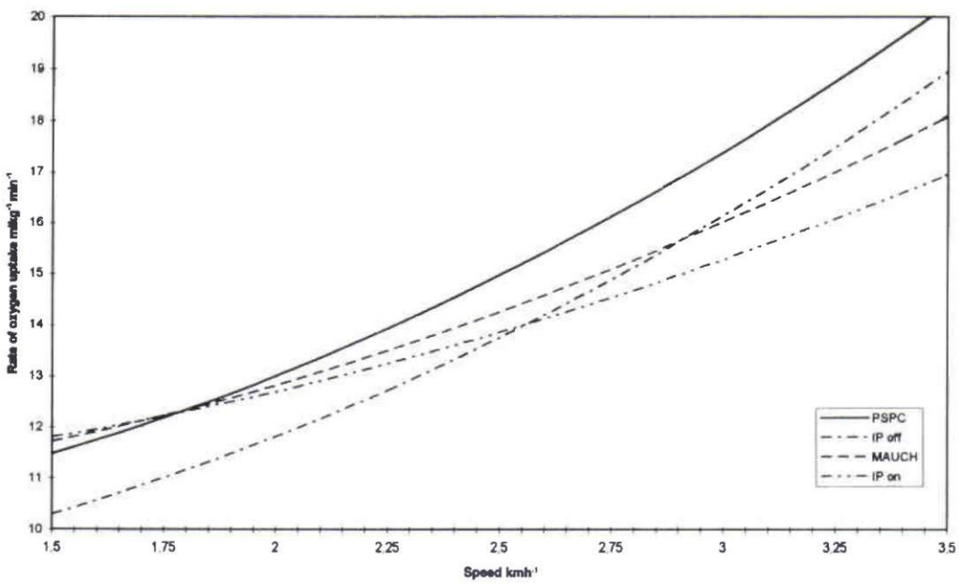


Fig. 1. Square law curves of the form $\dot{V}O_2 = 1+ kv^2$ fitted to oxygen uptake measurements for the Intelligent Prosthesis and three conventionally damped prostheses.

method of least squares (shown graphically in Fig. 1). ANOVA on the coefficients produced demonstrates a significant difference between velocity coefficients (k) for the different limb types ($p<0.05$).

Respiratory quotient ($\dot{V}O_2$ / $\dot{V}CO_2$) did not rise above 1.0 indicating that aerobic work was being done.

No relationship was found between heart rate and prosthesis type.

*Cadence*

Table 2 shows the cadences adopted for the four prosthesis types in each trial. Cadence was almost constant during the period of each walk, varying by 1 step $min^{-1}$ at most. However the test-retest differences were considerable, as illustrated in Figure 2.

Table 2. Cadence for two trials of each limb type at the three test speeds

| | Speed ($kmh^{-1}$) | 2.0 | 2.6 | 3.2 |
|---|---|---|---|---|
| Cadence ($steps\ min^{-1}$) | PSPC | 76<br>71 | 93<br>100 | 106<br>114 |
| | MAUCH | 77<br>80 | 91<br>94 | 104<br>101 |
| | IP on | 70<br>85 | 96<br>96 | 104<br>108 |
| | IP off | 81<br>80 | 88<br>93 | 106<br>108 |

## Discussion

It is clearly dangerous to generalise from results obtained with only one subject. However this experience is illustrative of the clinical problems presented by a high level, active amputee who could be expected to exploit fully the capabilities of the IP.

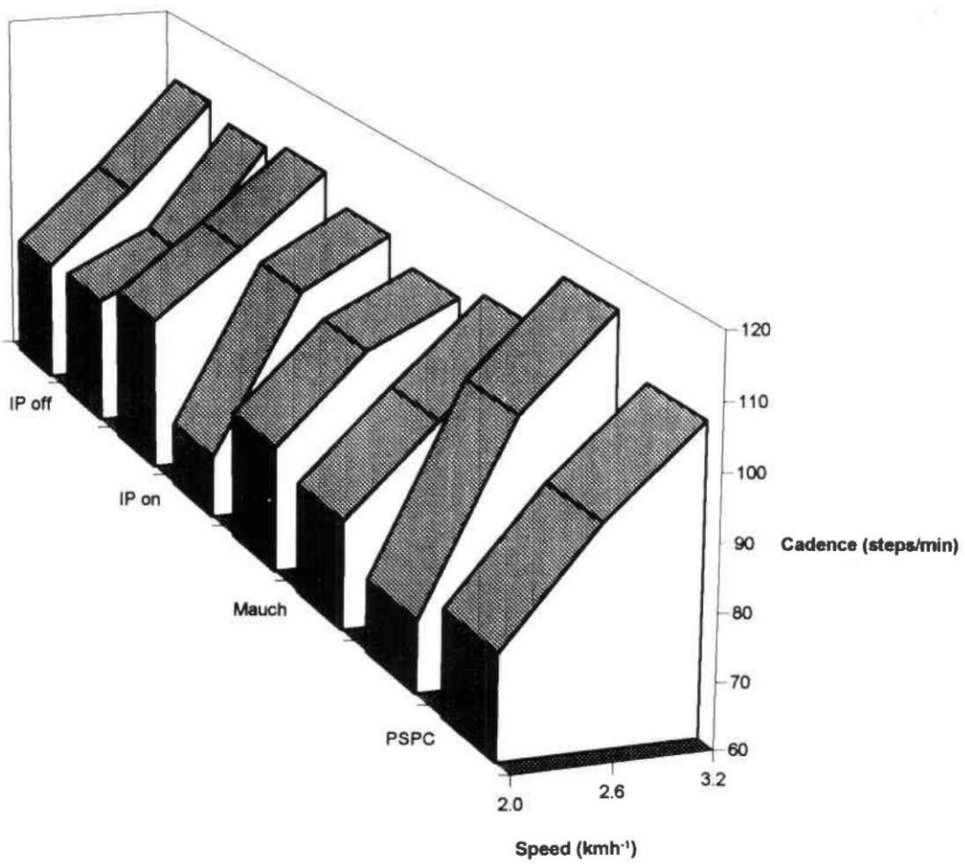


Fig. 2. Cadences adopted for two trials of each prosthesis type at the three test speeds.

The empirical, subjective method of assessing CWS as recommended by the manufacturer calls into question the validity of programming the IP for a range of speeds around this CWS, particularly in a device which it is claimed will alter the ease of walking at different speeds and even increase the CWS (Zahedi 1993). The manufacturer in fact recommends re-evaluation of CWS after a period of experience with the prosthesis. The authors were unable to evaluate changes in CWS as it had not been measured prior to IP use.

The subjects initial corridor walk seemed typical of his usual speed of ambulation at work in the limb centre. This CWS of 5.1 $kmh^{-1}$ is considerably higher than that found by Waters and Yakura (1989) and could not be sustained for 5 minutes of treadmill walking. The determination of CWS using treadmill walking was felt to yield speeds more representative of a middle distance walk — for example, when walking in the street. This CWS was closer to those found by Waters and Yakura (1989). Although it may be suspected that there are physiological differences between treadmill walking and free walking (Mattsson, 1989), several workers have used similar methods of CWS determination on a treadmill (Herbert *et al.*, 1994; Jaegers *et al.*, 1993). Waters and Yakura (1989) found no significant differences in the energy expenditure of non-amputees between free walking and treadmill walking.

Workers looking at normal walking speeds have generally compared different people walking in the same situation. For example Finley and Cody (1970) made covert measurements on people walking a 50 foot straight line in outdoor urban locations. Similar results were found by Waters *et al.* (1988) using an outdoor circular track with an instruction to walk at a comfortable pace. Various workers including Gailey *et al.* (1994) and Nene (1993) have used 'L' shaped or 'figure-of-eight' indoor tracks. Although intra-study CWS measurements will be valid, differences between studies would be important when selecting a 'typical' CWS for use in all walking situations, as in the case of IP set-up. For the full exploitation of the IP's adaptability further work should be done on the influence of environment on the range of amputee walking speeds.

$\dot{V}O_2$ measurements indicate that energy savings at low speeds are not significant. At speeds of 3.2 $kmh^{-1}$ and above energy savings of from 5% (MAUCH) to 15% (PSPC) may be obtainable for treadmill walking. Extrapolating a square law equation to the subjects normal walking speed as initially assessed (5.1 $kmh^{-1}$) would again give greater savings.

Considerable test-retest variation is present in $\dot{V}O_2$ measurements. Little has been reported on test-retest variation of $\dot{V}O_2$ measurements on amputees, however Herbert *et al.* (1994) found a test-retest $\dot{V}O_2$ variation of the same order as the difference between amputee and non-amputee children walking at CWS. Changes in resting $\dot{V}O_2$ are unlikely to be attributable to changes in fitness of the subject over the period of the study.

There is considerable variation in the cadence patterns adopted between test pairs. This is in contrast to the unvarying nature of intra-test cadence and the results of Jaegers *et al.* (1993) who suggested that amputee cadence varies less than that of non-amputees. However variability in cadence will result in increased variability in $\dot{V}O_2$. Lukin *et al.* (1967) demonstrated that each step entails the raising of ones centre of gravity with its attendant work in acquiring potential energy. Thus ambulation at the same speed with increasing cadence will expend increasing amounts of energy. Empirical models of energy expenditure incorporating both cadence and stride length developed for non-amputees have not been validated for amputees (Zarrugh *et al.*, 1974). Cadence variation may result from the subject's attempts to cope with the unfamiliar technique of treadmill. It might be expected that the IP's method of measuring step time in order to adapt to different cadences might increase the range of cadences possible for a given speed. This was not evident. However, further work is needed to explore the relationship between speed, cadence and energy expenditure particularly in the use of IP.

Change in heart rate and physiological cost index are widely accepted as measures of energy expenditure. This study found no relationship between heart rate and walking speed. This finding was predicted because of the action of antihypertensive drugs on heart rate control mechanisms.

It is possible that measuring energy expenditure at constant speed is not the most

sensitive measure of ease of ambulation when evaluating prosthetic lower limbs. Amputees walk more slowly than non-amputees and differences in energy expenditure are only evident at higher speeds. Also a large percentage of ambulation is spent manoeuvring around objects, walking on uneven terrain, changing speed, sitting down and standing up rather than steady level walking. Although some manoeuvres do not involve knee flexion it is possible that evaluation under more realistic conditions may be more revealing.

## Conclusion

Oxygen uptake measured at slower speeds showed no demonstrable difference between limb types when used by this high level amputee. At higher speeds (=>3.2 $kmh^{-1}$) the IP gave a lower oxygen uptake of about 10%.

Relatively large variations in cadence were observed between tests on the same prosthesis type (although not within tests), contributing to variation in energy expenditure.

## Acknowledgement

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# The potential for ambulation by severely handicapped cerebral palsy patients

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**Abstract**

Whereas walking for paraplegic patients is now a routine clinical option, ambulation for heavily handicapped cerebral palsy patients is less well established. There are good reasons for supposing that therapeutic benefits similar to that achieved with paraplegic patients are possible for this group. However, the biomechanical problems which must be overcome are different and in many ways more difficult to address.

The most important factors are identified as appropriate truncal support, control of abduction/adduction, rotation and flexion range at the hip, and of knee flexion.

A means of applying these controls is described as being a combination of a walking frame and orthosis. The special walking frame provides adjustable support at thoracic, abdominal and sacral levels and incorporates castor steering and upper limb support. Controls on lower limb movements are applied through a special orthosis which has a readily available variation of specification to accommodate the wide range of conditions met in cerebral palsy.

Practical application of a prototype system has shown that it can enable patients to walk unaided. However, in most cases it was used indoors only.

Evaluation of the experience in applying the system has enabled the fundamental principles to permit more practical applications to be defined.

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## Introduction

Walking for paraplegic patients is now a routine clinical option which is widely adopted (Douglas *et al.*, 1983; Motloch, 1980; Kirtley, 1992; Lissons, 1992; Butler and Major, 1987). None of the systems described has sought to address the problem of providing walking for total body involved cerebral palsy patients. There are good reasons for this in that lack of control in the upper limbs and trunk creates a different challenge of patient stabilisation. In the lower limbs there is also a different biomechanical situation with the potential for harnessing the available active motor power, despite the inherent lack of motor control.

The objectives of walking for paraplegic patients have been widely proposed as (Rose, 1983; Menelaus, 1987):

therapeutic benefit;

improvement of independence.

Confirmation that these objectives can be met in high lesion spina bifida patients has been reported by Mazur *et al.* (1989) who showed not only that non-walkers had five times the number of pressure sores and twice the number of bone fractures, but also that paediatric patients who walk are more than three times more likely to be able to move around the community independently. The success of properly controlled clinical provision of ambulation to spina bifida patients suggests that the benefits can be reproduced for other clinical groups.

Whereas many congenital or neonatal pathologies have been decreasing, cerebral palsy has remained stubbornly consistent at 3.5 per 1000 live births (Pharaoh *et al.*, 1990). Since survival rates of cerebral palsy patients

are now increasing sharply (Alberman and Botting, 1991) ever increasing numbers of those who are severely handicapped are presenting at orthopaedic clinics. There is often an ambition of patient, parent and/or carer for walking to be achieved and the experience with spina bifida patients reported by Mazur *et al.* (1989) indicates that there is therapeutic benefit in such activity. Innovative solutions are necessary for the stability requirements of the total body involved cerebral palsy patient and the harnessing of the active power available in their lower limbs. However, if successful walking systems can be developed for this group they could bring about important benefits, and there is clear justification for a fundamental examination of the problems which need to be addressed. Early attempts (ORLAU, 1979) centred on simple modifications of paraplegic walking orthoses. Whilst these achieved limited success they also highlighted significant differences which demanded more in-depth examination. Empirical development gradually led to a more detailed understanding of the requirements of ambulation for severely handicapped cerebral palsy patients (ORLAU, 1982, 1984 and 1987; Thompson and Patrick, 1990). Further consideration of the empirical experience has enabled some basic principles to be established, and these can lead to further improvements in the future.

## The biomechanical problem

Whereas the biomechanics of providing ambulation for the paraplegic patient are clearly definable (Stallard *et al.*, 1986) the requirements for stabilisation and propultion in total body involved ataxic or athetoid cerebral palsy patients are much less clear cut. A further complication is the variability of the condition, which is in contrast to the paraplegic patient where the only variabilities are level of lesion, patient shape, available joint ranges and upper limb strength and co-ordination. Nevertheless careful observation of athetoid and ataxic patients shows that there is sufficient commonality in the condition to premit some basic design criteria to be established.

Aberrant movements in the upper limbs make them a poor source of propulsive input. Overlying this is a lack of truncal control which means that the trunk cannot be relied upon to provide a stable platform from which to control the available active power in the lower limbs. The lower limbs also have poor voluntary control, though they are generally capable of generating powerful torques about all of the joints. This means that they are unable to provide the necessary internal stabilisation to prevent collapse of the skeletal structure when standing nor produce controlled propulsive forces.

If a patient is given the necessary truncal support to prevent collapse a variety of aberrant movements in the lower limbs may be observed which would require external control to permit an appropriately patterned reciprocal walking action:

(i) abduction/adduction;
(ii) internal/external rotation about the hip joint;
(iii) intermittent excessive hip flexion;
(iv) intermittent knee flexion.

It would appear to be rare for all of these requirements to be present in all patients. Therein lies the major aspect of the variability and this is compounded by the severity and frequency of the aberrations.

## Orthotic solutions

The fundamental biomechanical problems which need to be addressed were observed during experimental provision of orthoses and walking frames in previous studies of cerebral palsy patients (ORLAU, 1987; Thompson and Patrick, 1990). It became apparent that two primary elements are required:

(i) a walking frame which provides appropriate truncal alignment and support;
(ii) a hip-knee-ankle-foot orthosis (HKAFO) which resists and controls the aberrant hip adduction, abduction and rotation, and limits hip and knee flexion and extension ranges.

A walking frame requires to give both sagittal and coronal truncal support and additionally may need to give sagittal support, anteriorly and posteriorly, at the sacral level. The ORLAU Walking Frame (Thompson and Patrick, 1990) (Fig. 1) was designed to provide all of these elements of support combined with a handle to support the arms and provide a means of pushing for forward progression of the frame. Each element is adjustable to accommodate individual patients. It has four wheels and permits an option of castored or

Fig. 1. The ORLAU Walking Frame.

non-castored steering action on the front wheels. The aberrant movements of the cerebral palsy patient to cause significant stability problems. To counter these, low level weights are an option commonly used to enhance the Walking Frame stability. Whilst the Walking Frame was designed as part of an experimental cerebral palsy walking system it has been widely used in other applications where the range of options has proved attractive.

Controlling the aberrant movements of the lower limbs requires an orthosis which crosses the hip joint, knee joint and then ankle joint (HKAFO). Crossing the knee and ankle joint is necessary not only to permit control of those joints but also to ensure there is adequate resistance to the rotational movements of the hip joint. This control is most important since the conflicts between the two limbs arise not only through bilateral hip abduction but also through swing leg hip rotation combined with knee flexion. Experience has shown that the use of a simple thigh cuff in cases where knee and ankle control are not considered necessary is inadequate as the limb is able to rotate within it.

Since the orthosis is theoretically required merely to resist uncontrolled movements it was initially considered that a lightweight structure would suffice. Comparisons with the supportive requirements of the ParaWalker orthosis (Butler and Major, 1987), which is intended to provide walking for thoracic lesion paraplegic patients, initially reinforced the view of lower structural demands necessary in a controlling orthosis. It was recognised there was a need to proceed in an empirical manner, as it would not be possible to measure the aberrant torques during walking without providing the orthotic resistance to these. In the event experience showed, with a number of patients, that the aberrant abductor/adductor torques generated were of the same order of magnitude as those required to provide lateral hip joint support in paraplegic walking devices. An evolutionary approach demonstrated with patients who had progressed beyond the infant stage that structural rigidity rivalling that achieved in the ParaWalker would be necessary to counter the lower limb scissoring which is such an ubiquitous effect in the target group of patients.

The improvement which a control orthosis can provide was observed in a number of early patients. This qualitative impression was reinforced in one patient by monitoring the change in relative energy cost between the patient using the walking frame only and the control orthosis together with the walking frame. Physiological cost index (PCI) was used, in which heart rate and speed are combined to give a reading of heart beats per metre (MacGregor, 1981; Butler *et al.*, 1984). A standard test of five walks of six metres with one minute rest between each was used (Nene and Jennings, 1992) and the result was:

| | |
|---|---|
| walking frame only | 3.8 beats/metre |
| walking frame and control orthosis combined | 1.8 beats/metre |

It can be seen that there was a 52% reduction in PCI (2 beats/metre). When placed in the context of paraplegic walking in the ParaWalker (average 3.82 beats/metre) or RGO (average 5.34 beats/metre) (Bowker *et al.*, 1992) there is an indication of the potential practicality of walking for cerebral palsy patients in combined walking frame and control orthosis systems.

**Variability of biomechanical specifications**

The variability of the detailed biomechanical problems is also an issue which requires to be addressed. Common differences in the required control at hips and knees were found to be:

(i) hip flexion range;
(ii) fixed standing position at the hip;
(iii) knee flexion range;
(iv) knee extension assist.

Until walking has commenced it is usually not possible to identify whether or not these controls are appropriate and ideally an orthosis needs to permit options to be selected during the patient training phase of treatment.

As a result of progressive empirical experience a design specification for a variable specification orthosis (VSO) has been evolved within ORLAU. This is based on the structure of the ParaWalker but incorporates the following variable joint control features:

(i) *Hip joint*
A free or limited range of hip flexion and extension is incorporated. (This can be adjusted to permit a standing stop but then the flexion/extension adjustment is removed). Sitting override is provided in all cases. An optional extra component is available to permit both standing stop and limited hip flexion/extension range.

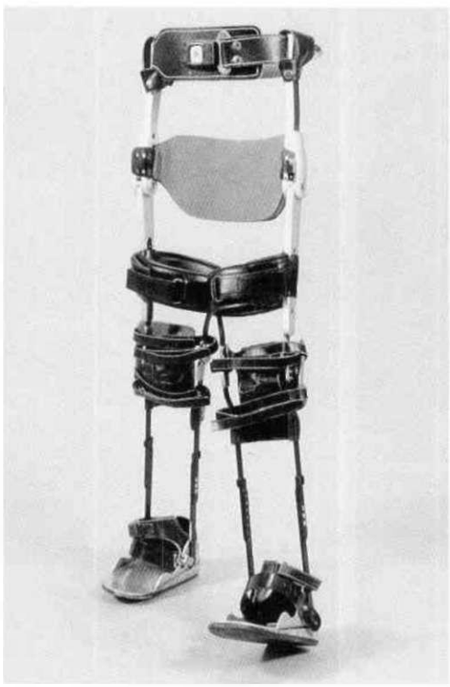

Fig. 2. The ORLAU Variable Specification Orthosis.

(ii) *Knee joint*
A free or limited knee flexion range with knee extension assist as an optional extra feature.

(iii) *Hip and knee joint alignment*
Adjustments to the distance between the hip and knee joint provided to ensure that the anatomical and orthotic joints can be maintained in alignment as the patient grows.

(iv) *Ankle joint*
Fixed with shoe plate enabling standard shoes to be used. Additional coronal plane ankle controls available as optional extras. Quick release attachments to knee-ankle-foot orthosis (KAFO) side members for conventional polypropylene ankle-foot orthosis (AFO) available as an option. (Farmer *et al.*, 1993).

Figure 2 shows the ORLAU Variable Specification Orthosis. Its generic similarities to the ParaWalker will be apparent. However, though there are visually common features, specifically with regard to the structural design, there are important and significant differences in their control functions which renders them non-interchangeable.

**Treatment system**

As with ORLAU ParaWalker (Butler and Major, 1987) it has become obvious that a walking system for severely handicapped patients will require an overall treatment approach if it is to be successfully and appropriately applied. Each patient will require careful orthopaedic and physiotherapy assessment. When a decision to proceed is taken there is a requirement for very accurate patient measurement with particular attention being paid to alignment of anatomical and orthotic joints. When delivered the orthosis requires careful fitting to promote the best possible comfort and function and reduce the risk of tissue damage, and the walking frame needs to be set up to accommodate the particular requirements of the individual. Training the patient also includes optimising the system specification and advising parents, carers and physiotherapists etc.

Once the training phase is successfully completed there is a need for regular on-going assessment at approximately six month intervals.

**Treatment outcomes**

Fourteen patients have been treated in the prototype ORLAU Locomotive Guidance System. Walking was generally restricted to indoor use but the majority of patients were able to walk with only limited supervision. In many cases physiotherapists were able to allow patients to move at will within the confines of the Special School. Some patients used the system at home and there were reports in a few cases of the child assisting with simple chores, such as laying the table.

Commonly patients and parents found the system worthwhile even though the distances walked were relatively small, as might be expected in the environment of home or school. Regular clinical reviews revealed that patients did generally continue to use the system over a period of years, the motivation expressed by parents being that it broadened the scope of activity and improved the outlook of their child. As they got older some of the patients gave up ambulating because it became too inconvenient. However, eight are still using the system and five of these have been doing so for more than five years.

Quite commonly patients used the orthosis for activities other than walking. The additional stability which it provides during sitting permitted some children to concentrate on upper limb activities so that they were able to accomplish particular tasks more speedily. Using the orthosis as a standing device with the hips locked, together with an additional stabilising device (ORLAU, 1990 and 1985), enabled some patients to undertake activities alongside their peers in the school environment.

In a small number of cases patients improved their overall motor control over an extended period during which they were regular users of the system. One child developed an ability to stand independently and then later still take a few steps. A different child was, again after several years' use, able to walk using the walking frame without the orthosis. Clearly it is not possible to ascribe such improvements specifically to the system. However, applying appropriate biomechanics through orthotic intervention has in lower levels of handicap in cerebral palsy patients led to motor learning (Butler and Major, 1992; Butler *et al.*, 1992; Major and Butler, 1995). This would suggest that the improved motor control achieved with some patients could be attributable to the Locomotor Guidance System. At the very least it demonstrates that such walking activity does not interfere with acquisition of motor skills.

**Discussion**

The results achieved with the experimental ORLAU Locomotor Guidance System have been greatly encouraging with the majority of patients who have had the opportunity to use it. However, its use has raised as many questions as it has provided answers to the dilemma of providing walking for an extremely handicapped group.

Careful analysis of the walking patterns adopted by the majority of the patients involved in the trials revealed a number of factors which need to be investigated if the system is to become both more widely applicable and of greater utility to the patients and their families.

An upright truncal posture during walking is an important requirement if appropriate walking patterns are to be stimulated and social interchange during walking is to be promoted. Observation of patients walking in the ORLAU system also revealed that patients require some vertical support if reciprocal lower limb patterning is to be promoted.

Some patients were clearly frustated by the need for their arms to be constrained to operate the walking frame, and steering the system by this method was in many cases an uncomfortable compromise.

It was observed that getting the patient into the complete system of orthosis, then the Walking Frame, was a time consuming and inconvenient activity which made great demands on carers. As the children became older (beyond 8 years) it was not uncommon for three carers to be required for this to be achieved safely.

It has become clear that the existing ORLAU Locomotor Guidance System has successfully addressed the orthotic demands of walking for the heavily handicapped cerebral palsy patient. The structural requirements were found to be much more demanding in the over 8 year old group than had been anticipated. Substantial orthosis strength and stiffness is an essential requirement to ensure both utility and reliability of the system, as is the range of variable options for control because of wide differences in patient motor impairment.

In examining the deficiencies of the walking frame for the target group a review of options was undertaken. Other approaches to walking frames were identified and in particular rear support systems as initially promoted by Motloch (Motloch, 1980; Bleck, 1987) and taken up later by Hart (1990) were seen as potentially suitable arrangements. Superficial examination of these suggests that in their existing form they would not provide sufficiently controlled vertical support or structural rigidity to accommodate the larger patients using the existing ORLAU Walking Frame. It was also clear that transfer of heavy patients into this type of walking frame would be potentially more difficult than with the existing system.

Much encouragement is taken from the results so far achieved. However, further work is now to be undertaken so that the problems of wider practicality can be addressed.

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# Stiffness control in posterior-type plastic ankle-foot orthoses: effect of ankle trimline
# Part 1: a device for measuring ankle moment

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**Abstract**

A device was developed to measure the dorsi- and plantar flexion moment of plastic ankle-foot orthoses when deflected. It is operated by manually controlling a lever which is used to apply a nearly static force. Various orthoses can be classified according to the characteristics of the correcting force measured by this device. Simplicity and high reproducibility are the major advantages. However, to obtain measurements approximating the characteristics of orthoses under wearing conditions its use is restricted to orthoses made of low-viscosity materials.

## Introduction

Plastic ankle-foot orthoses (AFOs) correct abnormal ankle motion by virtue of their stiffness, which defines their orthotic characteristics (Stills, 1975; 1977). Resistance to manual ankle deflection provides a subjective indication of stiffness. The authors have developed an original device which objectively measures orthotic ankle moment.

## Materials and methods

The device consisted of an orthosis-model complex, a pair of metal bars, a tensiometer, a plastic foot enclosure, and a protractor. The orthosis was bent passively with this device and its resistance was an indicator of its stiffness.

A moulded plaster foot and leg model was combined with a plastic AFO so as to permit deformation of orthosis in the same way as when it is worn. The foot model was attached to the frame by a single axis joint at the anatomical ankle axis. The leg model incorporated a pipe which extended from the ankle joint. The foot/leg model was fixed to a plastic AFO with screws and a calf cuff. A vice kept the orthosis-model complex horizontal, so that gravity did not affect the ankle dorsi- or plantar flexion moment.

The 2 metal bars, coupled at the ankle axis, functioned as a lever (Fig. 1). A digital tensiometer was attached to one side bar, so that the loadshaft located on the opposite bar perpendicularly at 0.4 or 0.5m from the axis. This right angled triangle rotated around the ankle axis. When the tensiometer was pulled, the opposite bar pulled the foot through an upright rod in the plastic foot enclosure thus tending to dorsiflex it. The tension force times the lever arm represented the ankle moment applied to the orthosis. A protractor was placed over the metal bars to set the deflection angle at 2.5° intervals (adjustable up to 25°). A screw, which

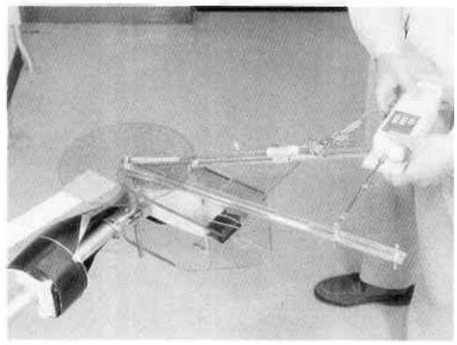

Fig. 1 A whole view of the moment measuring device, during plantar flexing.

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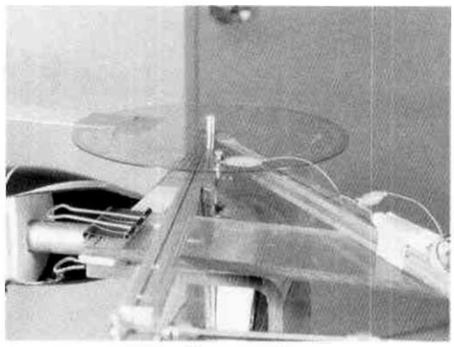

Fig. 2 The touching switch.

penetrated the protractor at 0°, worked as a touch switch (Fig. 2). The opposite bar came in contact with the screw to record the moment value at the preset angle. An experimenter pulled the tensiometer very slowly at an angular velocity of approximately 2°/s so that the opposite bar touched the screw lightly. Thus, the measured value approximated the static moment. The tensiometer had an accuracy of ± 0.2%.

The hingeless plastic AFO deflected mainly in the postero-inferior region when bent in dorsi- or plantar flexion. This caused the leg model to slide along the pipe proximally in plantar flexion and distally in dorsiflexion. Pipe model-interface friction was minimised by using a lubricant.

The whole device was tested for accuracy by measuring the moment of a polypropylene AFO. The test confirmed measurement reproducibility at 4 force levels, equivalent to 0-5Nm, 5-10Nm, 10-20Nm, and 20-30Nm. One experimenter carried out 100 repetitive measurements at each force level, allowing appropriate intervals for the orthosis to completely recover between measurements.

## Results

The maximum moment measured with this device reached 40Nm. This measurable range demonstrated the experimenter's maximum performance, combining force application and velocity control. The lever had to be elongated over 0.5m to apply a moment over 40Nm. The ordinary semi-rigid plastic generated from 30 to 40Nm when plantar flexed 15°.

The ratio of 1 standard deviation to the mean value expressed as a percentage was used as the index of measurement reproducibility. The results were 7.7% at 0-5Nm, 3.9% at 5-10Nm, 2.1% at 10-20Nm, and 1.6% at 20-30Nm level. These low percentages demonstrated the high reproducibility provided by this device.

## Discussion

There are no definitive methods for measuring orthotic load/deflection characteristics. The device described has the advantage of being simpler than methods described previously (Robin *et al.*, 1968; Condie and Meadows, 1977: Rubin and Dixon, 1973; Yamamoto *et al.*, 1993a and b; Miyazaki, 1993). It is simply constructed from readily available materials and involves no complicated mechanisms.

The measurements would reflect the characteristics of the orthosis when worn on the body. Plaster models are incapable of accurately reproducing physical softness and ankle motion. However, they do provide consistent experimental conditions, simplifying the contributory factors.

Measurement errors arise from the friction between the pipe and the leg model, the influence of gravity on ankle motion, and with the manual application of force and control of velocity creating the greatest applied error. However, the high reproducibility provided by this device in the repetitive test confirms that the applied errors can be minimized by having the same person perform all the tests paying the utmost attention to the operation. The protractor rotates if the bar hits the screw, which indicates an incorrect measurement.

Plastic AFOs deflect cyclically during ambulation, and produce a dynamic moment accompanied by material fatigue. Ankle angular velocity and the viscosity of the material have a large influence on dynamic moment. However, the device described is not appropriate for dynamic measurements. The indications for this device should be limited to plastic AFOs made of low-viscosity materials (Lipskin, 1971; Showers and Strunk, 1985), including polypropylene, to minimise the discrepancy between experimental and actual use. Furthermore, comparison of the measured moment should be restricted to static or nearly static conditions.

## Conclusion

This manually operated device permitted the measurement of orthotic moment with a simple mechanism and provided high reproducibility. Although only approximately static values are

measurable, the maximum force range rose to 40Nm with the lever arm length set at 0.5m. The materials of the orthosis should be of low viscosity when this measuring device is used to assess orthotic characteristics during ambulation.

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# Stiffness control in posterior-type plastic ankle-foot orthoses: effect of ankle trimline
# Part 2: orthosis characteristics and orthosis/patient matching

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**Abstract**

The hingeless plastic ankle-foot orthosis (AFO) changes stiffness largely depending on how much plastic is trimmed around the ankle. To support proper selection of the orthosis and final adjustment of the orthotic stiffness, the correlation between the posterior upright width and the resistance to dorsi- and plantar flexion movements was measured in 30 posterior-type plastic AFOs. The posterior upright width was varied by regularly trimming around the ankle in nine stages. The resistance to dorsi- and plantar flexion movements was measured by bending the plastic AFOs 15° with the measuring device described in Part 1. All the plastic AFOs decreased in their resistance to both movements in proportion to the reduction of the posterior upright width. The maximum resistance to plantar flexion movement was about 28 Nm, which was strong enough to assist dorsiflexion in patients with severe spasticity. On the other hand, the maximum resistance to dorsiflexion movement measured was about 10 Nm, which was insufficient to stabilise the ankle in patients who lacked in plantar flexion strength. These findings suggested that this type of plastic AFO should be prescribed for patients who predominantly require dorsiflexion assist, and that the orthotic stiffness could be finally adjusted by trimming to exactly meet individual requirements.

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## Introduction

The plastic ankle-foot orthosis (AFO) assists the swing phase by maintaining the ankle in a neutral position, controlling plantar flexion immediately after heel contact to absorb the impact of body weight, and supporting forward propulsion of the body by stabilising the ankle during terminal stance. It also controls eversion and inversion to provide adequate mediolateral stability.

Plastic AFOs without ankle joint articulations provide these functions in relation to the stiffness of the plastic around the Achilles tendon region. The ankle trimline is the most important among the several factors which affect the stiffness (Stills, 1975; Stills, 1977). Final adjustment of the ankle trimline is needed to meet the individual patient's requirements exactly even with proper selection of orthosis.

There are prescription criteria provided for plastic AFOs without quantitative data to define the final adjustment (LeBlanc, 1973; Lehmann, 1979; Lehneis *et al.*, 1973; Sarno and Lehneis, 1971; Sarno, 1973). The influence of the ankle trimline on orthotic stiffness has been evaluated without consistently regulating the trimming form (Condie and Meadows, 1977; Lehmann *et al.*, 1983; Rubin and Dixon, 1973) The objective of this research was to analyse quantitatively the change in orthotic stiffness corresponding with regulated ankle trimlines, and to advance prescription criteria. The posterior-type of AFO was selected for analysis because of its high frequency of prescription (Ofir and Sell, 1980; Sumiya *et al.*, 1993).

## Materials and methods

*Laboratory experiments*

Two experienced orthotists fabricated 30 posterior type plastic AFOs, 24 for patients and 6 for healthy adults, from 3 mm thick standard grade polypropylene using the vacuum forming technique. The proximal trimline was set 3 cm below the fibular head and the distal trimline was extended to the end of the toes.

The ankle axis was positioned as shown in Figure 1 to serve as a fulcrum for bending the orthosis with a lever. Although this axis did not coincide with the anatomical ankle axis (Isman and Inman, 1969), it was considered from previous test experience to be appropriate. The following opinions support this consideration. The talocrural and subtalar joints act together to create a universal joint-like linkage between the leg and the foot (Wright *et al.*, 1964). However, the orthotic ankle axis allows the talocrural joint alone to move. Accordingly, orthotic and anatomical ankle axes should not be congruent (Kubota, 1981).

Ankle trimlines consisted of circular arcs and their tangents (Fig. 2a). The centres of the arcs were placed on the ankle axis as defined above. The tangents and other straight lines were extended to complete the entire trimline according to the dimensions of the orthosis. The nine different radii, 20%, 25%, 30% . . . , 60% of the lateral malleolus height, provided the nine-stage trimlines.

Endoskeleton below-knee models were prepared for each plastic AFO to be dorsi- and plantar flexed artificially in a manner which resembles actual deformation during walking (Fig. 2b). The leg and foot parts were moulded from plaster and fixed to the plastic AFO with a calf-strap and screws. The leg part slid smoothly along the pipe by using a lubricant.

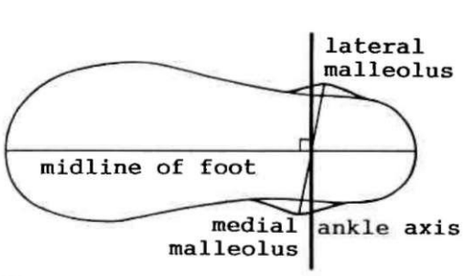


Fig. 1. The location of the ankle axis in a horizontal plane at lateral malleolus height.

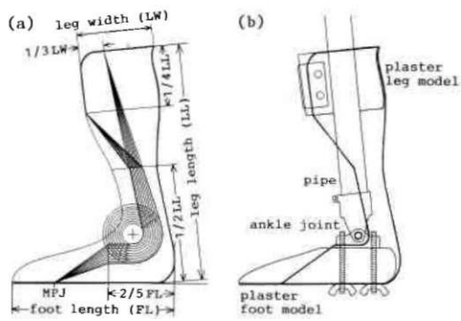


Fig. 2. (a) The nine-stage ankle trimlines.
(b) The endoskeleton below-knee model coupled with the plastic AFO.

The orthosis-model complex was placed horizontally , as described in Part 1, to eliminate the influence of gravity on the ankle movements (Fig. 3). The ankle was dorsi- and plantar flexed 15° at intervals of 2.5°, similar to the normal ankle angle range during walking (Peizer *et al.*, 1969; Stauffer *et al.*, 1977; Sutherland *et al.*, 1980). The ankle movement was measured 10 times at each angle. The orthosis was permitted to recover by leaving appropriate intervals between the measurements.

*Simultaneous clinical assessment*

A 55 year-old male with left sided hemiplegia, one of the 24 patients, wore the nine-stage trimmed orthosis. He had severe spasticity in the affected limbs with limited ankle dorsiflexion range. Careful observation of gait patterns and interview questions were made at each trimline stage to determine the optimal trimline for this particular patient.

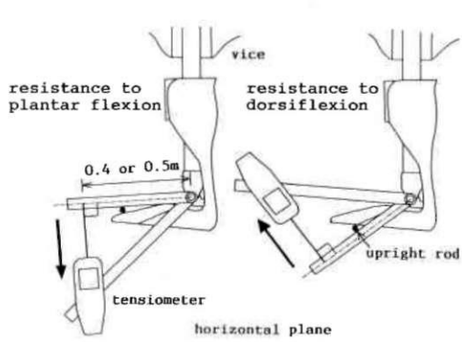


Fig. 3. The system of measurement. The resistance to plantar flexion movement indicated dorsiflexion moment. Similarly, the resistance to dorsiflexion movement represented plantar flexion moment.

## Results

*Laboratory experiments*

A mean of 10 measurements was taken at each deflection throughout the experiments. The ankle moment measured with this device represents an approximately static situation.

The results of tests on 30 plastic AFOs are summarised in Figure 4, displaying the resistance to 5° and 15° of dorsi- and plantar flexion corresponding to each trimline stage. The results showed variation in the stiffness of the plastic AFOs. Both resistance to dorsi- and plantar flexion movements decreased almost in inverse proportion to the trimline stages.

The maximum resistance to plantar flexion movement measured was 27.5 Nm, SD 7.2 Nm when plantar flexed 15°, whereas that to dorsiflexion movement was measured as 10.5 Nm, SD 2.7 Nm when dorsiflexed 15°.

*Clinical assessment*

The subject displayed changes in his gait pattern with changing trimline. Without an orthosis, the stance phase started with toe-contact. With a 20% trimmed orthosis, the stance phase started with heel-contact accompanied by rapid knee flexion. With 30% trimming, plantar flexion occurred immediately after heel-contact and dorsiflexion during the terminal stance became apparent. With 40%, he progressed forward smoothly with the ankle dorsiflexed during the terminal stance. With 50%, he could achieve heel-off just before pre-swing. With 60%, the stride length on the sound side increased, but stability during the stance phase on the affected side decreased and toe-dragging appeared at pre-swing.

## Discussion

The above statements offer the biomechanical grounds for the interpretation of the results.

A locked ankle orthosis provides good toe clearance during the swing phase and spasticity inhibition for hemiplegics (Perry, 1969). However, the rigid plantar flexion stop makes the knee unstable by producing a flexion moment at heel strike. The dorsiflexion assist with spring reduces the knee flexion moment by plantar flexing at heel strike without accelerating spasticity (Lee and Johnston, 1973; Lee and Johnston, 1974). The requirements for dorsiflexion assist for toe clearance during the swing phase and for knee stabilisation at heel strike complement each other. The former should be set at a minimum to permit the latter in flaccid paralysis (Lehmann *et al.*, 1970; Lehmann, 1979; Lehmann *et al.*, 1986). These findings suggest that the orthotic dorsiflexion assist should be minimised such that the swing phase can be carried out safely.

The triceps surae muscle resists dorsiflexion to stabilise the ankle and the knee during the midstance (Perry, 1992; Simon *et al.*, 1978; Sutherland *et al.*, 1980), which contributes to

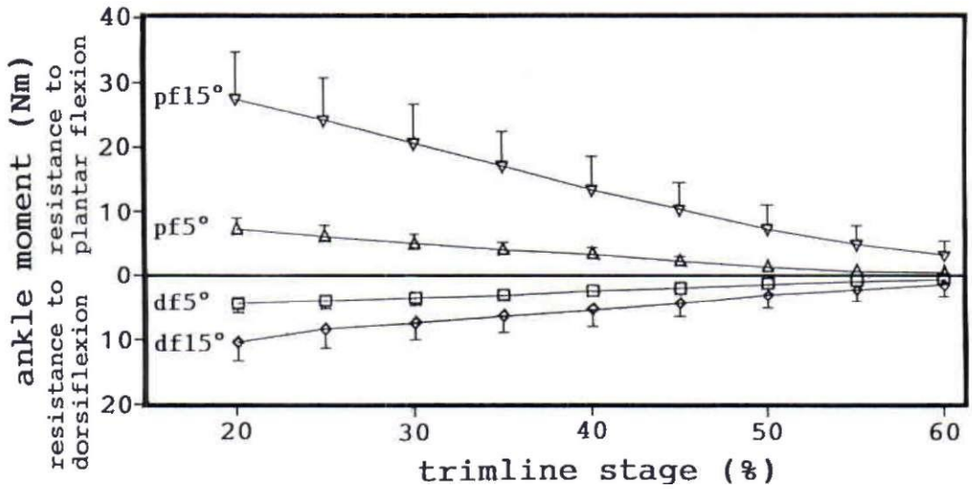


Fig. 4. The correlation between ankle moment and trimline stage. Dots and error bars represent mean and 1 standard deviation of 30 individual plastic AFOs.

forward propulsion of the body during the terminal stance (Brandel, 1976; Dubo *et al.*, 1976; Inman, 1966; Perry, 1974; Winter, 1983). The orthosis with anterior stop successfully substitutes for this muscle function in flaccid paralysis (Lehmann and Delateur *et al.*, 1980; Lehmann *et al.*, 1985; Perry *et al.*, 1995).

On the other hand there is no established indication for plantar flexion assist in spastic paralysis. Hemiplegic patients exhibit weak plantar flexors (Peat *et al.*, 1976). The anterior stop assists them to achieve heel-off, resulting in push-off phase elongation (Lehmann *et al.*, 1987). On the contrary, the hinged plastic AFO with free dorsiflexion reduces spasticity in children with cerebral palsy by stretching the Achilles tendon and saves quadriceps muscle energy consumption (Middleton *et al.*, 1988). Therefore, the orthotic plantar flexion assist should be determined comprehensively on the basis of muscle tone, gait pattern, and energy consumption.

The results can be interpreted based on the above considerations (Fig. 4). Curve-pf15° indicates the dorsiflexion assist at heel-strike for controlled plantar flexion, curve-pf5° curve the moment necessary for toe clearance during the swing phase, curve-df5° the resistance to dorsiflexion during the midstance for knee stabilisation, and curve-df15° the moment opposing free dorsiflexion during the terminal stance. These four requirements must be considered in matching the orthosis to the individual.

The maximum resistance to plantar flexion movement, about 28 Nm in curve-pf15°, is strong enough to control plantar flexion immediately after heel strike in patients with severe spasticity. However, the maximum resistance to dorsiflexion movement, about 10 Nm in curve-df15°, is insufficient to prevent the ankle from breaking down into dorsiflexion during the terminal stance in patients with complete plantar flexor paralysis (Lehmann *et al.*, 1985). In this case, reinforcement of the orthosis will be necessary to provide sufficient ankle support (Clark and Lunsford, 1978; Fillauer, 1981).

The availability of the four moment curves for the final adjustment of the ankle trimline is illustratively demonstrated in the case of the hemiplegic patient (Fig. 5). He required dorsiflexion assist exceeding 1.6 Nm (50% trimline) for toe clearance during the swing phase, but less than 14.4 Nm (40% trimline) for controlled plantar flexion at heel strike. Therefore, the trimlines from 40% to 50% produced the optimal dorsiflexion assist for this patient. On the other hand, he required as much ankle dorsiflexion as possible to transfer the centre of gravity forward during the midstance, overcoming the structural ankle stiffness (Thilman *et al.*, 1991). Consequently, the 50% trimline created the best condition in the posterior-type plastic AFO, and fortunately was a good match. An articulated plastic AFO with free dorsiflexion could possibly replace the

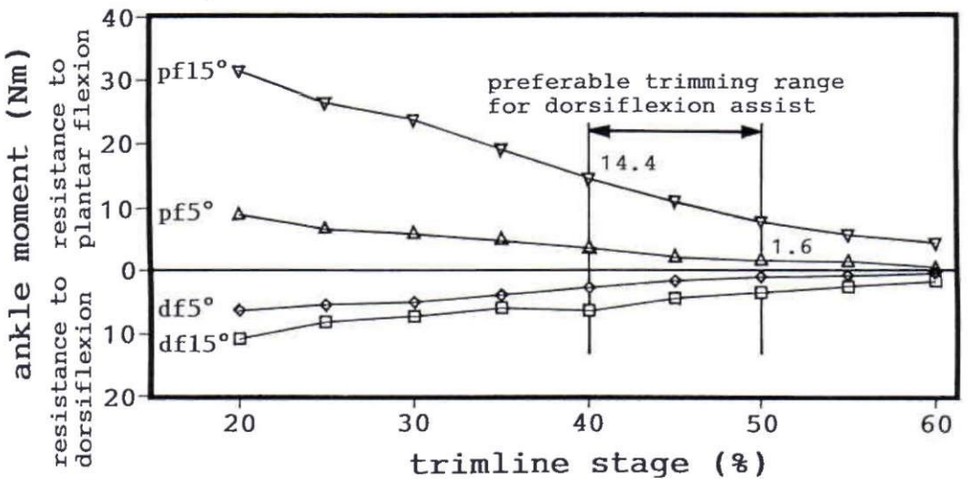


Fig. 5. Clinical application. Data of the hemiplegic patient's plastic AFO are shown.

posterior-type if the same resistance to plantar flexion was available.

## Conclusion

The posterior-type plastic AFO decreased in resistance to dorsi- and plantar flexion movements nearly in proportion to the reduction of posterior upright width. The maximum resistance to plantar flexion movement was sufficient to assist dorsiflexion even under severe spasticity, but that to dorsiflexion movement was only about a third of the former. Accordingly plastic AFOs of this type should be prescribed for patients who predominantly require dorsiflexion assist, and the ankle stiffness must be adjusted by trimming to provide the optimal degree of support.

## Acknowledgements

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# Technical Note

# A CAD CAM digitizing adapter for spinal casts

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**Abstract**

An adapter was designed to hold a negative cast of a torso in position for mechanical digitization. Once a spinal shape is digitized, the data file can be loaded into a CAD CAM system for modification and manufacturing of a positive model for an orthosis. The adapter currently fits a Seattle Digitizer (M+IND, Seattle, WA); however, the interface unit can be changed to accommodate other commercial digitizers. By using this adapter, current lower limb prosthetic CAD CAM systems can be used to design and manufacture spinal orthoses.

## Introduction

Computer Aided Design/Computer Aided Manufacture CAD CAM systems are used in many prosthetic clinics as a manufacturing, research, and/or educational tool (Houston *et al.*, 1992; Krouskop *et al.*, 1989; Lemaire, 1994). While CAD CAM benefits have been shown for prosthetics (Engsberg *et al.*, 1992; Oberg *et al.*, 1993), few orthotic CAD CAM applications have been reported in the literature.

One reason for the lack of orthotic CAD CAM applications could be the difficulty in obtaining the necessary surface topography to produce an orthosis. This is especially true for lower limb orthotics since leg/ankle/foot topography deviates from the conical stump shape encountered in prosthetics. Spinal orthoses, however, have a cylindrical shape which can be accommodated by most prosthetic CAD software packages. Previous literature on CAD CAM and spinal orthosis design involved dedicated spinal cast digitizers to convert a spinal cast shape into a computer data file (Rashke, 1989; Rashke *et al.*, 1990) or used physical measurements to create mathematically an orthosis shape (Ramos *et al.*, 1994).

Physical measurement based systems required custom software to produce the computerized orthotic shape and, while they are currently used as service tools, do not accommodate severe spinal deformities. The cast digitization approach may require a larger custom digitizer for large models; however, existing prosthetic cast digitizers can be adapted to accommodate the majority of spinal orthotic shapes. Since a positive model can be carved out of medium density foam (IPOS carver[1]), orthotic technicians benefit by not having to work with the heavy, plaster spinal cast models during fabrication (foam blanks are approximately 90% lighter than plaster blanks). One contraindication of the foam blanks is that lining materials cannot be stapled on to the foam; therefore, liners cannot be moulded into the orthosis.

To permit spinal cast digitization on a standard prosthetic cast digitizer (Seattle Digitizer, M+IND[2]), an adpater has been developed to hold a spinal cast in position without requiring special modifications to the cast. This Technical Note describes the design criteria, fabrication procedure, and operation of this device.

## Methods

### *Design criteria*

The CAD CAM spinal cast adapter was designed using the following criteria:

- accommodate all spinal cast shapes within the physical limits of the digitizer (58 cm diameter),

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[1] IPOS GmbH & Co, KG Zeppelinstraße 30, 21337 Lüneburg, Germany.

[2] M+IND, 861 Poplar Place South, Seattle, WA 98144, USA.

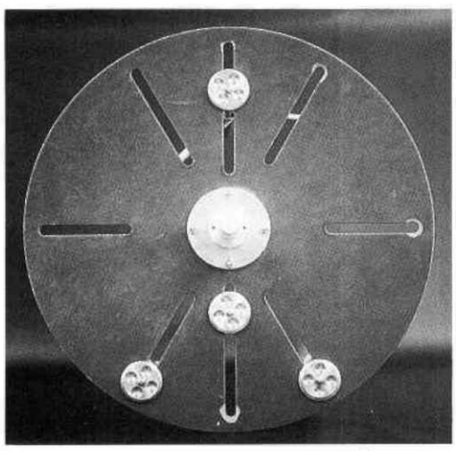

Fig. 1. Inferior view of spinal cast adapter.

- easy cast positioning and digitizer attachment,
- no interference with digitizer operation,
- no cast motion within the adapter frame during digitizing,
- simple device installation and maintenance,
- relatively inexpensive.

### *Components*

The spinal cast adapter has five main components: interface, round table, vertical bars, and attachment arms. The interface section was machined out of aluminium stock to the prosthetic cast holder's dimensions and secured to the bottom of the table with four screws (Fig. 1). Different interface sections could be attached to accommodate different digitizing machines.

Six tracks, 1.3 cm wide were cut in the circular (51.0 cm diameter) table to allow the vertical bars to move radially from the edge of the table (Fig. 2). Although all six tracks are not needed at the same time, the ability to move the vertical bars to a better angular location could help with extremely asymmetrical shapes. The table dimensions were chosen to provide the largest possible diameter without interfering with digitizer operation.

Four aluminium bars 55.0 cm long by 1.6 cm diameter were used to either support the cast or attach the horizontal bars. The vertical bars were threaded at the bottom so that the threaded end could be inserted through a track and secured underneath by a 5.0 cm threaded disk (Fig. 3).

To prevent axial rotation of the bar within the track, a 1.3 cm by 3.0 cm rectangular key was pinned to the mid-point of the threaded region. A 3.2 cm long, threaded cylindrical piece (threaded sleeve) and a 5.0 cm, unthreaded disk were attached at the top side of the threaded region. After the bar had been placed in the track and the lower disk screwed snugly against the key, the cylindrical piece can be tightened against the table by turning the threaded sleeve (thereby securing the vertical bar's position). This "top tightening/loosening" action allows an orthotist to move the vertical bars to the correct position without having to reach under the table to tighten a nut or screw.

The attachment arms were used to hold the cast in a vertical position. The arms consisted of an aluminium rod 15.0 cm by 1.0 cm diameter

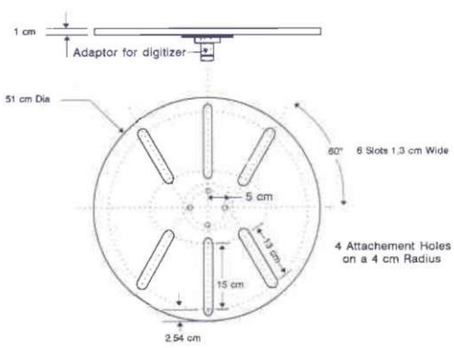


Fig. 2. Base for cast adapter (round table).

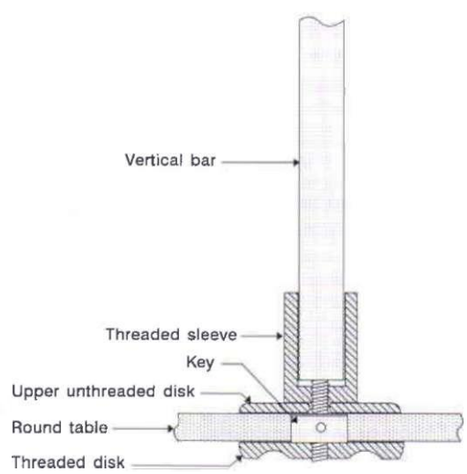


Fig. 3. Vertical bar attachment system.

with a 2.5 by 1.0 cm diameter knurled piece screwed on to the distal end (Fig. 4). This smaller piece was oriented perpendicular to the horizontal bar and was free to rotate about its mid-point to accommodate the cast's contoured surface. An aluminium threaded sleeve was attached to the proximal end of the 15.0 cm rod to secure the attachment arm to the vertical bar.

The locking mechanism is shown in Figure 4. This mechanism simplifies the process of holding the cast in position, moving the attachment arm to the cast's exterior surface, and tightening the arm in position (i.e. hand tightening only, no tools are required to tighten the vertical or horizontal bars). Since the screw section is attached to the bar and the collar, rotating the threaded sleeve will push the locking piece tightly against the horizontal bar.

## Installation – operation

Before installing the cast adapter in the digitizer, the orthotist should ensure that the horizontal bars are in the desired tracks and secured to the table. Orthotists at the Rehabilitation Centre have found that having three horizontal bars on the back tracks and one on the centre front track was appropriate for the majority of spinal casts (Fig. 5).

It is important that the cast be aligned correctly when digitizing to facilitate future CAD modifications. Push the centre-back vertical bar forward until it touches the back of the cast and tighten the bar's locking screw. An attachment arm is not required for this vertical bar since the bar acts as a positioning support for the cast. Move the front vertical bar to approximately 5 cm in front of the cast. Two attachment arms are used on this bar to hold the front of the cast. The two attachment arms are rotated to the left and the right of the bar

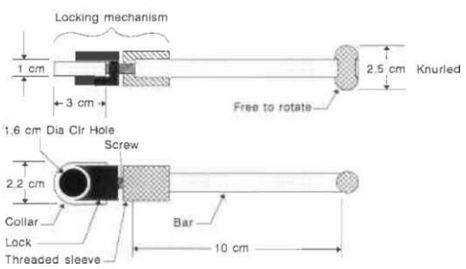


Fig. 4. Horizontal bar design.

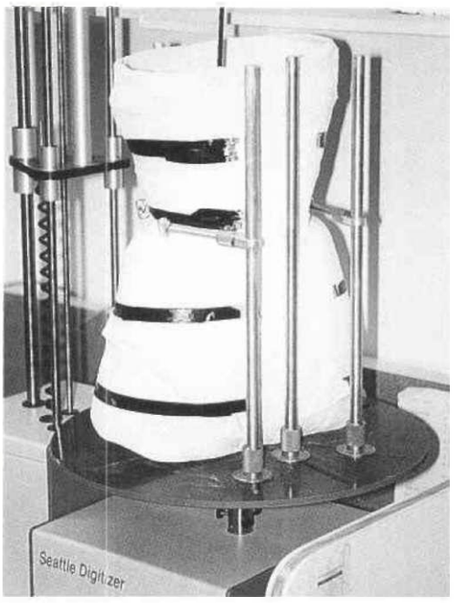


Fig. 5. Spinal cast secured in the digitizing frame.

respectively, pushed tight against the cast, and tightened in place. The remaining two vertical bars are placed 5-8 cm from the sides of the cast. The single attachment arms on these bars are rotated to the side of the cast at a level just above the iliac crests.

During digitizing the clinician or technician may have to slow down the table's angular velocity to ensure that the tracking arm does not skip over high contour areas.

## Acknowledgements

The authors wish to acknowledge the technical assistance of Paul Washer and Harold Gay and the administrative assistance of Guy Martel.

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# Calendar of Events

## National Centre for Training and Education in Prosthetics and Orthotics
## Short Term Courses 1996-97

### Courses for Physicians, Surgeons and Therapists

NC504 Lower Limb Orthotics, 18–22 November, 1996
NC505 Lower Limb Prosthetics, 27–31 January, 1997
NC517 Trans-Tibial Pressure Casting Techniques, 12 February, 1997
NC514 Orthotic Management of the Diabetic Foot, 26-27 February, 1997
NC515 Orthotics and G.P. Fundholders, 25 March, 1997
NC516 Clinical Audit and Outcome Measurement for Lower Limb Amputees, 9 April, 1997
NC511 Clinical Gait Analysis, 10–11 April, 1997
NC510 Wheelchairs and Seating, 15-17 April, 1997
NC506 Fracture Bracing, 28 April–2 May, 1997

### Courses for Prosthetists and Orthotists

NC220 Upper Limb Prosthetics *(in conjuction with BAPO);*
Module 1 20–24 January, 1997
Module 2 19–23 May, 1997
Module 3 13–15 October, 1997
NC219 Orthotic Assessment for Orthotists, 6-7 February, 1997
NC517 Trans-Tibial Pressure Casting Techniques, 12–13 February, 1997
NC218 The TEC Interface System, 10-12 March, 1997

### Course for Orthotists and Therapists

NC217 Ankle-Foot Orthoses for the Management of the Cerebral Palsied Child; 27-29 November, 1996

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

**3-5 September, 1996**
42nd Annual Scientific Conference of the American Paraplegic Society, Las Vegas, USA.
Information: American Paraplegic Society, 75-20 Astoria Boulevard, Jackson Heights, NY 11370-1177, USA.

**9-11 September, 1996**
Instructional Course on Gait Analysis, Dublin, Ireland.
Information: Ms. Ann Jenkinson, Gait Laboratory Manager, CRC, Vernon Ave., Clontarf, Dublin 3, Ireland.

**10-12 September, 1996**
2nd Annual National Conference of the Institution of Physics and Engineering in Medicine and Biology, Leeds, England.
Information: The Secretariat, IPEMB Annual Conference, 4 Campleshon Rd., York YO2 1PE, England.

**12-14 September, 1996**
5th Annual Meeting of the European Society for Movement Analysis in Children, Dublin, Ireland.
Information: ESMAC, Gait Laboratory, CRC, Vernon Ave., Clontarf, Dublin 3, Ireland.

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

**18-21 September, 1996**
50th Annual Meeting of the American Academy for Cerebral Palsy and Development Medicine, Minneapolis, USA.
Information: AACPDM, 6300 N.River Rd., Suite 727, Rosemont, IL 60018, USA.

**3-6 October, 1996**
Annual Meeting of the Biomedical Engineering Society, Pennsylvania.
Information: Rita Kline, Bioengineering Program, Pennsylvania State University, 205 Hallowell Building, University Park, PA 16802-6804, USA.

**10-14 October, 1996**
73rd Annual Meeting of the American Congress of Rehabilitation Medicine, Chicago, USA.
Information: American Congress of Rehabilitation Medicine, 4700 West Lake Ave., Glenview, IL 60025, USA.

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

**14-19 October, 1996**
Eurospine '96, Zurich Switzerland.
Information: ESS Secretariat, J Ricchert Schild, c/o Schaltess Klinik, Legghalde 2, CH-8008 Zurich, Switzerland.

**21-23 October, 1996**
International Conference on Quality of Life and Assistive Technologies, Montreal, Canada.
Information: Denise Mauger, 6300 Darlington Ave., Montreal, Quebec HS3 2J4, Canada.

**29 October-1 November, 1996**
American Orthotic and Prosthetic Association National Assembly, Cincinnati, USA.
Information: AOPA, 1650 King St., Suite 500, Alexandria, VA22314, USA.

**31 October-3 November, 1996**
18th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, Amsterdam, The Netherlands.
Information: Wim Rutten, Program Co-chair, Dept. of Biomedical Engineering, Faculty of Electrical Engineering, University of Twente, PO Box 217, 7500 AE Enschede, The Netherlands.

**14-16 November, 1996**
III Internacional Técnicas Ortoprotésicas, Madrid, Spain.
Information: Orto-96, San Mateo 20, 28004 Madrid, Spain.

**3-6 December, 1996**
International Congress on "People and Health" — Traumatology, Orthopaedics, Prosthetics and Rehabilitation.
Information: Congress Organising Committee, PO Box 204, St Petersburg, Russia, 191025.

**1997**

**13-18 February, 1997**
American Academy of Orthopaedic Surgeons Annual Convention, San Francisco, USA.
Information: AAOS, 1650 King St., Suite 500, Alexandria, VA 22314, USA.

**8-9 March, 1997**
Annual Scientific Meeting of British Association of Prosthetists and Orthotists, Blackpool, England.
Information: The Secretariat, BAPO, Dunoon and District General Hospital, Dunoon, Argyll PA23 7RL, Scotland.

**17-20 March, 1997**
1st Gulf Congress in Medical Rehabilitation, Kuwait.
Information: Congress Secretariat, Physical Medicine and Rehabilitation Hospital, PO Box 4079, 13041, Kuwait.

**11-14 April, 1997**
ISPO Update Course on Amputation and Prosthetics, Helsingborg, Sweden.
Information: May-Christine Friberg, Department of Orthopaedics, Helsingborg Hospital, S-251 87 Helsingborg, Sweden.

**6-9 May, 1997**
Orthopaedic and Rehabilitation Technology Trade Fair and World Congress, Nuremberg, Germany.
Information: Bundesinnungsverband fur Orthopadie-Technik, PO Box 10 06 51, D-4406 Dortmund, Germany.

**31 August-5 September, 1997**
8th World Congress of the International Rehabilitation Medicine Association, Kyoto, Japan.
Information: Japan Convention Services Inc., Nippon Press Center Bldg., 2-1, 2-chome, Uchisaiwai-cho, Chiyoda-ku, Tokyo 100, Japan.

**8-12 September, 1997**
Dundee 97: International Conference on Wheelchairs and Seating, Dundee, Scotland.
Information: The Secretariat, Dundee 97, Dundee Limb Fitting Centre, 133 Queen St., Broughty Ferry, Dundee DD5 1AG, Scotland.

**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.

**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.

**1998**

**28 June-3 July, 1998**
9th World Congress of the International Society for Prosthetics and Orthotics, Amsterdam, The Netherlands.
Information: Congrex (Holland) B.V., Keizersgracht 782, 1017 EC Amsterdam, The Netherlands.
Tel: +3120 626 13 72. Fax: +3120 625 95 74.