Accelerations due to impact at heel strike using below-knee prosthesis

H. W. L. VAN JAARSVELD*, H. J. GROOTENBOER* and J. DE VRIES†

University of Twente, Department of Mechanical Engineering, Netherlands
†Rehabilitation Centre, "Het Roessingh", Netherlands

Abstract

The acceleration in the sagittal plane of the prosthetic tube at heel strike in normal walking was measured in five healthy amputees with their definitive below-knee prosthesis, every subject using six different prosthetic feet, wearing sport shoes as well as leather shoes. The experiments were carried out in the rehabilitation centre "Het Roessingh", Enschede, The Netherlands.

Maximum accelerations were extracted from the acceleration-time-signal. Mean acceleration maxima of all subjects were calculated for each foot-shoe combination to eliminate the individual influence of the subjects. In the axial direction the maximal accelerations demonstrate a clear difference among the prosthetic feet and the shoes, while in dorsoventral (tangential) direction the inter-individual variation in the acceleration extremes dominates the difference between the types of footwear. In comparison with non-amputees the magnitude of the maximal axial acceleration at heel strike does not differ significantly.

Introduction

One of the important aspects in the design and selection of prosthetic feet is the comfort they offer the user in walking.

An item of comfort is certainly the absorption of high level accelerations (in fact negative acceleration, but further referred to as just acceleration), induced by the impact between (shoe) heel and ground at the end of the swing phase. Without sufficient damping these high accelerations can easily lead to an overload of the biological tissues of the stump and the skeletal joints. To evaluate the differences in absorption of high level accelerations among commercially available prosthetic feet and the influence of the

All correspondence to be addressed to H. W. L. Van Jaarsveld, Dept. of Mechanical Engineering, University of Twente, PO Box 217, 7500 AE Enschende, Netherlands.

shoe type on these differences, an experiment was designed in which these accelerations at heel strike were measured.

In a similar experiment as reported here, but with non-amputees Van Leeuwen et al. (1988) found in the tibia a peak acceleration at heel strike of $40 \text{m} \cdot \text{s}^{-2}$ in the axial direction of the tibia and of $80 \text{m} \cdot \text{s}^{-2}$ in the direction perpendicular to the tibia, while in the shoe sole these peak accelerations were $600 \text{m} \cdot \text{s}^{-2}$ and $150 \text{m} \cdot \text{s}^{-2}$ respectively. From these figures it is clear that for healthy subjects there is a great amount of shock absorption in the foot-ankle complex, expecially in the axial direction of the leg. The knee, hip and other skeletal joints take care of peak acceleration absorption as well. Mizrahi and Susak (1982) found that at impact with one straight leg, after fall from 50 mm, there only remains a maximum acceleration of $46 \text{m} \cdot \text{s}^{-2}$ at the level of the greater trochanter.

Johnson (1988) investigated the influence of shoe insoles on peak acceleration absorption. He defines a shock factor as the integral of the amplitude spectrum from 50 to 150Hz divided by the integral from 10 to 150Hz. Due to the use of an insole this shock reduction between 8% and 39% can be achieved with respect to a shock factor.

Method

Five male subjects with an unilateral below-knee amputation, of good physical condition and experienced in prosthetic walking, were each provided alternately with six different prosthetic feet. Information on the subjects and feet is presented in Tables 1 and 2. After a weekly period of habituation to each new prosthetic foot, the volunteers were invited to the gait laboratory where accelerometers were fixed to the prosthetic tube just above the foot-tube connection. Gait parameters as well as the axial and tangential acceleration were registered while walking a distance of 2 times 10m at a comfortable speed. The experiments were performed with two types

Table 1. Information on the below-knee amputees who co-operated with the experiment.

subject	age (Yrs)	body weight (kg)	initial foot		amputated since (Yrs)	walking speed (m/s)
1	59	102	Bock	Dynamic	21	1.33
2	52	75	″	Lager	44	1.78
3	23	—	″	″	—	1.44
4	32	92	″	″	10	1.32
5	30	88	″	″	10	1.26

Table 2. Information on the prosthetic feet.

1. Otto Bock SACH foot
2. Otto Bock dynamic foot
3. Otto Bock uniaxial foot
4. Rax foot
5. Seattle foot
6. Blatchford Endolite Multiflex foot (medium stiffness)

of shoes: a pair of normal leather shoes and a pair of flexible sports shoes.

The accelerations were registered with Bruël & Kjoer accelerometers Type 4375 for the axial acceleration and Type 4393 for the acceleration perpendicular to the tube axes in the sagittal plane. Both accelerometers had a sensititivy of $0.3pC/m \cdot s^{-2}$. The signal was amplified to $0.1V/m \cdot s^{-2}$ and with the aid of an analogue/digital-converter and an IBM-XT compatible computer sampled with a frequency of 2000Hz.

Each walking track of 10m included approximately 6 prosthetic heel strikes. Due to occasional wire fractures, resulting in a constant measuring-signal, some measurements had to be rejected afterwards as being incorrect.

Data analysis

In the experiment no use was made of foot-ground contact detectors to eliminate any disturbance from these devices. The moment of heel strike was accurately extracted from the registered acceleration signal. A reliable way of automatic detection of this moment was to determine the point where the absolute value of the first time derivative of this signal exceeded the value $1 \cdot 10^4 m \cdot s^{-3}$ (Fig. 1).

The first extreme acceleration after heel strike was determined. For each experiment the mean
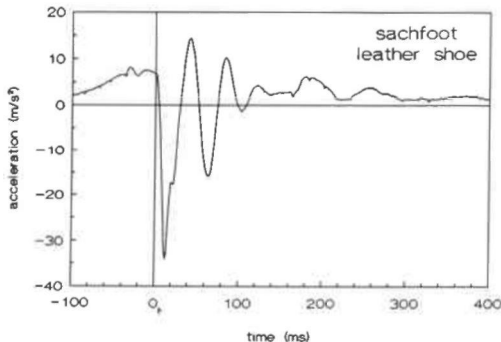
and variation of these peak accelerations were calculated. Due to the inter-individual differences in the gait patterns the peak accelerations per subject were normalised before calculating the mean of all five test subjects. This normalization was achieved by dividing for each subject the axial and tangential peak accelerations by the axial peak acceleration of walking with an Endolite foot with sports shoe and the tangential peak

Fig. 1. Axial acceleration around heel strike (0 msec.) of subject 1 wearing a SACH foot and leather sole.
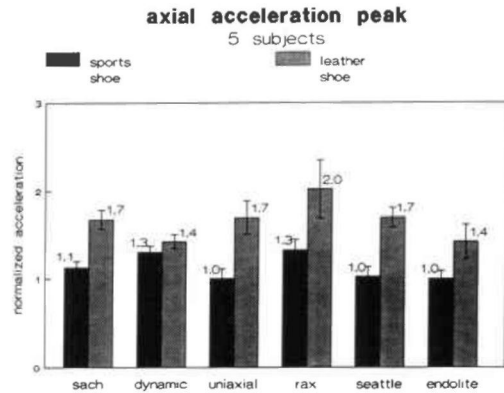
(a)
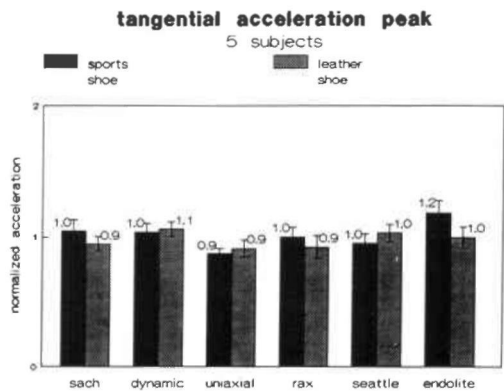
(b)

Fig. 2. Mean normalized acceleration maxima in axial (a) and dorso-ventral (b) direction ± 90% confidence interval.

acceleration of walking with an Endolite foot with leather shoe respectively.

Results

Figures 2a and 2b show the mean maximum axial and tangential accelerations, with a 90% confidence interval according to the Student-t-test, of all five subjects per foot design, wearing leather and sports shoes respectively.

It can be seen in Figure 2b that the influence of foot design and shoe stiffness on the peak tangential acceleration is not significant. The influence of these factors on the peak axial acceleration is larger (Figure 2a), although the differences between the mean accelerations are moderate. For the axial acceleration the added flexibility of the sports shoe leads in all cases to a reduction of the peak acceleration, a reduction which is most pronounced with the stiffer feet like the Rax.

The inter-individual differences were very large as can be seen in Figure 3, the mean maximal

axial and tangential accelerations are depicted for each subject, foot design and shoe. These large differences are caused by variation in weight, length, muscle strength and other physical properties.

The assumption is made that the walking patterns of the subject are constant in all experiments so differences for one subject are only caused by the variations in foot type. Small variations in gait throughout the experiments will occur due to physical and mental coincidences, but these influences can be largely eliminated by taking the mean of a reasonably large sample.

Discussion

The declination of peak accelerations at heel strike is an important aspect of the foot-ankle performance. The choice of the foot type influences the magnitude of the accelerations in the axial direction of the prosthetic tube at heel strike, leaving the tangential accelerations
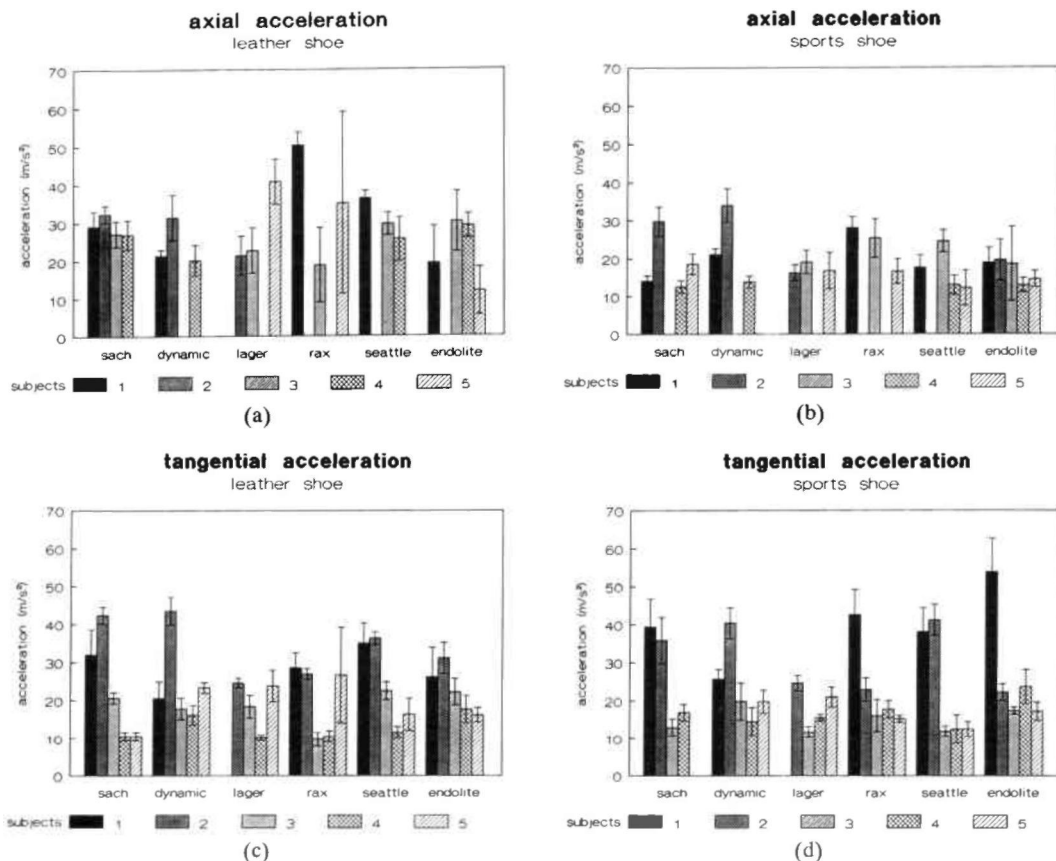

Fig. 3. Individual acceleration maxima in axial (a & b) and dorso-ventral direction (c & d) of all subjects wearing leather (a & c) and sports (b & d) shoes ± 90% confidence interval.

practically unaffected. A significant reduction of the maximal axial acceleration can be achieved by wearing sports shoes instead of normal rigid leather shoes.

The registered axial accelerations of the prosthetic tube of the unilateral below-knee amputees ranges from 20 to $50 \text{m} \cdot \text{s}^{-2}$ which is comparable with the $40 \text{m} \cdot \text{s}^{-2}$ for healthy subjects, as measured by Van Leeuwen et al. (1988). In tangential direction, however, the accelerations were lower in prosthetic walking ($10 - 40 \text{m} \cdot \text{s}^{-2}$) than in normal walking (about $80 \text{m} \cdot \text{s}^{-2}$).

REFERENCES

Johnson, G. R. (1988). The effectiveness of shock-absorbing insoles during normal walking. *Prosthet. Orthot. Int.* **12**, 91−95.

Leeuwen, J. L., Kahmann, L. R. M., Jans, H. W. J. (1988). An experimental approach to shock absorption of the human foot during heel strike. *Eur. Soc. Biomech., September* 11−15.

Michael, J. W. (1987). Energy storing feet: a clinical comparison. *Clin. Prosthet. Orthot.* **11**, 154−168.

Mizrahi, J., Susak, Z. (1982). In-vivo elastic and damping response of the human leg to impact forces. Trans ASME. *J. Biomech. Eng.* **104**, 63−66.