Prosthetics and Orthotics International, 1990, 14, 117–124

Stiffness and hysteresis properties of some prosthetic feet

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

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

Abstract

A prosthetic foot is an important element of a prosthesis, although it is not always fully recognized that the properties of the foot, along with the prosthetic knee joint and the socket, are in part responsible for the stability and metabolic energy cost during walking.

The stiffness and the hysteresis, which are the topics of this paper, are not properly prescribed, but could be adapted to improve the prosthetic walking performance. The shape is strongly related to the cosmetic appearance and so can not be altered to effect these improvements. Because detailed comparable data on foot stiffness and hysteresis, which are necessary to quantify the differences between different types of feet, are absent in literature, these properties were measured by the authors in a laboratory setup for nine different prosthetic feet, bare and with two different shoes. One test cycle consisted of measurements of load deformation curves in 66 positions, representing the range from heel strike to toe-off.

The hysteresis is defined by the energy loss as a part of the total deformation energy. Without shoes significant differences in hysteresis between the feet exist, while with sport shoes the differences in hysteresis between the feet vanish for the most part. Applying a leather shoe leads to an increase of hysteresis loss for all tested feet.

The stiffness turned out to be non-constant, so mean stiffness is used. Because very little is known about the optimal values of stiffness and hysteresis, and substantial differences in stiffness between different feet and shoes exist, further investigation into the importance of stiffness and hysteresis to the walking quality of a foot is necessary. Footwear counts too for this quality because it modifies the variation in stiffness among the feet.

All correspondece to be sent to Prof. dr. ir. H. J. Grootenboer, Lab. for Biomechanics, Faculty of Mechanical Engineering, University of Twente, P.O. Box 217, 7500 AE Enschede, The Netherlands.

Introduction

The influence of the mechanical properties of the prosthetic foot on different aspects of gait is not yet fully understood. In conjunction with the prosthetic knee joint and socket, two important mechanical conditions are to be fulfilled:

- the prosthesis has to support the body with maximal stability during the stance phase, which means for example that the resultant ground reaction force has to pass in front of the instantaneous centre of rotation of the knee joint.
- walking with a prosthesis has to demand as little energy as possible.

Four mechanical properties of the foot influence the stability and energy consumption and affect of the roll-over behaviour of the foot:

- the shape and the alignment of the foot, along with the pylon angle, determine the point of application of the ground reaction force on the foot. The shape also influences the vertical and horizontal movement of prosthesis and body during gait, as is shown by Koopman (1989). Foot shape is not considered in this paper.
- the mass and mass-distribution of the foot affect the swing behaviour of the leg (Van de Veen, 1989). Donn *et al.* (1989) showed in an experimental study that an optimal choice of the mass can significantly improve some symmetry coefficients of walking. The mass-

distribution will not be considered here.

- the stiffness determines the foot deformation during weight bearing and therefore affects the foot shape. It may be important for energy storage and release during the progress of weight bearing since a soft foot can store more energy than a stiff foot when the same load is applied.
- hysteresis related to stiffness is a pure energy issue and represents an energy loss due to internal friction when loading and unloading a deformable object. Minimizing this hysteresis of the foot is an easy and sure way to decrease the energy cost of walking provided that the stored energy is indeed returned in a profitable way.

The choice of a stiffness grade depends on the body weight and activity level of the amputee and is mostly restricted to the heel grade of one foot type. How different foot types differ in stiffness is unknown, while just through the increasing number of foot types it is necessary to know more about the particular advantages of different feet. The stiffness of a SACH foot is prescribed by the Veterans Administration Prosthetics Center (Daher, 1975).

The goal of this investigation is to measure the foot stiffness and hysteresis of the nine prosthetic feet which are listed in Table 1. For this purpose the feet are tested in a 3-D stiffness measuring device so, as opposed to a clinical test (e.g. Michael, 1987), a good reproducibility is achieved. The use of a measuring device also allows for a more objective qualitative comparison of different prosthetic feet than in clinical tests such as those performed by Winter and Sienko (1988) and Ehara *et al.* (1990). In these tests Ehara found considerable differences in the energy storage of 12 different prosthetic feet, while Winter found differences of 50% between a SACH and a Greissinger foot. To examine the effect of footwear on the mechanical properties of the feet, the same measurements were performed with a leather shoe and a sport shoe. Thorough stiffness and durability tests have been carried out by Daher (1975) and Skinner *et al.* (1985), but only with some SACH feet.

Table 1. The names of the tested feet with abbreviations.

1. Otto Bock dynamic	dy
2. Otto Bock uni-axial	un
3. Otto Bock SACH	sa
4. Blatchford Endolite Multiflex (medium stiffness)	mu
5. Hanger Quantum (50 to 70kg)	qu
6. Rax	ra
7. Ipos titanium spring rigid ankle	itr
8. Ipos carbon reinforced plastic spring rigid ankle	icr
9. Ipos carbon reinforced plastic spring flexible ankle	icf

The data obtained are not completely representative of the behaviour during gait but are especially useful for comparisons of several feet. Differences between practice and experiment are the loading speed and direction.

Application in prosthetic design of the principle that energy can be stored in an elastic element to be used later on for mechanical work is not new. Voisin (1987) designed a foot with two helical steel springs mounted in the sagittal plane between two plates and claimed an improvement in the energy restoring property of his D.A.S. foot. Also new foot designs using materials such as carbon reinforced plastics have been presented, like the Hanger and Ipos feet. To examine the energy restoring capacities of prosthetic feet, Michael (1987) did clinical tests with some older types like the SACH foot and some new feet like the Seattle, Carbon Copy II and Flex-Foot. The experiments were done with the use of a pogo stick with one of the feet mounted at the end. Michael (1987) used the maximum height achieved by the same person after ten hops as the comparative value, where the Flex-Foot turned out to be the best in returning energy. However, in this way the feet are tested in only one position and the reproducibility may not be very good.

A recent study of Ehara *et al.* (1990) showed considerable differences between 12 prosthetic feet in energy storage and release during walking.

Methods

The 3-D stiffness measuring device consists of a stiff rectangular frame, instrumented with 6 carriages, controlled by step-motors (Fig. 1). The prosthetic foot is mounted upside-down in the bottom part of the frame where five carriages are able to perform the horizontal ($x-$ and $y-$) translations and the three ($x-$, $y-$ and $z-$) rotations of the foot. A stiff horizontal aluminium plate, representing the floor, is mounted at the upper part of the frame and can be translated in the vertical ($z-$)direction by the last carriage thus applying a load to the foot.

To include the range which occurs during walking, the angle between the pylon and the

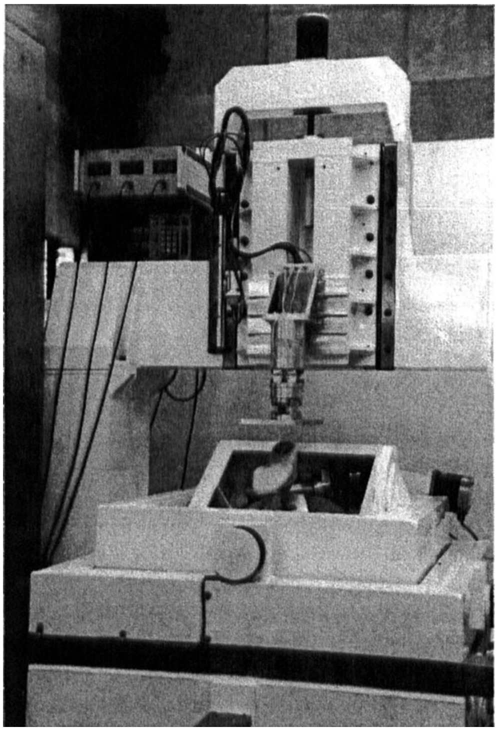

Fig. 1. The set-up of the test rig.

vertical (y–rotation) is varied from −30° (heel strike position) towards 35° (toe-off position) by increments of one degree. In the practice of prosthetic walking there is relatively little eversion or inversion, which justifies the experimental restriction to a two dimensional measurement. In all 66 positions a horizontal plate representing the floor is pushed down on the foot in stages of 1mm until a vertical force of 1000N or 35mm deformation of the foot is achieved (whichever occurs first), after which the procedure is reversed with decreasing stages of deformation. At each stage the vertical and horizontal force between foot and plate is registered.

When the horizontal force is too large, a slippage may occur between the foot and the plate. To prevent this slippage, the horizontal force is decreased in each stage by moving the plate in a horizontal direction whenever the horizontal force exceeds a value of 0.3 times the vertical force. This friction coefficient was chosen after initial experiments with the measuring device. The horizontal corrections are especially needed near the heel strike and toe-off positions.

Five measurements were carried out on the icf-foot to identify the repeatability, and the velocity influence was tested.

All feet were tested without footwear, with a leather shoe and with a sports shoe.

Data analysis

Two force-displacement curves are shown in Figure 2: in Figure 2a position −30° and in Figure 2b position 35°. The appearance of a hardening spring-like behaviour and the hysteresis loop are revealed at first glance.

To reduce the data two fourth grade polynomials are fitted on the force-displacement curves for loading and unloading. The irregularities in the force-displacement curve in the vertical direction are caused by the horizontal

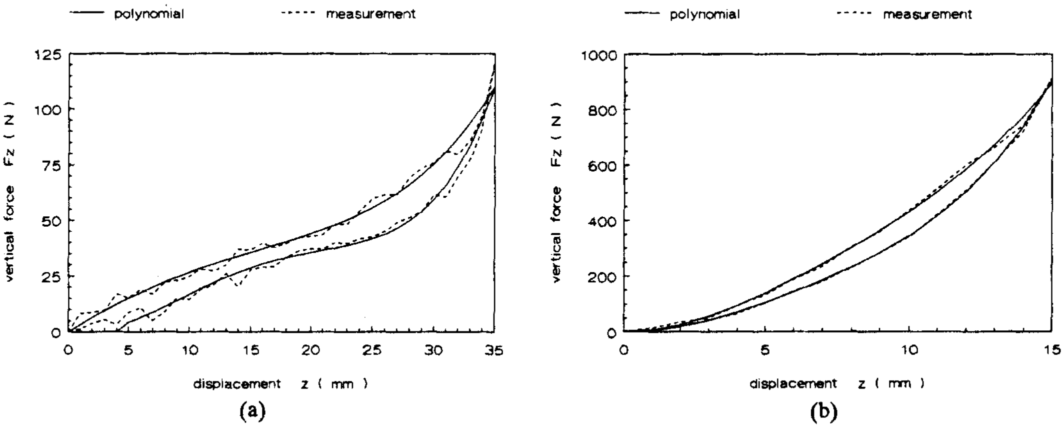

Fig. 2. Typical load-displacement curves showing the hardening spring behaviour. Auxilary is the best fitted polynomial according to the least square method. The depicted measurements are heel strike (a) and toe-off (b) from the Hanger Quantum foot without shoe.

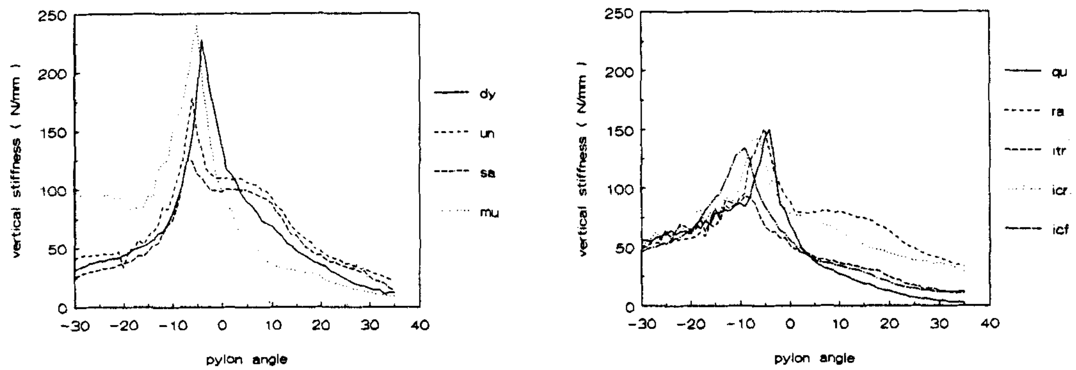

Fig. 3. Vertical secant stiffness of all feet without shoe (abbreviations see Table 1).

displacements of the plate (Figure 2). The curve fitting smooths these irregularities and results in the functions $f_i(z)$ and $f_d(z)$, where i stands for increasing loads and d for decreasing loads. The polynomials cross the z-axis at $z = 0$ for f_i and $z = z_0$ for f_d, where z_0 is a positive real value.

The stiffness depends on the displacement (z) so differentiation of $f_i(z)$ yields the rate of change of stiffness in the vertical direction as a function of z, resulting in 66 stiffness curves per foot. To further condense the data the mean stiffness only is presented as a function of the pylon angle. This mean stiffness is calculated at maximum load or maximum deformation.

The hysteresis is derived as a function of foot inclination from the loading and unloading curves and is the energy loss as a part of the total deformation energy under increasing loads.

Results and discussion

Stiffness (Figs. 3, 4 and 5)

Stiffness data will be discussed on a relative basis because a reference is not available. The foot flat position will be defined as the position where the maximal stiffness occurs, that is where the heel and the forefoot are equally loaded.

A high stiffness is found from heel strike to foot flat position for the Multiflex foot. The elastic material concentrated in the ankle device is apparently very stiff when the ankle deflects in the plantar direction. At foot flat position the Otto Bock dynamic foot is extremely stiff. The Ipos titanium rigid ankle foot is weak due to its soft rubber heel and weak titanium spring blade. At toe-off position the feet with a spring blade (the Hanger and all Ipos feet) are the most flexible with the exception of the Ipos carbon rigid ankle foot.

The influence of the presence and the type of the footwear is quite obvious. The point of maximal stiffness rotates about 8° forward for the leather shoe; for the sport shoe the point of maximum stiffness lies around −5°. The value of the maximum stiffness increases by about 50N/mm by adding a leather shoe except for the Otto Bock uni-axial foot where it increases by 180N/mm and the Hanger and dynamic foot where the

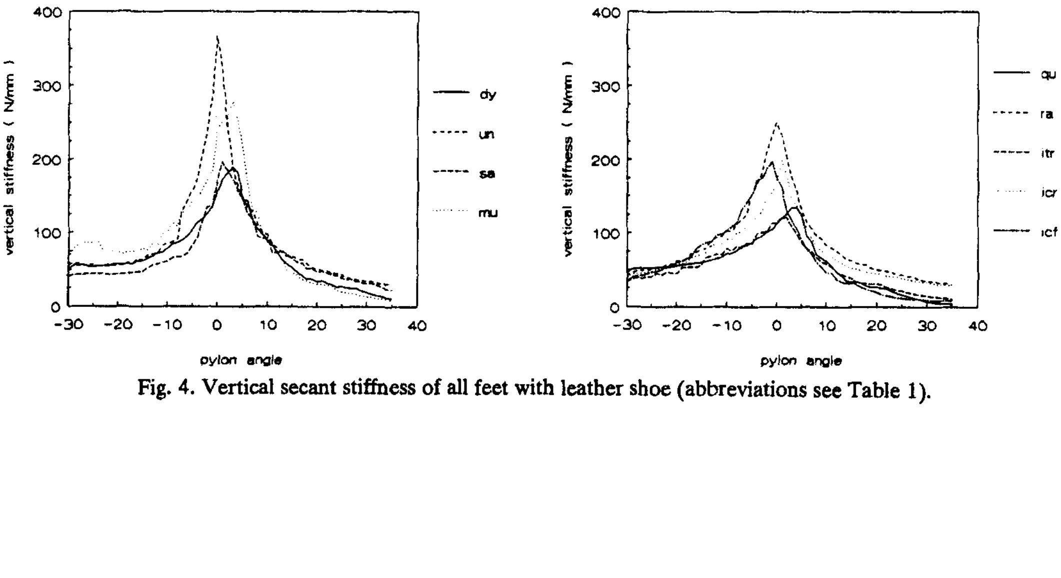

Fig. 4. Vertical secant stiffness of all feet with leather shoe (abbreviations see Table 1).

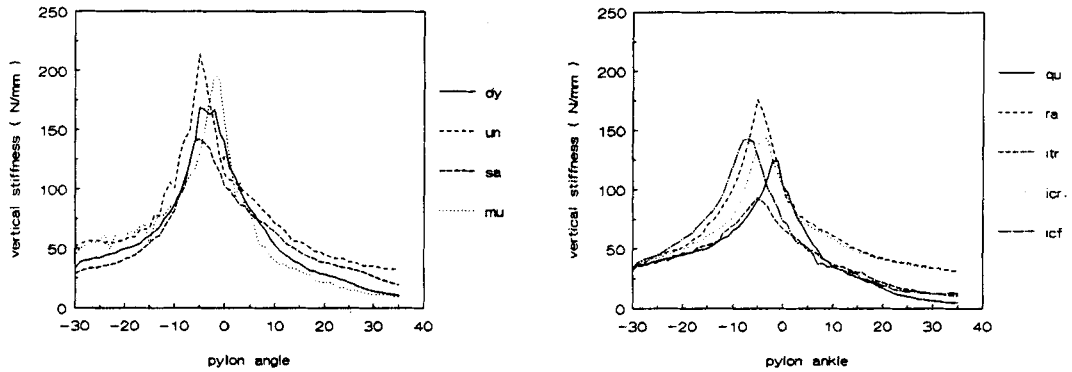

Fig. 5. Vertical secant stiffness of all feet with sport shoe (abbreviations see Table 1).

maximum stiffness decreases by 30N/mm. The maximum stiffness of the feet is less influenced by a sport shoe.

Hysteresis (Figs. 6, 7 and 8)

In contrast to the other properties of the foot such as stiffness or shape, the hysteresis can be directly interpreted. The smaller hysteresis loop the better, because the absorbed energy depends on the area of this loop and it is to be expected that a low value of hysteresis will reduce the energy needed for walking.

In general, rubber shows less hysteresis loss when deformed below 100% than at more deformation (Powell, 1983). Applying this to the authors' measurements explains why at foot flat position, where stiffness is high, and deformation is low, hysteresis is at a minimum.

Comparing some familiar feet like the three Otto Bock feet, it is conspicuous that the hysteresis characteristics are quite similar, except that in the case of the uni-axial foot with the cylindrical rubber plantar flexion stop, which is active at heel strike, it appears that the rubber stop absorbs more energy than the soft heels of the SACH and dynamic foot.

Comparison of the three Ipos variations leads to the logical conclusion that they do not differ significantly at foot flat because the inserted spring blades in the forefoot are not active and neither is the flexible ankle device. In the area where the spring is in action (angles larger than −10°) a significant difference can be seen between the foot with the titanium and the carbon reinforced polymer spring. The second foot restores more energy. Comparing the Ipos rigid ankle with the Ipos flexible ankle, the conclusion can be drawn that the flexible ankle absorbs much more energy in all positions except foot flat.

The Quantum foot absorbs hardly any energy in these experiments, partly because at heel strike a highly elastic spring is active in contrast to the Ipos feet where at the heel strike position deformation of soft rubber results in high hysteresis. The Multiflex foot derives most of its flexibility from the rubber rings in the flexible ankle device and has the worst energy restoring

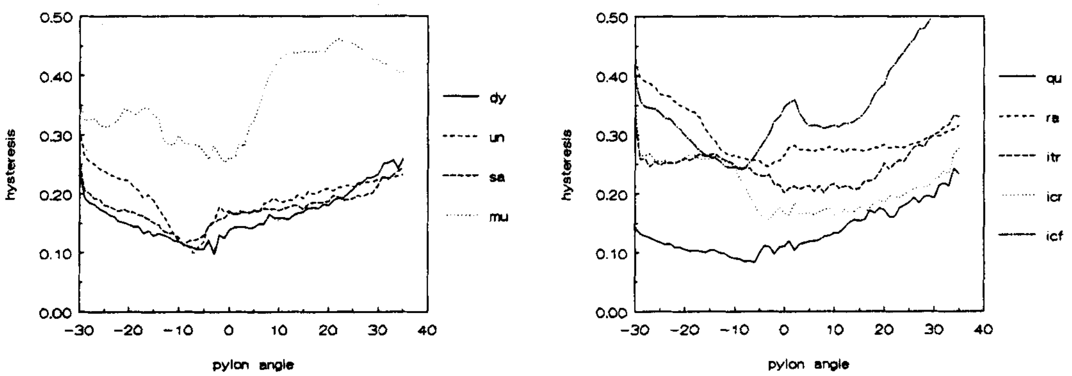

Fig. 6. Hysteresis data of all feet without shoe (abbreviations see Table 1).

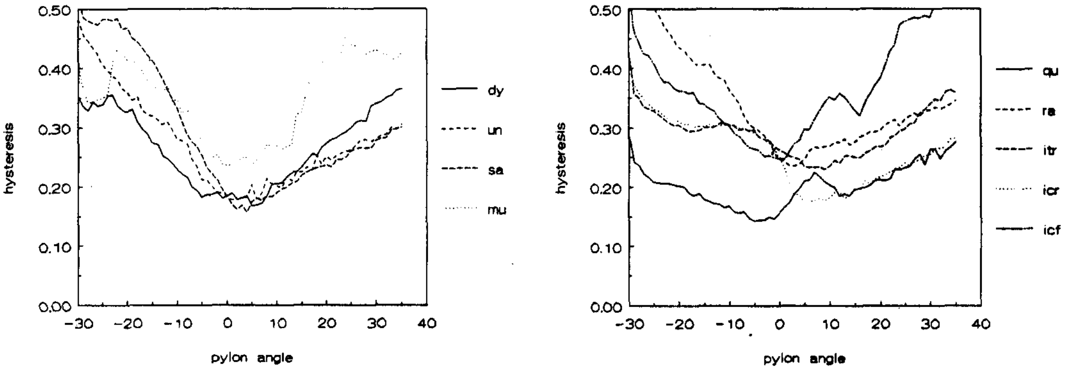

Fig. 7. Hysteresis data of all feet with leather shoe (abbreviations see Table 1).

capacity. Footwear has for the most part an increasing effect on hysteresis, particularly the leather shoe. An extreme increase is found in the case of the SACH foot in heel contact.

The reliability of the measurements is shown in Figure 8 in terms of the standard deviation, calculated using five measurements on the icf-foot. Also one measurement was carried out at half the speed of the usual protocol and the results show only a slight increase of hysteresis due to the lower testing speed.

During gait the forces on the foot will change much faster than in this simulation with the measuring device where the total test cycle of one foot lasts about 1½ hour. In practice the feet will probably have a lower energy absorption rate and higher stiffness due to a higher deformation velocity.

Conclusions

The foot shapes do not differ as much as would be needed to explain the differences in stiffness. The stiffness can only be explained by variations in material used in the manufacture. How the use of different materials can cause a particular stiffness-angle curve is not discussed here.

The influence of the shoes cannot be confined to compression of the sole alone, because this should decrease the stiffness as in a serial spring model. Increased stiffness can only be explained by the bending of the sole and the deformation of the shoe cover.

Although the roll-over characteristics cannot yet be judged, a low stiffness at toe-off position may be considered to obstruct a proper push off at the end of the stance phase; just as a low stiffness at heel strike may cause too great a vertical displacement of the overlying body segments.

To provide a better numerical survey of the characteristics during the stance phase values for the position −15° and 30° are printed in Table 2. These values are derived after smoothing the stiffness curves in Figures 3, 4 and 5 with a 4th grade polynomial for the part left of the maximum and a 5th grade polynomial for the right part.

The tendency for a "soft" foot to become stiffer

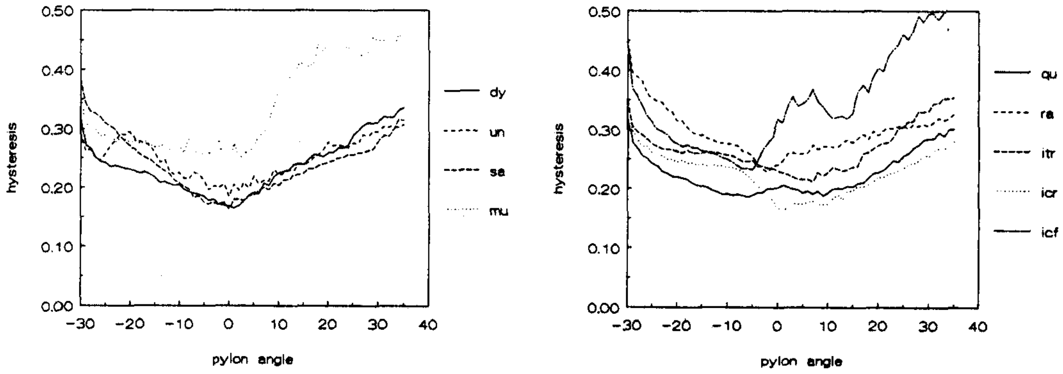

Fig. 8. Hysteresis data of all feet with sport shoe (abbreviations see Table 1).

Table 2. Stiffness values at two positions after smoothing with polynomials.
(wi = without shoe, le = with leather shoe, sp = with sport shoe.)

heel strike (−15°)	wi	le	sp	toe-off (30°)	wi	le	sp
mu :	89.9	76.6	66.0	ra :	40.2	35.5	34.7
icf :	85.6	73.9	72.7	icr :	36.7	32.9	33.8
icr :	77.2	69.8	57.0	un :	33.8	36.4	36.6
itr :	74.4	57.4	51.5	sa :	28.7	31.3	27.3
qu :	72.2	59.0	49.7	dy :	16.7	19.7	13.3
ra :	66.0	73.5	63.8	mu :	13.3	13.7	11.7
un :	54.6	63.4	66.1	itr :	12.4	15.2	12.5
dy :	53.8	61.3	57.0	icf :	11.6	8.6	11.7
qu :	44.0	49.2	49.5	qu :	5.2	7.1	6.0

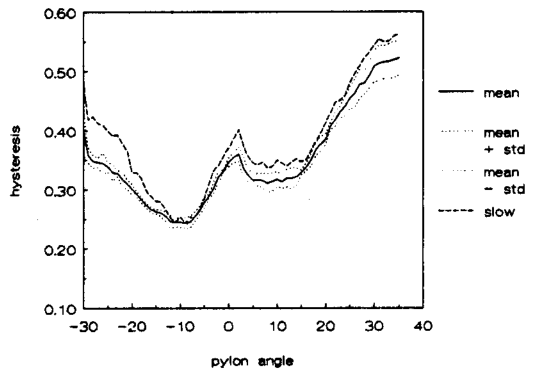

Fig. 9. Reliability indication of the hysteresis after five tests and one test at half the speed.

and a "stiff" foot softer by using a shoe in both positions is remarkable. This tendency can be explained by the deformation of the shoe cover and the bending of the shoe sole, the effects of which are large in comparison to a flabby foot and small in comparison to a stiff foot. Trying to represent this in a spring model would give the foot in parallel with the shoe-cover and both in series with the shoe-sole. In Table 2 the distinction between the "soft" and "stiff" feet is shown by the dotted line and the exceptions to the rule are printed in a bold type face. At 30° this change over for a sport shoe is much smaller than for a leather shoe, caused by the large difference in stiffness of the cover at the forefoot of both shoes.

An indication is given to explain the modifying influences of the footwear on several feet. To actually prove the spring model thesis, measurements with shoe parts would be required.

As far as the hysteresis is concerened, if as low a value as possible is to be recommended, the Hanger Quantum foot proves to be the best of all feet examined using this experimental setup.

In general we may conclude that uni- and multi-axial feet absorb more energy than feet with a rigid ankle device due to the hysteresis in the deformation of the rubber parts and friction in the axis. When adding a leather shoe the hysteresis increases for all feet except for the Multiflex foot, and for the Ipos foot with the flexible ankle which shows only a little increase.

To investigate the behaviour of the footwear at −15° and 30°, an 8[th] grade polynomial is used to smooth the data before presentation in Table 3. At 30° almost all feet show an increase in hysteresis on adding a shoe and it appears that the smaller the hysteresis without a shoe, the larger the increase with one. At −15° where there is a high hysteresis without shoe a decrease occurs for the sport shoe which is accented by the dotted line in Table 3 but on the other hand an increase is

Table 3. Hysteresis values at two positions after smoothing with polynomials.
(wi = without shoe, le = with leather shoe, sp = with sport shoe.)

heel strike (−15°)	wi	le	sp	toe-off (30°)	wi	le	sp
mu :	0.33	0.40	0.28	icf :	0.50	0.50	0.49
ra :	0.31	0.42	0.30	mu :	0.43	0.42	0.44
icr :	0.27	0.31	0.25	itr :	0.30	0.34	0.34
itr :	0.27	0.30	0.26	ra :	0.30	0.33	0.31
icf :	0.26	0.35	0.26	dy :	0.24	0.33	0.31
un :	0.19	0.33	0.27	un :	0.22	0.28	0.30
sa :	0.15	0.43	0.25	icr :	0.22	0.25	0.26
dy :	0.14	0.29	0.22	sa :	0.21	0.28	0.28
qu :	0.10	0.19	0.21	qu :	0.20	0.26	0.28

observed for the leather shoe. Because the reducing effect is most active at −15° for the sport shoe it may be concluded that the heel deformation of the sport shoe causes less auxiliary hysteresis in comparison to the leather shoe.

The smaller hysteresis for the sport shoe when compared with the leather shoe is a striking result. Taking into consideration the increased shock absorption of prosthetic feet when combined with sport shoes, compared to leather shoes, amputees can for the improvement of this particular characteristic, be recommended to walk on sport shoes instead of stiff leather shoes.

The Ipos and Hanger feet are designed on the principle which separates two functions of the foot: the mechanical properties and the cosmetic appearance. A soft rubber coating is necessary for cosmetic reasons and scarcely contributes to the mechanical properties. Hanger manufacturers succeeded in avoiding dissipative rubber elements in their design, resulting in a low energy absorbing foot.

REFERENCES

Daher, R. L. (1975). Physical response of SACH feet under laboratory testing. *Bull. Prosthet. Res.* **10(23)**, 4–50.

Donn, J. M., Porter, D., Roberts, V. C. (1989). The effect of footwear mass on the gait patterns of unilateral below-knee amputees. *Prosthet. Orthot. Int.* **13**, 140–144.

Ehara, Y., Bepppu, M., Nomura, S., Kunimi, Y., Takakashi, S. (1990). Energy analysis of energy-storing prosthetic feet. In: *Proceedings International Symposium Gait Analysis,* 2–3 February 1990, Berlin.

Koopman, H. F. J. M. (1989). *The three-dimensional analysis and prediction of human walking.* Ph.D. thesis. Enschede, The Netherlands: University of Twente.

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

Powell, P. C. (1983). *Engineering with polymers.* London: Chapman and Hall.

Skinner, H. B., Abrahamson, M. A., Hung, R. K., Wilson, L. A., Effeney, D. J. (1985). Static load response of the heels of SACH feet. *Orthopedics* **8**, 225–228.

Veen, P. G. van de (1989). *An investigation of design criteria of modular endoskeletal lower limb prosthesis.* Ph.D. thesis, Enshede, The Netherlands: University of Twente.

Voisin, J. P. (1987). Dual-ankle springs (D.A.S.) foot-ankle system. *Orthot. Prosthet.* **40**, 27–36.

Winter, D. A., Sienko, S. E. (1988). Biomechanics of below-knee amputee gait. *J. Biomech.* **21**, 361–367.