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Energy storage and release of prosthetic feet
Part 1: biomechanical analysis related to user benefits

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Abstract

The energy storing and releasing behaviour of 2 energy storing feet (ESF) and 2 conventional prosthetic feet (CF) were compared (ESF: Otto Bock Dynamic Pro and Hanger Quantum; CF: Otto Bock Multi Axial and Otto Bock Lager). Ten trans-tibial amputees were selected. The study was designed as a double-blind, randomised trial. For gait analysis a V1CON motion analysis system was used with 2 AMTI force platforms. A special measuring device was used for measuring energy storage and release of the foot during a simulated step.

The impulses of the anteroposterior component of the ground force showed small, statistically non-significant differences (deceleration phase: 22.7-23.4 Ns; acceleration phase: 17.0-18.4 Ns). The power storage and release phases as well as the net results also showed small differences (maximum difference in net result is 0.03 J kg). It was estimated that these differences lead to a maximum saving of 3% of metabolic energy during walking. It was considered unlikely that the subjects would notice this difference. It was concluded that during walking differences in mechanical energy expenditure of this magnitude are probably not of clinical relevance.

Ankle power, as an indicator for energy storage and release gave different results to the energy storage and release as measured with the special test device, especially during landing response. In the biomechanical model (based on inverse dynamics) used in the gait analysis the deformation of the material is not taken into consideration and hence this method of gait analysis is probably not suitable for calculation of shock absorption.

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Introduction

The general concept of energy storage and release of prosthetic feet is that they store energy during mid-stance and release the energy when it is desired, i.e. during push-off. These events are based on two major phases (Winter and Sienko, 1988) consistently seen in ankle power graphs in normal subjects. A long energy dissipation phase, A1, is thought to be a result of eccentric contraction of plantar flexors as the leg rotates forward over the flat foot, which is followed by a large energy generation phase, A2 (see Figure 4, sound side). This phase is due to concentric activity of the plantar flexors before toe-off. These parameters are calculated with the use of a biomechanical model, based on inverse dynamics.

In conventional prosthetic feet most of the stored energy is dissipated in the material. In so called energy storing feet most of the energy is said not to be dissipated in the material, but stored in the spring mechanism that should release it during push-off. Quantities of energy storage and release, as calculated from gait analysis, are not only dependent on the material and construction of the prosthetic foot, but also on many variables concerning the user, such as walking speed and body weight. Besides, footwear has a major influence on the properties of prosthetic feet (Jaarsveld *et al,* 1990).

One of the main contributors in the calculations of energy storage and release is the ground reaction force. The pattern of the ground reaction force may be a valuable indicator, because different authors suggest that a larger energy release of the prosthetic foot, during push-off, results in an increase of the second maximum of the anteroposterior ground reaction force (Blumentritt *et al,* 1994; Wagner *et al,* 1987). This second maximum of the ground reaction force is a 'direct' measured parameter without being influenced by assumptions of a biomechanical model. However the literature is not unanimous and different authors were not able to confirm an increase in this parameter for energy storing feet (Arya *et al,* 1995; Barr and Siegel, 1992; Menard *et al,* 1992; Torburn *et al,* 1990 and 1995).

Kinematics and stride characteristics are often used to prove differences between conventional and energy storing prosthetic feet. Barr and Siegel (1992), Lehmann *et al.* (1993), Perry and Shanfield (1993) and Wagner *et al.* (1987) reported for the energy storing feet a statistically significant increase in the ankle range of motion, especially in late stance dorsiflexion. The increase in dorsiflexion range is probably dependent on the construction of the prosthetic foot. It is not clear however whether there is a relation between properties of energy storage and release and late stance dorsiflexion.

Aim of this part of the study

The aim of the study was to obtain a better understanding related to user benefits of energy storing and release behaviour of some prosthetic feet that are used regularly in patient care. It was clear that there were some subsidiary

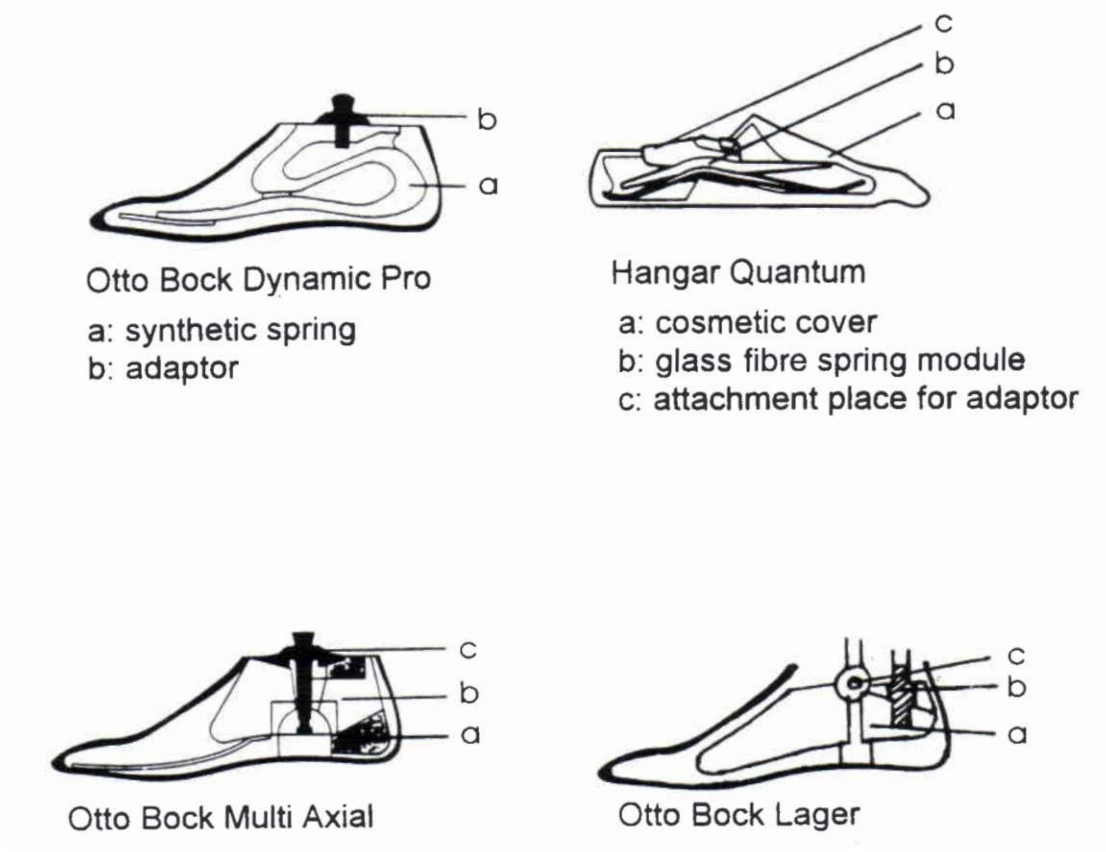


Fig 1. Basic constituents of 4 prosthetic feet.

questions which needed to be answered, such as:

- does gait analysis show differences in kinematic data and mechanical energy storage and release between so called energy storing feet and conventional feet when the feet are selected from the same price range? If differences do exist, are they clinically relevant?
- does energy storage and release as measured in a special test device (Biomedical Engineering Group of the University of Twente) with a standardized stance phase differ from the energy storage and release as calculated from the gait analysis? In other words what is the influence of the subject, in the laboratory situation, on the storage and release of energy of the prosthetic feet?

Materials and methods

Prosthetic feet

The following 4 designs of prosthetic feet were chosen, Otto Bock Multi Axial (CF1), Otto Bock Lager (CF2), Otto Bock Dynamic Pro (ESF1) and Hanger Quantum (ESF2). The first 2 are of a conventional variety and the other 2 are known to have energy storing properties. Some experienced orthopaedic technicians described the subjective characteristics of these 4 as follows:

- Otto Bock Multi Axial: stiff, mobile in sagittal and frontal plane.
- Otto Bock Lager: supple, mobile in plantar flexion direction (with ankle axis), stiff in dorsiflexion direction.
- Otto Bock Dynamic Pro: stiff in all directions.
- Hanger Quantum: very supple in all directions.

Figure 1 shows the basic constituents of the feet. The energy storing feet both show a spring mechanism while the others do not.

Since it is generally known that the properties of the shoe (e.g. stiff or supple) influence the properties of the prosthetic foot during walking all subjects were provided with the same brand of supple shoes.

Subjects

Ten trans-tibial amputees were selected. They were all active walkers who were able to walk at least 1 kilometre without any problem. None of the subjects had any stump problem. All subjects were informed in detail about the study and signed an informed consent form. Table 1 summarises the descriptions of the subjects.

Study design and data analysis

The study was designed as a double blind, randomised trial. Neither the investigator nor the subjects know which variety of foot was mounted on the prosthesis. A co-worker carried out the randomisation with the aid of a dice. The code was broken after the entire trial was completed. Every time a foot was supplied there was a habituation period of 2 weeks. The measurements were then carried out. After the measurements the foot was replaced with a different kind. Correct alignment of the prosthesis is very important because it influences the 'walking properties' of the foot and consequently the energy absorption and release. The feet were therefore always fitted and an alignment carried out by the same orthopedic technician who was not otherwise

Table 1. Description of subjects

Subject	M/F	Age (years)	Body-weight (kg)	Years since amputation	Reason for amputation	Own prosthetic foot
01	m	34	87	16	traumatic	Quantum
02	m	34	85	34-24 (reamputation)	congenital	Seattle
03	m	63	95	28	traumatic	Quantum
04	f	52	64.5	26	traumatic	Seattle
05	m	58	82	23	traumatic	1H32
06	m	66	84	46	traumatic	Endolite
07	m	50	82	20	traumatic	Endolite
08	m	43	80	23	traumatic	1D10
09	m	50	76	2	vascular	1D10
10	m	42	85	31	traumatic	1H32

involved in the trial.

Gait analysis

All the measurements were carried out at the Roessingh Research and Development Laboratory based at Het Roessingh, Centre for Rehabilitation. The gait analysis was carried out using a VICON motion analysis system (Oxford Metrics Ltd, Botley, Oxford, UK). The system consisted of 5 standard ccd cameras fitted with infrared filters linked to an Etherbox data acquisition system and a host computer (Micro VAX 3100). AMASS software for three-dimensional data collection was used for capturing kinematic data. The marker detection rate was 50 Hz. Two AMTI force platforms, operating at a sampling rate of 200 Hz, were used in conjunction to determine the ground reaction forces. VICON Clinical Manager software as used to compute walking speed, kinematic and kinetic data.

During the gait analysis 13 reflective markers were taped to both sides of the body at designated anatomical landmarks such as the sacrum, hips, upper legs, knees, lower legs, ankles and at the dorsal aspect of the metatarsal phalangeal joint II.

The subject was positioned at the end of the walkway and was asked to walk at a comfortable speed (free velocity). The distance between the force platforms was adapted, based on estimated step length of the subject, to accomplish a clean foot strike on each force platform. The walkway was 10 metres long but only a length of 4 metres in the middle of it was designated for data collection. The subjects were not informed about the use of the force platforms and they were not asked to hit them. At each session 10 trials were selected in which both feet hit the force platforms cleanly. Using the VICON Clinical Manager an average of the 10 trials was calculated. Then the kinematics and kinetic parameters were ascertained. Walking speed and cadence were measured in order to determine if differences exist which are likely to influence the ground reaction force.

The following parameters were calculated from the gait analysis:

- walking speed.
- cadence ($step.min^{-1}$).
- range of movement at hips, knees and ankles, with early stance plantar flexion and late stance dorsiflexion.
- impulse of deceleration and acceleration phase of the anteroposterior component of the ground reaction force. The impulse is the time-integral between the zero crossings.
- energy storage (Al phase), release (A2 phase) and final net values are calculated from the total ankle power.

Hysteresis

Hysteresis (internal friction) of the material of a prosthetic foot results in loss of energy when variable loading on the foot is applied. This loss of energy for the 4 test feet was measured using a special test device of the biomedical Engineering Group of the University of Twente, as described by Van Jaarsveld *et al.* (1990), but with the difference that the foot is loaded continuously while rotating from heel to forefoot (artificial roll-off movement). All of the feet were of the same size and were mounted on the test device with the same shoe.

Energy is calculated as the integral of force with respect to displacement. The force generated by the test device was equal to the measured vertical ground reaction force as a function of the shank floor angle of a subject (amputee, good walker) of mass 80 kg. The displacement was taken as the deformation of the foot as a result of this applied force. The horizontal ground reaction force was not applied. In this study the angle between the shank and the floor was 32° at initial floor contact and 40° at toe-off.

Figure 2 shows the graph of energy storage and release as measured with the test device. It shows the amount of energy (J, Y-axis) against shank-floor angle (X-axis). Initially there is storage of energy by the hindfoot during landing (shock absorption) as indicated by the downward direction of the graph during early stance (A). The amount of energy stored (A') is a measure of the stiffness of the hindfoot. The less the energy stored the stiffer is the hindfoot. From foot-flat to mid-stance (B) some energy is returned (B'). The amount of energy which is not returned (B") is mainly dissipated although part might be transferred to the forefoot and returned during push-off. However, it is assumed that the amount of energy transferred to the forefoot in this way is negligible. After mid-stance until push-off (C) the origin of the ground reaction force shifts to the forefoot. The

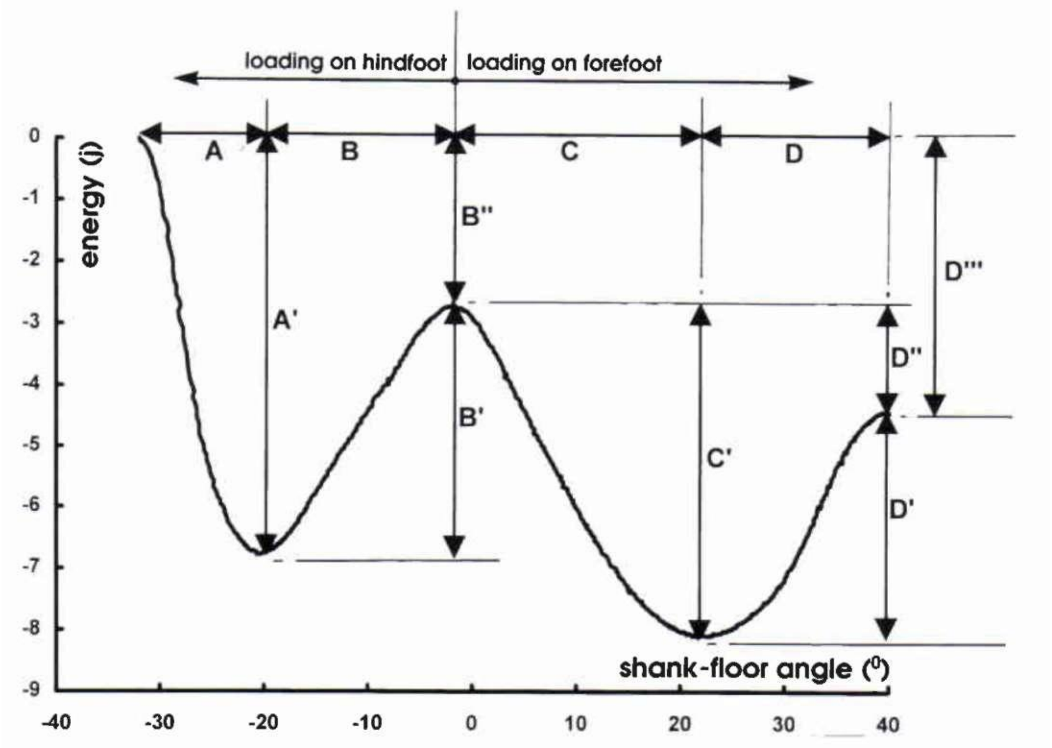


Fig. 2. Energy storage and release of a prosthetic foot, as measured with the test device (for explanation see text).

forefoot is deformed and therefore stores energy. The amount of energy stored (C) during the latter half of the stance phase (from mid-stance until push-off) is a measure of the stiffness of the forefoot. During push-off (D) part of that energy is returned (D'). The difference between the value at mid-stance and the value at toe-off (D") is the energy loss as a result of absorption of energy at the forefoot. B"+D" gives the amount of the total loss of energy, due to absorption.

Statistics

For statistical calculations multivariate analysis of variance was used with repeated measures' design with difference contrast. Thus the results for the different feet of every subject are compared within this subject, the only within subject factor was the type of prosthetic foot.

Results

One subject did not show up for the measurements with one of the feet. This resulted in a missing value. All statistical procedures, concerning the gait analysis, were performed on the results of 9 subjects.

Walking speed and kinematic data

Walking speed and kinematic data are shown in Table 2. During every session the subjects were asked to walk at a comfortable walking speed.

The mean walking speed was almost the same for the 4 feet (1.3-1.36 ms^{-1}). The mean difference between the 4 feet was $0.02 m s^{-1}$.

The mean cadence was almost the same for the 4 feet (0.87-0.88 steps min^{-1}). The mean difference between the 4 feet was 0.01 steps min^{-1}.

The range of motion of the hip with the CF2 was statistically significantly (p=0.04) larger than with the ESF1 and the CF1. The difference was less than 2°.

The range of motion of the knee showed no statistically significant differences between the 4 feet (p=0.117). The range of motion at the ankle with the CF2 was greater than that of the other 3 feet. The difference was statistically significant (p=0.003). The larger range of the

Table 2. Walking speed and kinematic data of amputated side; mean values and standard deviations.

Kinematic data	ESF1 (s.d.)	ESF2 (s.d.)	CF1 (s.d.)	CF2 (s.d.)
walking speed (m s^{-1})	1.35 (0.20)	1.34 (0.21)	1.36 (0.19)	1.34 (0.20)
cadence (steps min^{-1})	0.88 (0.07)	0.88 (0.06)	0.88 (0.05)	0.87 (0.07)
hip ROM (^{0})	46.6 (5.0)	46.7 (4.8)	45.6 (4.4)	47.3 (4.7)*
knee ROM (^{0})	65.2 (7.7)	62.4 (13.5)	58.1 (12.6)	60.0 (11.9)
ankle ROM (^{0})	16.8 (3.6)	17.2 (2.8)	17.2 (3.8)	21.2 (2.6)*
late stance dorsiflexion (^{0})	13.4 (2.9)	13.7 (2.5)	12.4 (3.0)	11.9 (2.2)
early stance plantar flexion (^{0})	3.4 (1.6)	3.2 (1.5)	5.3 (2.1)	9.4 (3.6)*

*Statistically significant.

motion with the CF2 was probably due to the ankle mechanism, which allows the foot to make a plantar flexion movement directly after heel contact. The early stance plantar flexion with the CF2 was 9.4°, while for the other feet this motion was 4 to 6° smaller (p=0.001). There was no statistically significant difference in late stance dorsiflexion for the 4 prosthetic feet (p=0.145).

Anteroposterior component of the ground reaction force

Table 3 gives the averages of the impulses of the anteroposterior component of the ground reaction force in the deceleration phase and in the acceleration phase. The measurements with the 4 prosthetic feet did not show any statistically significant difference in the impulses of the deceleration (p=0.307) or acceleration (p=0.179) phase.

Energy storage and release as calculated from the total ankle power

Table 4 gives the mean values of energy storage during phase Al and energy release during phase A2 with all prosthetic feet, calculated from the total ankle power. The mean storage of the ESF2 (0.17) was more than that of the other feet (which varied from 0.13 to 0.15 J kg^{-1}). The differences however were not statistically significant (p=0.675). The differences in the release of energy, between the 4 feet, were smaller (0.03-0.05 J kg^{-1}).

However these differences were statistically significant. The release of energy of the ESF2 was greater than that of the CF2 (p=0.026) and the release of energy of the ESF1 was greater than that of the other feet (p=0.025). The net result of the energy storage and release gives the absorption during the Al and A2 phases. There were only small differences in the net results of the energy storing and releasing phases and these differences were not statistically significant (p=0.549).

Energy storage and release as calculated with the special test device

Figure 3 gives the curves of the energy storage and release of 4 prosthetic feet (same size, same shoe). It shows the amount of energy (J, Y-axis) needed to reach a specified shank angle (X-axis). The values are given in Table 5.

Table 3. Impulses (N s) of the deceleration and acceleration phases of the anteroposterior ground reaction forces with standard deviation shown in brackets.

Phase	ESF1 (s.d.)	ESF2 (s.d.)	CF1 (s.d.)	CF2 (s.d.)
Deceleration	22.7 (4.5)	23.1 (4.2)	23.4 (4.2)	23.1 (3.9)
Acceleration	18.4 (3.9)	17.0 (3.5)	18.3 (4.1)	17.6 (4.1)

Table 4. Energy storage (Al phase), energy release (A2 phase) and net results (Al and A2) of the ankle in J kg^{-1} with standard deviation between brackets

Energy	ESF1 (s.d.)	ESF2 (s.d.)	CF1 (s.d.)	CF2 (s.d.)
storage (A1)	0.125 (0.03)	0.17 (0.02)	0.13 (0.06)	0.15 (0.10)
release (A2)	0.05 (0.02)	0.04 (0.02)	0.04 (0.01)	0.03 (0.02)
absorption (net result A1 and A2)	0.10 (0.02)	0.12 (0.02)	0.09 (0.05)	0.12 (0.09)

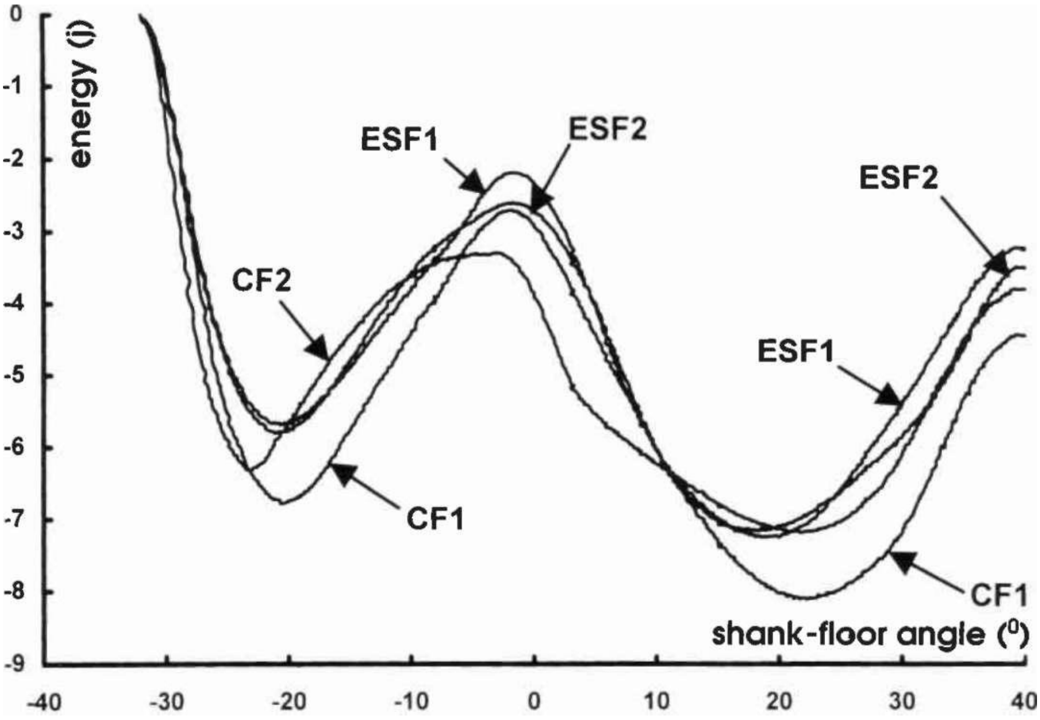


Fig. 3. Energy storage and release of 4 prosthetic feet: ESF1 (Otto Bock Dynamic Pro), ESF2 (Hanger Quantum), CF1 (Otto Bock Multi Axial), CF2 (Otto Bock Lager) as measured with the test device (J).

The results can be summarised as:

- hindfoot, energy storage (A'): the hindfoot of both non-energy storing feet store more energy.
- hindfoot, energy release (B'): most energy was released by the CF1.
- hindfoot, energy dissipation: the largest dissipation of the hindfoot, was found in the CF2. Both energy storing feet showed the lowest dissipation.
- forefoot, energy storage (C): the forefoot of the CF1 stored the most energy, while the CF2 stored the least energy.
- forefoot, energy release (D'): most energy was returned by the ESF1.
- forefoot, energy dissipation (D"): lowest dissipation was in the CF2 and highest in the CFl.
- total dissipation (B"+D"): this was lowest for the ESF1 and highest for the CF1.

Table 5. Energy storage and release as measured with the test device. The values are in J for a subject of 80 kg mass. In brackets are the values per kg body mass.

Foot	A'	B'	C'	D'	E'	F'	G"
ESF1	5.7 (0.071)	3.5 (0.044)	2.2 (0.028)	5.0 (0.063)	4.0 (0.050)	1.0 (0.013)	3.2 (0.040)
ESF2	5.8 (0.073)	3.1 (0.039)	2.7 (0.034)	4.5 (0.056)	3.7 (0.046)	0.8 (0.010)	3.5 (0.044)
CF1	6.8 (0.085)	4.0 (0.050)	2.8 (0.035)	5.3 (0.066)	3.7 (0.046)	1.6 (0.020)	4.4 (0.055)
CF2	6.3 (0.079)	3.0 (0.038)	3.3 (0.041)	3.9 (0.049)	3.4 (0.043)	0.5 (0.006)	3.8 (0.048)

A'=energy stored hindfoot
B'=energy release hindfoot
B"=dissipation hindfoot (A'-B')
C'=energy storage forefoot
D'=energy release forefoot
D"=dissipation forefoot (C'-D')
B"+D"=total dissipation

Discussion

Walking speed and kinematic data

Walking speed and cadence: in the laboratory situation the mean of the differences of walking speed and cadence were so small that it is assumed that it is unlikely these differences influenced the parameters which are based upon the ground reaction force. This is in agreement with most reports in literature.

Hip and knee range of motion: in the results reported here the CF2 showed a very small, but statistically significant (p=0.040) larger hip range of motion than the ESF1 and the CF1. The difference however was so small that it hardly could be of clinical importance. The differences of the knee range of motion were much bigger, but these differences were statistically not significant (p=0.117).

Ankle range of motion: most studies demonstrate an increase in the range of motion of the ankle of energy storing feet (Barr and Siegel, 1992; Lehmann *et al.,* 1993; Perry and Shanfield 1993; Wagner *et al.,* 1987).

The results reported here were different from these findings. It was found that the range of motion at the ankle of the CF2 was clearly and statistically significantly (p=0.003) greater than the range of motion at the ankle of the other feet. This was probably due to the ankle mechanism in this foot.

Looking at the ankle motion, the total range is the sum of motions during early stance in the plantar flexion direction, and then in late stance in dorsiflexion. The early stance plantar flexion of the CF2 was clearly and statistically significantly greater than that of the other feet (p=0.001).The early stance plantar flexion of the CF1 was greater than that of the ESF1 and the ESF2 (p=0.050). Both conventional feet (with ankle axis) showed greater early stance plantar flexion motion. The late stance dorsiflexion motion in the ankle showed no statistically significant difference (p=0.145). Various authors have reported a greater late stance dorsiflexion for energy storing feet. However, mostly the SACH foot was used as the conventional foot, while probably other conventional feet, especially those with an ankle joint, would have shown a greater range of motion at the ankle. Besides an ankle axis, the stiffness of the foot also will have an influence on the range of motion at the ankle.

Not only foot-related factors play a role in the range of motion at the ankle. Different authors hag reported that the physical condition of the amputee, traumatic amputees versus vascular amputees, is an important influence on gait parameters such as the late stance dorsiflexion and walking speed (Barth *et al.,* 1992; Casillas *et al.,* 1995; Hermodsson *et al.,* 1994; Torburn *et al.,* 1990). This means that the more active the subject, the more late dorsiflexion he needs/makes. In the authors' study the group of subjects was too small and too homogeneous (all active walkers) to enable this to be confirmed.

Importance of late stance dorsiflexion

It can be argued that the late stance dorsiflexion of the prosthetic foot is related to balance control. Balance control is, among other things, dependent on proprioceptive feedback. Proprioceptive feedback is impaired in patients suffering from polyneuropathy (i.e. due to diabetes).

An increase in late stance dorsiflexion results in an increase of knee flexion moment and thereby decreases knee stability. With poor balance control this leads to unsafe situations and therefore these amputees prefer prosthetic feet which allow only limited late stance dorsiflexion. This is in agreement with the conclusion of Casillas *et al.* (1995) that 'the vascular patient seeks easy proprioceptive support control on the amputated side - in other words, maximum safety'. Late stance dorsiflexion permits a supple roll-off movement. With limited dorsiflexion the roll-off movement will be impaired. Most active walkers do not prefer this, they prefer a supple roll-off. High level active subjects, without impairment of balance control, therefore should be provided with a prosthetic foot which allows a wide extent of dorsiflexion (Barth *et al.,* 1992).

Energy storage and release

In the literature different methods are described to assess energy storage and release of prosthetic feet. Some authors calculated an efficiency parameter from energy storage and release (Barr and Siegel, 1992; Schneider *et al.,* 1992). The energy release is expressed as a percentage of that stored. This is a relative parameter which gives information about the properties of the foot materials. However it gives no information about the absolute quantities of energy storage and release and thus it gives no information about the amount of energy dissipation which in the authors' opinion is essential with respect to energy expenditure.

Others consider the sum of the absolute values of the Al and A2-power bursts as a measure of efficiency (Ehara *et al,* 1983; Goh *et al,* 1994). With respect to energy expenditure however this gives no clear information, because the 'same energy' is appraised twice (first during the storing phase and secondly during the releasing phase).

Sacchetti *et al* (1994) looked at both bursts (Al and A2) separately and also at the net value of both bursts as a measure of energy storing and releasing capacities of the prosthetic foot. In this way information about energy storage and release, as well as total amount of dissipation is obtained. The authors consider this is preferable and therefore used this method of describing and discussing energy storage, release and dissipation.

This study showed only little differences in energy storage, release and dissipation, despite the differences in construction of the conventional and energy storing feet. The storage during the Al phase showed no statistically significant differences between the 4 feet. The differences in energy release during the A2 phase were even smaller, however, statistically significant. The net results also did not show a statistically significant difference. It is possible that with a greater number of subjects the differences in energy storage and in the final net results also become statistically significant. Even if this were true, are differences of this size of clinical relevance? A lower leg amputee needs ± 10% more energy than a normal subject. A normal subject needs about 3.1 to 3.3 J $kg^{-1}m^{-1}$ and a lower leg amputee needs about 3.4 till 3.6 J kg m of metabolic energy to walk at a speed of 5 km h^{-1} (Corcoran 1971; Donn and Roberts, 1992). A stride, about 1.5 m long, needs 5.1 till 5.4 J kg^{-1}. The biggest difference in the net absorption between the 4 feet was 0.03 J kg per stride (mechanical energy). This means that the subject needs 0.03 J kg less for walking with the foot with the least net absorption. The power, necessary for walking, is the result of muscle activity. By a rough estimation, the efficiency of muscles is 20 to 25%. Therefore to supply 0.03 J kg^{-1} effectively for walking, the muscles have to raise 0.12 to 0.15 J kg^{-1}, which is less than 2.5 to 3% of the energy needed for 1 stride. There does not appear to be data in the literature identifying the difference in the expenditure of metabolic energy which can be noticed by subjects, while walking at comfortable speed. It is however unlikely that subjects notice such small differences during normal walking, or that a gain of this size is of clinical importance. Only at the top level of sport could differences of this size be of importance.

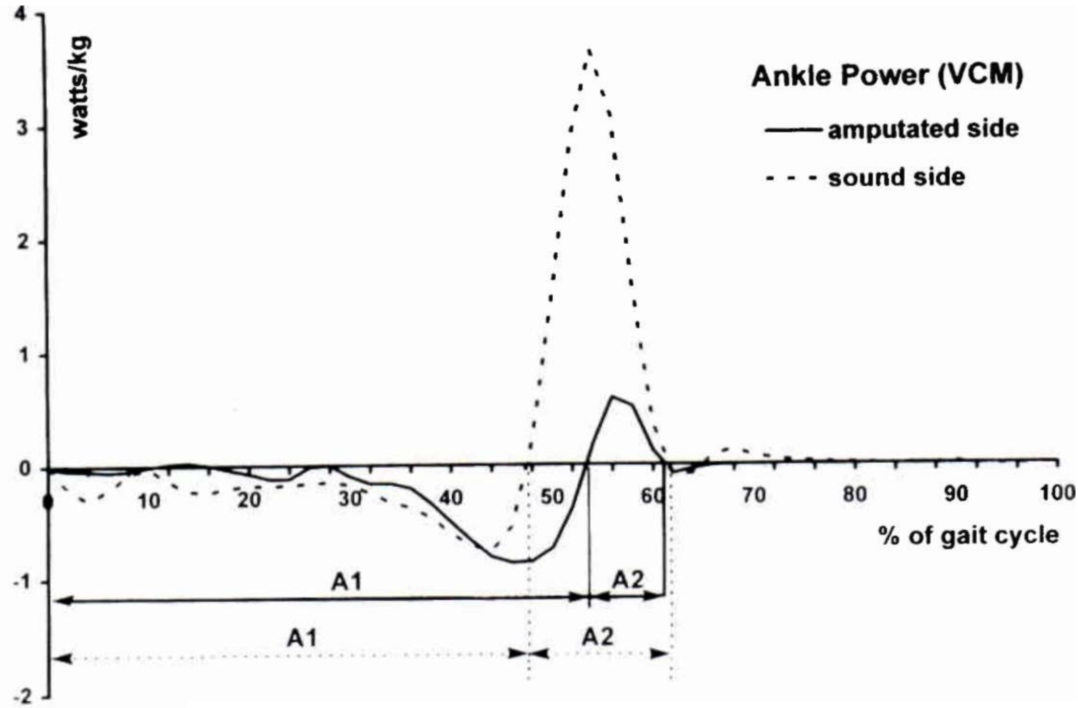
Gait analysis versus test device

A comparison of the energy storage and release of the hindfoot, as shown in the graph of the total ankle power and as shown in the graph of the test device showed some remarkable differences. Figure 3 shows the pattern of mechanical energy storage of the hindfoot, as measured with the test device. The storage takes place at heel loading. It is directly followed by a release of a part of the stored energy, until mid-stance. Although the storage of energy in the hindfoot during loading was a little larger for both conventional feet, the subjects did not experience a clear difference during heel loading. The storage and release of energy during the first part of the stance phase, as measured with the test device, was not seen in the graph of the ankle power, as is shown in Figure 4.

The reason for the observed difference of energy absorption during loading seems to be twofold. Firstly: the special test device uses for calculation forces and displacement, caused by deformation of the heel. The gait analysis uses for calculation an inverse dynamic model, based on rigid bodies approximation. This model does not take into consideration the deformation, and therefore it maybe not accurate enough to measure absorption during loading. Secondly: the accuracy of the gait analysis measuring system may be insufficient to cope with this problem.

It is concluded that the A1-power burst, as described by Winter and Sienko (1988), seems to contain only limited information about the amount of stored energy during shock absorption. Therefore it is not appropriate to consider the A1-power burst as an adequate indicator of energy storage caused by shock absorption.

The shock absorption however seems to be important in relation to comfortable walking since it has been proven that trans-tibial amputees prefer prosthetic feet which develop greater



A1=energy storage
A2=energy release

Fig.4. Total ankle power
X-axis: % gait cycle
Y-axis: Watts kg^{-1}, (pos=release; neg=storage)

damping at loading, in other words more energy dissipation (Wirta *et al,* 1991). This means that a lower net energy storage and release is not automatically related to a 'better' foot.

Both measurement systems were also compared in relation to the data of the energy storage and release of the forefoot.

The absolute data calculated from the Al and A2 phases of the ankle power and the test device showed some striking differences. The amount of storage as measured with the test device is 2 to 3 times smaller than is calculated from the Al phase, while the amount of energy release is about equal for both measurement systems. The net results, storage minus release, of the test device were about 10 times smaller than those of the Al and A2 phases.

These differences cannot be explained only by influences on the subjects such as different walking pattern, speed, non-linear weight influences etc., which are measured automatically in the gait analysis, but not with the test device. It is likely that differences in method of calculation of the 2 measurement systems and possibly differences in accuracy, are mainly responsible.

Conclusion

When comparing the 2 energy storing and the 2 conventional feet, there are no clear differences in kinematic data or in kinetic data. The range of motion at the ankle of the CF2 is bigger than that of the other feet, but this is due to the ankle mechanism and not to energy storing features.

The differences in mechanical energy storage and release (net results), as calculated from the ankle power, are small and not significant; the mean net absorption of the ESF1 is smallest. It is unlikely that differences in net dissipation of energy of the magnitude found in this study can be noticed during normal walking and therefore these differences are probably not clinically relevant.

The data of the test device (simulated stance phase) are in favour of the 2 energy storing feet. The total dissipation of energy for both ESF is less than for the CF. The size of the differences however, are again small and are unlikely to be clinically relevant.

During loading, energy is absorbed by the deformation of the foot material. This is measured with the test device (integral of force with respect to displacement) but not with the gait analysis system which uses for calculations an inverse dynamic model, based on rigid bodies approximation. Therefore this kind of gait analysis may not be suitable for calculation of energy storage due to shock absorption.

With respect to energy expenditure, in normal walking, energy storage and release of the prosthetic foot, seem only to be important when the gain in net absorption is much larger than for the energy storing feet in this study. A wooden foot would give the 'best' results (almost no energy storage, nor energy release) but would be also very uncomfortable, among other things, due to lack of energy absorption during loading and the fixed roll-off.

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REFERENCES

ARYA AP, LEES A, NIRULA HC, KLENERMAN L (1995). A biomedical comparison of the SACH, Seattle and Jaipur feet using ground reaction forces. *Prosthet Orthot Int* 19, 37-45.

BARTH DG, SCHUMACHER L, SIENKO THOMAS S (1992). Gait analysis and energy cost of below-knee amputees wearing six different prosthetic feet. *J Prosthet Orthot* 4, 63-75.

BARR AE, SIEGEL KL (1992). Biomechanical comparison of the energy-storing capabilities of SACH and Carbon Copy II prosthetic feet during the stance phase of gait in a person with below-knee amputation. *Phys Ther* 72(5), 344-354.

BLUMENTRITT S, SCHERER HW, WELLERSHAUS U (1994). Biomechanisch-ganganalystische bewertung von prothesanfiiBen. *Med Orth Tech* 114, 287-292.

CASILLAS J-M, DULIEU V, COHEN M, MARCER I DIDIER JP (1995). Bioenergetic comparison of a new energy-storing foot and SACH foot in traumatic below-knee vascular amputations. *Arch Phys Med Rehabil* 76, 39-44.

CORCORAN PJ (1971). Energy expenditure during ambulation. In: Physiological basis of rehabilitation medicine. Downey JAR, Darling RC.-Philadelphia, W.B. Saunders.

DONN JM, ROBERTS C (1992). A review of the energy expenditure of disabled locomotion with special reference to lower limb amputees. *Physiotherapy Theory Practice* 8, 97-108.

EHARA Y, BEPPU M, NOMURA S, KUNIMI Y, TAKAHASHI S (1983). Energy storing property of so-called energy-storing prosthetic feet. *Arch Phys Med Rehabil* 74, 68-72.

GOH JCH, TAN PH, TOH SL, TAY TE (1994). Gait analysis study of an energy-storing prosthetic foot - a preliminary report. *Gait Posture* 2, 95-101.

HERMODSSON Y, EKDAHL C, PERSSON BM, ROZENDAL G (1994). Gait in male trans-tibial amputees: a comparative study with healthy subjects in relation to walking speed. *Prosthet Orthot Int* 18, 68-77.

VAN JAARSVELD[1] HWL, GROOTENBOER HJ, DE VRIES J, KOOPMAN HJFM (1990). Stiffness and hysteresis properties of some prosthetic feet. *Prosthet Orthot Int* 14, 117-125.

VAN JAARSVELD[2], HWL, GROOTENBOER HJ, DE VRIES J (1990). Accelerations due to impact at heel strike using below-knee prosthesis. *Prosthet Orthot Int* 14, 63-66.

LEHMANN JF, PRICE R, BOSWELL-BESSETTE S, DARALLE A (1993) Comprehensive analysis of dynamic elastic response feet: Seattle Ankle/Lite foot Versus SACH foot. *Arch Phys Med Rehabil* 74, 853-861.

MENARD MR, MCBRIDE ME, SANDERSON DJ. MURRAY DD (1992). Comparative biomechanical analysis of energy-storing prosthetic feet. *Arch Phys Med Rehabil* 73, 451-458.

PERRY J, SHANEFIELD S (1993). Efficiency of dynamic elastic response prosthetic feet. *J Rehabil Res Dev* 30(1), 137-143

SACCHETTI R, SCHMIDL H, GRONINGEN MV (1994), Orienterende undersuchung uber energie-speidherung und energieruckgabe von prothesenfuBen. *Med Orth Tech* 114, 293-295.

SCHNEIDER K, HART T. ZERNICKE RF, SETOGUCHI Y, OPPENHEIM W (1992), Below-knee child-amputee gait: SACH foot verus FLEX foot. In: European Symposium on Clinical Gait Analysis, April 1-3, 1992.-Zurich:Laboratorium fur Biomechanik der ETH Zurich

TORBURN L, PERRY J, AYYAPPA E, SHANFIELD SL (1990). Below-knee amputee gait with dynamic elastic response prosthetic feet: a pilot study. *J Rehabil Res Dev* 27, 369-384.

TORBURN L, POWERS CM, GUITEREZ R, PERRY J (1995). Energy expenditure during ambulation in dysvascular and traumatic below-knee amputees: a comparison of five prosthetic feet. *J Rehabil Res Dev* 32, 111-119.

WAGNER J, SIENKO S, SUPAN T, BARTH D (1987). Motion analysis of SACH vs. Flex-Foot in moderately active below-knee amputees. *Clin Prosthet Orthot* 1, 55-62.

WINTER DA, SIENKO SE (1988) Biomechanics of below-knee amputee gait. *J Biomech* 21, 361-376.

WIRTA RW, MASON R, CALVO K, GOLBRANSON FL (1991). Effect on gait using various prosthetic ankle-foot devices. *J Rehabil Res Dev* 28(2), 13-24.