Prosthetics and Orthotics International, 1989, 13, 140–144

The effect of footwear mass on the gait patterns of unilateral below-knee amputees

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Abstract
This study reports an investigation into the effect of shoe mass on the gait patterns of below-knee (BK) amputees. Ten established unilateral BK, patellar-tendon-bearing prosthesis wearers were assessed using a VICON system of gait analysis. Incremental masses of 50g (up to 200g) were added to the subjects' shoes and data captured as they walked along a 15m measurement field. Coefficients of symmetry of various parameters of the swing phase (knee frequency symmetry, swing time symmetry, maximum flexion to heel strike time symmetry) were measured and their correlation was tested with the patient's preferrerd shoe mass and also their own shoe mass, all expressed as a proportion of body mass.

The subjects' 'preferred' shoe mass (139–318g) showed the greatest symmetry in all the parameters examined (correlations 0.78–0.81 $p<0.01$ and <0.005), whereas there was no correlation between the subjects' own shoe mass (121-325g) and the symmetry coefficients measured.

Introduction

Over five and a half thousand amputees are now being referred to Disablement Services Centres around the United Kingdom each year for prosthetic prescription, the majority of whom suffer from peripheral vascular disease, (Ham et al. 1989). Any advice given to them by members of the management team (therapists, prosthetists, bioengineers, doctors etc.) regarding their rehabilitation is very important, especially if this advice helps to improve their function during normal prosthetic use. The advice regarding footwear for the amputee is either non-existent, or that "the lighter the footwear the better". Lighter footwear is often thought to minimally affect the design performance of the prosthesis, despite the fact that there is little published evidence of consideration of footwear in prosthetic design.

In an early study of the effect of prosthetic foot mass, Godfrey et al. (1977) described the effect of changing the foot mass of a prosthesis on above-knee amputee gait. Stride length and heel rise velocity were considered to be of major importance because of the pendular quality of the above-knee prosthesis. However, no significant differences of performance were noted between the three different masses tested.

Several authors have looked at the different prosthetic feet available and their effects on gait. Doane and Holt (1983) compared the SACH foot and the single-axis foot in the gait of unilateral below-knee amputees. Temporal and kinematic data were obtained for a complete gait cycle on both sides. Of the data collected, only one parameter was thought to have any clinical significance and that was the difference of 6.5° in the range of ankle movement found between the single-axis foot and the SACH foot. In addition, the SACH foot also showed a limited range of plantarflexion.

Burgess et al. (1985) have evaluated the Veterans Administration (VA) Seattle foot. In their study the role of the energy storing/returning keel (which provides extra push at toe-off) was assessed, as were the natural feel and stability of the foot, the durability of the materials used and the mass of the foot. Currently prosthetic feet weigh between around 230g and 690g, with the VA Seattle foot

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weighing 460g. Consumer acceptance was the final measure of the Seattle foot and the amputees' comments were noted. The relative ability (compared with the use of other feet) to run at different speeds and on different surfaces was favourably commented upon, as was the extra push perceived at toe off. There was however no mention of the effect of the mass of the prosthetic foot.

A review of current prosthetic feet by Edelstein (1988) commented on many areas, such as the ability to accommodate different heel heights, the performance of the feet during stance phase, the energy storing materials used, and also their cost of production. Edelstein commented that to accommodate a change of more than 0.5cm in heel height either higher or lower, a change in the alignment of the prosthesis may be required, most designs of prosthetic feet were able to accommodate changes of up to 2-3cm, (this being achieved either by mechanically altering the alignment of the keel of the shoe to its heel, or by wedging material under the heel, as in the Seattle foot). The new energy storing feet, for which superior performance is claimed, use a variety of materials, from double carbon-fibre composite in the Carbon Copy II foot, to acetyl polymer and Kevlar fabric in the Seattle foot. The type of shoes used in their assessments may have influenced the performance of the various feet but they were not mentioned.

In the belief that many of these studies may have overlooked a factor which may significantly influence prosthetic use, this present pilot study looks at the subject of footwear, specifically the effect of shoe mass on the gait patterns of unilateral below-knee amputees.

Subjects

Ten unilateral BK amputees were assessed in this investigation. Nine were male, three subjects having undergone left-sided amputations and seven right-sided amputations. Body mass ranged from 57.0kg to 95.5kg with a mean of 79.9kg (±14.6). All subjects were patellar-tendon-bearing prostheses wearers and could walk at least 100m without the use of a walking aid. The subjects were all wearing single-axis feet except one, who was wearing a Quantum foot.

Method

The mass of the shoe with which each patient arrived at the laboratory was recorded, as was their body mass and the mass of the shoe the subject preferred during the tests.

The mass of shoes available commercially was tested and ranged from approx 250g (a light canvas shoe), to 460g (a heavy leather shoe). For the afternoon of the test the subjects wore a pair of Drushoes*. This is a make of orthopaedic footwear readily available in many hospitals and often issued to amputees with their first prosthesis. Each Drushoe weighed 260g. Investigations were carried out with the unweighted shoes and then following the sequential addition of plasticine (50g, 100g, 150g, 200g) to the front of each Drushoe. The plasticine was attached to the shoes as close to the ankle joint as possible to reduce any effect on ankle moments to a minimum. The range of shoe mass (Drushoe plus extra mass) was thus 260–460g, effectively covering the 'normal' range of shoe mass.

Kinematic information was obtained using the VICON 3-dimensional gait analysis system consisting of four infra red cameras and a PDP 11/23 computer. Foot switches were used in order to obtain accurate foot contact timing information and thus swing phase times.

Each subject was required to wear shorts and reflective markers were placed on the subject's skin at the following anatomical landmarks on the non-prosthetic side:

1 anterior superior iliac spine,
2 greater trochanter,
3 lateral line of the knee,
4 lateral malleolus,
5 head of the fifth metatarsal.

Markers were also placed at equivalent positions on the prosthetic side. Marker trajectories were recorded whilst the subjects walked down at 15m walkway at their own comfortable pace. From these marker trajectories information was obtained about hip, knee and ankle joint angle variations in the sagittal plane, throughout the swing phase of one gait cycle on each side. Joint angles were computed for the swing phase for both left and right sides. Each data set was normalised in time to 64 data samples using linear interpolation. In order to compare the change

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* John Drew (London) Ltd.

in joint angle values rather than the absolute magnitude, the mean angle value for each data set was subtracted from all points in that set. A Fast Fourier Transform was then performed in order to obtain the power spectrum for each angle waveform.

$$r=(\Sigma xy-(\Sigma xy)/N)/\sqrt{\{(\Sigma x^2-((\Sigma x)^2)/N)(\Sigma y^2-((\Sigma y)^2)/N)\}} \qquad \ldots\ldots\ldots\ldots 1$$

The Index of Symmetry was calculated as the correlation index using Equation 1 (above). In this equation x represents the amplitude of the angle at each harmonic for the left hand joint, and y the amplitude at the same frequencies for the right hand joint. The coefficient was calculated for the first ten frequency samples with the exception of the first since the joint angle offset had been previously removed. A symmetry coefficient of 1.0 would represent perfect symmetry, with the coefficients ranging from 0–1, this is the same procedure as followed by Hannah & Morrison (1984) and Pepino et al. (1986). Symmetry Indices were calculated for the hip, knee and ankle (Hip Frequency Symmetry, Knee Frequency Symmetry, Ankle Frequency Symmetry).

Swing time symmetry was calculated from the swing phase times obtained from heel and toe switch information. Left and right feet were compared and a symmetry coefficient (SWTS) was obtained.

The time taken from maximum knee flexion to heel strike was calculated from the knee joint angle vs swing time waveform. Again left and right were compared and a symmetry coefficient (MF-HS) obtained.

Results

From the temporal and kinematic data obtained, symmetry coefficients were calculated for three parameters at each mass tested. The three parameters were, Knee Frequency Symmetry in the Swing Phase (KFSS), Swing Time Symmetry (SWTS) and Maximum Flexion to Heel Strike Time Symmetry (MF-HS). Table 1 shows the information collected for one subject and is typical of the data collected on all ten. From this information the shoe mass which gave the most symmetrical gait, as indicated by the three parameters, could be identified and recorded.

Table 1. Symmetry coefficients for Subject 5.

Shoe mass tested (g)	KFSS	SWTS	MF-HS
260	0.921	0.833	0.893
310	0.932	0.709	0.786
360	0.987	0.935	0.935
410	0.933	0.833	0.921
460	0.942	0.719	0.719

The subject's own shoe and their preferred shoe mass was expressed as a proportion of their body mass, to allow comparison of results between subjects. Table 2 shows this and also whether the subjects preferred heavier or lighter shoes than their own.

Table 2. Body/shoe mass proportion related to preference.

Subject	Body mass (kg)	Own shoe mass proportion	Preferred mass proportion	Heavier/ Lighter
1	57.0	121	139	Lighter
2	60.0	300	193	Heavier
3	63.5	128	138	Lighter
4	76.0	325	245	Heavier
5	80.2	221	174	Heavier
6	84.0	258	183	Heavier
7	89.0	168	194	Lighter
8	95.5	281	265	Heavier
9	95.5	382	208	Heavier
10	98.6	253	318	Lighter

The shoe mass which indicated the most symmetrical gait in each of the parameters looked at was also expressed as a proportion of body mass (Table 3). The body to shoe proportion which gave the most symmetrical gait patterns for the subjects ranged from 146–318.

Table 3. Symmetry—body/shoe mass proportion.

Subject	KFSS	SWTS	MF-HS
1	158	184	158
2	146	146	231
3	155	176	176
4	245	245	245
5	223	223	223
6	233	271	183
7	247	217	217
8	233	233	208
9	208	208	208
10	318	274	274

Since all shoe masses were expressed as a proportion of body mass a correlation between indices is possible. In order to achieve this the shoe mass which provided the optimum results for each of the parameters was identified. A correlation was then tested between these shoe

Table 4. Correlation coefficients.

KFSS	
Own shoe correlation	0.44
Preferred mass correlation	0.78 ($p<0.01$)
MF–HS	
Own shoe correlation	0.57
Preferred sho correlation	0.80 ($p<0.05$)
SWING PERIOD PARAMETERS	
Own shoe correlation	0.49
Preferred mass correlation	0.81 ($p<0.005$)

masses and both the original shoe mass with which the subject arrived, and with the shoe mass which the subject preferred.

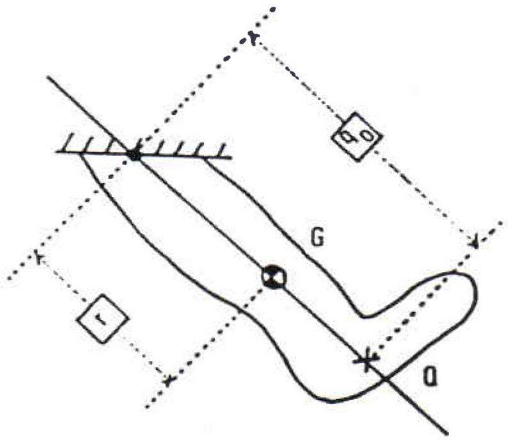
Table 4 shows the correlation coefficients calculated for KFSS, MF-HS and a combination of all three parameters looked at. The correlation calculated for the swing period parameters, used an average of the body to shoe mass proportion which produced the most symmetrical gait in each of the three parameters (KFSS, SWTS, MF-HS), for each subject.

Although the results showed that the optimum shoe mass exhibited little correlation with the subjects own shoe mass, it was however significantly correlated with their preferred mass in each parameter.

Discussion

Since changing footwear mass is most likely to change the gait pattern in the swing phase, this present study has been confined to only that part of the gait cycle. During the swing phase of the gait cycle the knee is flexing and then extending under pendular action. If the prosthetic shank and foot act as a simple pendulum the natural frequency and therefore the period of swing would be unaffected by the pendulum mass (i.e. prosthetic foot and shoe) but would be dependent only on the distance of that mass from the axis of rotation ie. the knee joint.

However, it can be argued that a prosthesis may more realistically resemble a compound pendulum due to the mass being distributed throughout the stump, prosthetic shank and foot, with the added restraints of muscular control of the knee. Figure 1 shows a prosthesis represented as a compound pendulum, (Steidal 1971). The position of the plasticine mass, close to both the normal and prosthetic ankle joints prevents a significant shift in the centre of mass of the shank and foot in an antero-posterior



$$f = \frac{1}{2\pi}\sqrt{\frac{g}{q_0}}$$

G – centre of mass

Q – centre of percussion

$$q_0 = \frac{K_0}{r}$$ (K_0 – radius of gyration)

= distance from the axis to Q

Fig. 1. Compound pendulum.

direction. By adding mass to the footwear, the position of the centre of mass will however be shifted distally therefore increasing r and consequently altering q°, (the distance from the knee joint to the centre of percussion). Note that the natural frequency of the pendulum is inversely proportional to q_o, which in turn will be affected as suggested, by the shoe mass.

In this present study, evidence was found that altering the shoe mass changed the swing time, as shown by the swing time symmetry (SWTS). Components of gait symmetry, ie. knee frequency symmetry, maximum flexion to heel-strike time symmetry and SWTS have been examined to see how much the prosthetic side altered and approximated to the natural side during the swing phase. The possible correlation between gait symmetry and the subjects' acceptance of the shoes has also been examined.

The lack of correlation between the subjects' own shoe and symmetry could have been for two reasons. Either the subjects had received the wrong advice about their footwear (six out of the ten subjects preferred walking with shoes that were heavier than their own) or the subjects were accustomed to walking with the same type of shoes and had not experimented or tried any other type of shoe.

Consumer acceptance and approval was the final measure of success when the VA Seattle foot was evaluated (Burgess et al. 1985), and so it should be for the amputee's total prosthetic prescription. Footwear should be included as an item on the final prescription if it is found to have an influence on their walking patterns. It would seem from this present study that footwear *does* influence the amputee's gait pattern, both from an objective and subjective point of view.

Conclusions

Although only a preliminary study, the data collected are sufficiently convincing to be able to draw certain conclusions. Firstly, changing footwear mass *does* alter the gait pattern, as shown here, during the swing phase. Secondly, lightweight footwear does not necessarily provide the most symmetrical gait as shown by the collected data, or the most acceptable gait as preferred by the amputee. Thirdly, amputees should be encouraged with their choice of footwear to find a pair which suits them individually. This may be their only way of finely tuning their artificial limb to their everyday needs. Their footwear would then become part of their total prosthetic prescription.

Acknowledgements

With grateful thanks to Mr. J. M. Regan, Senior Prosthetist, Chas Blatchfords & Sons Ltd. Roehampton, formerly of Vessa Ltd. Also to the registered charity Action Research for The Crippled Child who sponsored J. D. for a Research Training Fellowship which enabled her to finish this work.

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