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Below-knee amputation: a comparison of the effect of the SACH foot and single axis foot on electromyographic patterns during locomotion

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Abstract

The purpose of this investigation was to measure the effect of two terminal prosthetic components, the SACH foot and the single axis foot, on the locomotion patterns of unilateral below-knee amputees. The ten subjects who participated in the study were evaluated on two occasions, once following prescription of a PTB prosthesis and one terminal device and once following its replacement with the second device. The two devices were allocated alternately at the time of prosthetic prescription to ensure that five of the subjects used the SACH foot initially and five used the single axis foot. The electromyographic activity of the vastus lateralis and medial hamstrings was recorded bilaterally during gait using Beckman surface electrodes. The EMG signals were full wave rectified and low pass filtered to obtain the linear envelope. Pressure sensitive footswitches were used to correlate the EMG signals with components of the gait cycle. The pattern of quadriceps and hamstring muscle activity of the contralateral limb was similar to that reported for normal individuals and was unaffected by changing the terminal device on the prosthetic limb. The pattern of quadriceps and hamstring activity of the prosthetic limb differed from that of the contralateral limb and was influenced by the change in terminal device. With both devices the muscles were active for a greater percentage of the stance phase when compared to the contralateral limb. With the SACH foot attachment there appeared to be more co-contraction of the quadriceps and hamstrings during the mid stance phase of gait.

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Introduction

Surveys conducted in the United States have revealed that 60-90% of the lower extremity amputations performed are due to complications of peripheral vascular disease (Davies et al, 1970; Kay and Newman, 1974; Kerstein et al, 1974). Kay and Newman (1974) reported that 62% of all lower extremity amputations and 88% of amputations due to disease were performed on persons over 50 years of age. The increasing emphasis on saving the knee joint since the mid 1940's has resulted in a greater potential for this patient population to achieve independent ambulation. Successful rehabilitation depends in part upon the appropriate prescription of a prosthesis to replace "the missing limb segment. The prosthesis must be able to provide stability as well as to compensate in part for the lost joint movement and muscle action thereby minimizing the increased demands placed on more proximal joints and muscles.

The patellar-tendon-bearing (PTB) socket, designed in 1959 by the Biomechanics Laboratory, University of California, is most commonly prescribed for subjects with below-knee amputation (Fishman et al, 1975). The terminal devices most often used with the PTB socket are the SACH (solid ankle cushion heel) foot and the single axis ankle joint and foot (Davies et al, 1970; Fishman et al, 1975).

The SACH foot, as its name implies, has no ankle joint mechanism. A wedge of soft rubber inserted into the heel of this device compresses with weight bearing allowing the forepart of the foot to come in contact with the floor during initial stance, thus simulating plantar-flexion. A wooden or metal keel running forward from the ankle position to the mid-forefoot region is shaped to permit the leg to roll forward over the forefoot during the stance phase. A flexible forefoot allows roll-over at the end of stance.

The single axis ankle joint and foot, sometimes referred to as the conventional foot, is attached to the prosthesis by a hinge which allows 15-20° of plantar-flexion and 5-8° of dorsi-flexion. Both movements are limited by the action of rubber bumpers placed respectively anterior and posterior to the ankle axis. A pliable forefoot allows the toe section of the foot to bend during terminal stance.

Since the mechanism by which each of these devices attempts to substitute for ankle movement differs, it is conceivable that the adaptation required in the more proximal joints and muscles may also differ. Previous research, comparing these two devices, has indicated that there are minimal differences in the temporal parameters and in the knee joint angle during the gait cycle (Doane and Holt, 1983; Culham et al, 1984). The patterns of activity in the proximal musculature however have not been investigated. The purpose of this study therefore was to evaluate the effect of the terminal prosthetic component on the electromyographic activity of the quadriceps and hamstring muscle groups during gait.

Subjects

Subjects who met the following criteria were recruited for the study: unilateral below-knee amputation, suitability for fitting of a temporary PTB prosthesis, relatively pain-free stump with no skin abrasions, and residency within the London, Ontario area. Eight male and two female subjects participated in the study. They were between the ages of 32 and 79 years with a mean age of 61 years. Mean height and weight were 174 cm and 75.5 kg respectively. Peripheral vascular disease was the cause of amputation in nine of the subjects and trauma in the tenth.

Methodology

All subjects were fitted with a temporary PTB prosthesis with various methods of suspension. The terminal devices, the SACH foot and the single axis-ankle joint and foot, were alternately allocated at the time of prosthetic fitting so that five of the subjects were using the SACH foot initially and five were using the single axis foot. Subjects then received routine gait training by a physiotherapist. When able to ambulate independently for a distance of at least 10m subjects were evaluated in the Locomotion Laboratory, Department of Physical Therapy, University of Western Ontario, London, Canada. The mean time from surgery to initial evaluation in the laboratory was 5.5 months.

Following evaluation of gait using the first terminal device, the second was applied and the appropriate alignment changes were made by a prosthetist. Subjects were then allowed a minimum of one week to adjust to the change in terminal device. They then returned to the laboratory for evaluation of their gait when using the second terminal device. The mean time between the first and second evaluation was 24 days.

Data collection

Electromyographic activity of the vastus lateralis and medial hamstrings was recorded bilaterally using Beckman, silver-silver chloride surface electrodes, 16mm in diameter, with an active electrode size of 9mm. The skin of the area to which the electrodes were attached was shaved when necessary, and cleansed thoroughly with gauze soaked in alcohol to remove the superficial layer of dead skin and protective oils. The electrodes were filled with gel and attached to the skin, in a bipolar fashion along the long axis of the muscle, using adhesive electrode collars. Interelectrode distance was kept constant by placing electrodes such that the adhesive collars met but did not overlap. A ground reference electrode was attached to the skin over the lateral femoral condyle.

A system of contact closing footswitches* was used to relate the electromyographic activity of these muscles to components of the gait cycle. Three pressure sensitive switches were taped to the sole of the subjects' shoes; at the heel, head of the 5th metatarsal and the great toe. Closure of a switch or series of switches resulted in an electrical signal indicating contact of any part of the foot with the floor.

The footswitch and electromyographic data were collected as subjects walked along the walkway at their preferred cadence. The signals were transmitted from the subject to a processing and recording unit by an FM telemetry system.** The EMG signals were

* Model T4-025, Mountain West Alarm Company, Pheonix, Arizona.

** Conestoga Medical Electronics, Waterloo, Ontario.

subsequently amplified, full wave rectified and filtered to provide the linear envelope representative of the EMG signal. Both the footswitch signals and the linear envelopes were recorded on a Gould Brush 8 channel ink recorder.

Analysis

The EMG and footswitch data were visually inspected. Three consecutive strides, which appeared most representative of the subject's gait, were chosen for analysis. Each stride was divided into 10% intervals with initial contact representing 0% of the gait cycle. The lowest level of EMG activity recorded over the three strides was used as a reference point. The amplitude of the EMG linear envelope in microvolts above this baseline was then determined at each 10% marker. The data from the three consecutive strides were averaged to obtain a representative EMG pattern over one stride for each muscle.

In order to permit between subject comparison of EMG activity levels, the microvolt measurements were transformed into relative amplitudes. The method of normalization used was similar to that reported by Knutsson and Richards (1979) and Yang (1984). The average peak amplitude of the three strides for each subject was given a value of 100% and the mean values at the remaining points in the gait cycle were expressed as a percentage of this mean peak amplitude. Mean values for all subjects were then calculated at each 10% point of the gait cycle and this data was subjected to statistical analysis.

A Dunns Multiple Comparison procedure was used to test whether there were significant differences in EMG activity at any point in the gait cycle when the two terminal devices were compared.

Results

To facilitate interpretation of the electromyographic data, a summary of the temporal parameters of gait, with the subjects using each of the terminal devices is presented in Tables 1 and 2. This data, as well as knee joint angle measurements throughout the gait cycle has been previously reported (Culham et al, 1984).

† Model 2800, Gould Inc., Instrument Systems Division, Alan Crawford Associates, Mississauga, Ontario.

Table 1. Comparison of mean values and standard deviations — velocity, cadence and stride length

Parameter	SACH foot	Single axis foot	p value
Velocity (metres/minute)	34.37±10.43	32.87±8.46	.501
Cadence (steps/minute)	70.60±9.98	67.10±7.19	.107
Stride length (metres)	1.173±.26	1.81±.19	.230

Table 2. Mean values of the temporal parameters of gait expressed as a percentage of the gait cycle

	Prosthetic limb			Contralateral limb		
	SACH foot	Single axis foot	p value	SACH foot	Single axis foot	p value
Stance	65.68	63.63	.041	70.73	70.79	.941
Swing	34.32	36.37	.041	29.14	29.21	.934
Single limb support time	29.14	29.21	.934	34.32	36.37	.041
Initial double limb support time	23.30	21.95	.267	13.14	12.42	.393
Terminal double limb support time	13.14	12.42	.393	23.30	21.14	.267
Total double limb support time	36.44	34.37	.036	36.44	34.37	.036

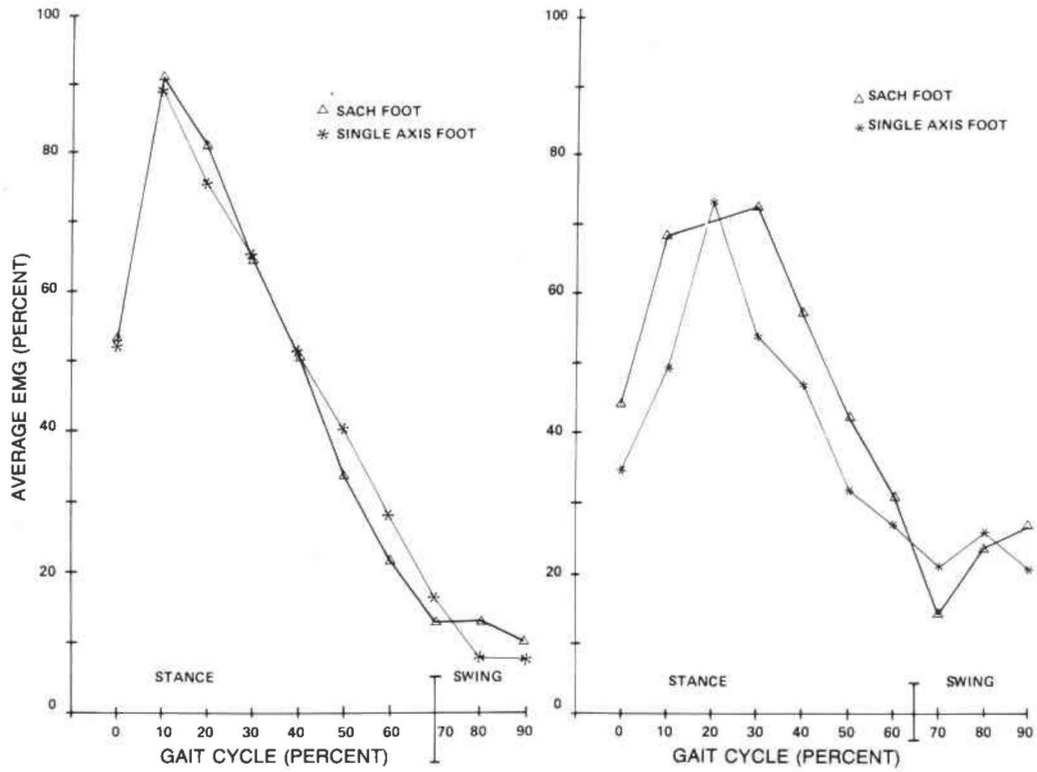

Fig. 1. Mean EMG pattern of vastus lateralis expressed as a percentage of the mean peak amplitude (N=10). Left, non amputated limb SACH foot: 100% = 149.7 microvolts. Single axis foot: 100% = 154.6 microvolts. Right, prosthetic limb SACH foot: 100% = 105.2 microvolts. Single axis foot: 100% = 122.5 microvolts.

The patterns of quadriceps (vastus lateralis) muscle activity of the contralateral (non-amputated) limb were very similar for the two terminal devices and did not differ statistically at any point in the gait cycle (Fig. 1, left). The highest levels of activity were recorded during early stance and peak activity occurred at 10% of the gait cycle with both devices. From this point activity levels decreased sharply throughout the rest of stance and remained at relatively low levels throughout most of the swing phase.

In the prosthetic limb, peak quadriceps activity occurred later in the stance phase at 20% and 30% of the gait cycle with the SACH foot and single axis foot respectively (Fig. 1, right). Following this peak the level of activity decreased throughout the remainder of stance and early swing with both devices.

The level of EMG activity did not differ significantly at any point in the gait cycle when the two terminal devices were compared.

The highest levels of hamstrings activity for the contralateral limb were recorded during early stance with both terminal devices (Fig. 2, left). Peak activity occurred at initial contact with the SACH foot and at 10% of the gait cycle with the single axis foot. From this point, activity levels fell sharply and relatively low levels of electromyographic activity were observed during late stance and initial and mid swing. The patterns were very similar and there were no statistically significant differences in the level of activity at any point in the gait cycle.

In the prosthetic limb, differences in the pattern of hamstring muscle activity were observed during the stance phase when the terminal devices were compared (Fig 2, right). When the SACH foot was used, the hamstrings

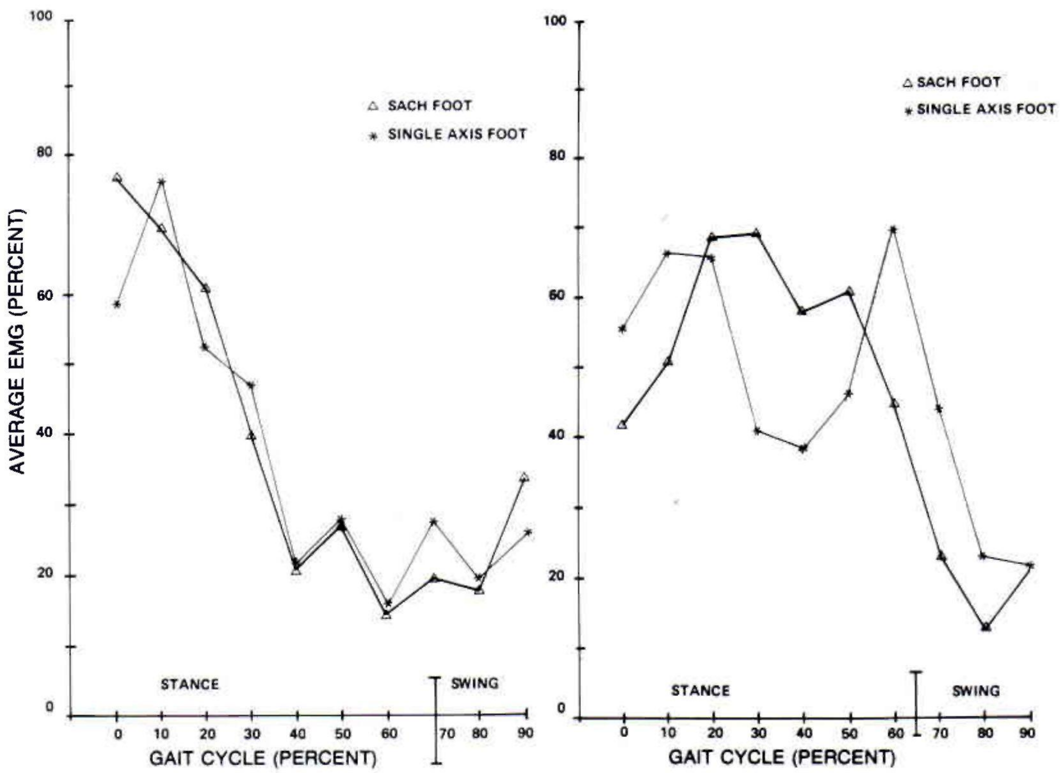

Fig. 2. Mean EMG pattern of medial hamstrings expressed as a percentage of the mean peak amplitude. Left, non-amputated limb SACH foot: 100% = 179.6 microvolts. Single axis foot: 100% = 167.6 microvolts. Right, prosthetic limb SACH foot: 100% = 115.2 microvolts. Single axis foot: 100% = 83.3 microvolts.

were active throughout the early and mid stance phase and peak activity occurred at 30 per cent of the gait cycle. The level of activity decreased from 50-80% of the gait cycle, corresponding to terminal stance and initial and mid swing. When the single axis foot was attached to the prosthesis two peaks of hamstring activity were observed, the first during early stance, at 10% of the gait cycle, and the second during the stance/swing transition, at 60% of the gait cycle. The level of activity decreased sharply during initial and mid swing. Statistically significant differences in the level of activity occurred at 30 and 60% of the gait cycle when the two terminal devices were compared ($p<.05$).

Discussion

The electromyographic patterns of the quadriceps and hamstring muscles in the contralateral limb of the amputee subjects were unaffected by the change in terminal device and the patterns, with both devices, were similar to those reported for normal healthy individuals (Battye and Joseph, 1966; Dubo et al, 1976).

Peak quadriceps activity of the contralateral limb occurred during early stance, at 10% of the gait cycle, regardless of which terminal device was attached to the prosthesis. At this point in the gait cycle the knee is flexing and the centre of gravity falls posterior to the knee joint (Perry, 1974). Quadriceps action is necessary to restrain the flexion force at the knee and thereby preserve stability (Perry, 1974).

The sharp decrease in the level of quadriceps activity following weight acceptance, observed in the contralateral limb of subjects in this study is also typical of normal gait (Sutherland, 1966; Perry, 1974). Soleus is active during early stance and functions to restrain the forward movement of the tibia over the fixed foot (Murray et al, 1978; Simon et al, 1978). As momentum carries the body forward over the stabilized tibia, the centre of gravity moves anterior to the knee joint resulting in passive knee extension. The quadriceps are no longer needed to stabilize the knee and they become relatively silent by mid stance (Perry, 1974).

A second small peak of quadriceps activity has been reported to occur at the stance/swing transition in some normal individuals (Battye and Joseph, 1966; Milner et al, 1971). It has been postulated that quadriceps are active at this time to stabilize the knee and counteract the flexion tendency created by the gastrocnemius during 'push off' (Battye and Joseph, 1966). This second peak was not observed in the contralateral limb of the amputee subjects. Absence of this second peak may be related to the relatively slow walking speed of the subjects, decreasing the requirement for reactive stabilizing forces at the knee.

The highest levels of hamstring muscle activity in the contralateral limb of the amputee subjects were recorded during early stance, 0 and 10% of the gait cycle with the SACH and single axis foot respectively. This is similar to the pattern seen in normal locomotion in which predominant hamstring activity occurs during the deceleration period of swing activity continuing into the early part of stance (Milner et al, 1971) ; Dubo et al, 1976). During initial stance the centre of mass falls posterior to the knee joint. This combined with the downward and forward moving trunk results in a flexion force at the hip and knee. Hamstrings and quadriceps act to resist this flexion tendency by stabilizing the hip and knee during this part of the gait cycle.

In some normal individuals a second peak of hamstring activity occurs at the end of stance phase (Battye and Joseph, 1966; Milner et al, 1971). This second phase of activity may be related to knee flexion in preparation for swing. Two smaller peaks were observed in the contralateral limb of subjects in this study, the first during stance, at 50% of the gait cycle, and the second at the end of stance, at 70% of the gait cycle. The second peak may be related to swing phase flexion but the function of the peak at 50% of the gait cycle is unclear.

With both terminal devices the pattern of quadriceps muscle activity in the prosthetic limb of the amputee subjects differed from that of the contralateral limb. Peak activity occurred later in the stance phase and the period of activity appeared to be prolonged regardless of the terminal device used. The delay in the occurrence of peak activity is probably related to the lengthened period of initial double limb support. Body weight was not fully accepted by the prosthetic limb until 20-25% of the gait cycle had been completed. Quadriceps were active throughout this prolonged period of weight acceptance and the highest levels of activity occurred at the beginning of the single limb support phase when the prosthetic limb was most vulnerable.

The prolongation of quadriceps activity during stance may also be related to the absence of the restraining action of the soleus muscle from initial to mid stance necessitating a greater contribution from the quadriceps in order to stabilize the flexed knee. Prolonged quadriceps femoris activity in the absence of triceps surae has been reported by other investigators (Breakey, 1976; Murray, 1978). Breakey (1976) found that quadriceps were active from initial contact to 27% of the gait cycle in subjects with below-knee amputation using a PTB prosthesis and the SACH foot attachment, compared to 0-18% reported for normal individuals. The increased duration of quadriceps activity during stance phase was felt to be related to the loss of the stabilizing effect of soleus on the knee joint.

The pattern of quadriceps activity of the prosthetic limb was also influenced by the change in terminal device. With the SACH foot attachment, the period of stance phase quadriceps activity appeared to be more prolonged than with the single axis foot although this difference was not statistically significant. In normal gait the ankle dorsiflexes to approximately 10 degrees during the mid stance phase of gait as the body is propelled forward over the stationary foot (Murray, 1967; Lamoreux, 1971). As the ankle dorsiflexes the centre of mass of the body passes anterior to the hip and knee resulting in passive extension of these joints (Perry, 1974). The SACH foot, with no ankle mechanism, does not permit dorsiflexion and possibly provides some resistance to forward progression during the mid stance phase of the gait cycle. Prolonged quadriceps action may be needed to actively extend the knee and contribute to forward movement of the body. In addition, heel rise must occur earlier with the SACH foot than it would in normal gait. Breakey (1976) found that heel rise of subjects with below-knee amputation occurred at 31% of the gait cycle compared to 43% reported for normal subjects. Early heel rise would tend to decrease stability and might contribute to the somewhat prolonged and higher levels of quadriceps activity observed during stance when the SACH foot was used.

The pattern of hamstring muscle activity in the prosthetic limb of the amputee subjects differed from that of the contralateral limb and was affected by the change in terminal device. With both terminal devices, the period of peak activity of the hamstrings was prolonged compared to the contralateral limb. This finding may be related to the longer period of initial double limb support and the increased demands placed on the hamstrings by loss of the stabilizing function of the triceps surae.

The hamstrings were active throughout the stance phase when the SACH foot was attached to the prosthesis. The level of activity at mid stance (30% of the gait cycle) was significantly higher with the SACH foot than with the single axis foot. A similar finding of abnormal hamstring activity during mid stance has been reported to occur in subjects who had undergone surgical ankle fusion (Mazur et al, 1979). The rigidity of the SACH foot may be responsible for this difference, and for the apparent co-contraction of the quadriceps and hamstrings at the mid stance point of the gait cycle. Additional hamstring muscle action may be necessary to actively extend the hip and overcome the resistance to forward progression offered by the SACH foot. Early heel rise of the SACH foot may also necessitate additional stabilization of the hip at mid stance.

A more phasic pattern of hamstring activity was observed when the single axis foot was attached to the prosthesis. Two phases of activity were observed, the first during stance and the second at the stance/swing transition. Hamstring activity during early stance would be necessary to control the postural demands created by the flexed hip and knee as the limb accepts a weight bearing function.

The second peak of hamstring activity, with the single axis foot, occurred at 60% of the gait cycle and at this point the level of activity was significantly greater than with the SACH foot attachment. This phase of hamstring activity may function to actively flex the knee in preparation for swing. The average mass of the single axis foot was 1.298kg greater than the SACH foot. This difference may have contributed to the higher levels of activity observed with the single axis foot at the end of the stance phase. Previous research has demonstrated that foot mass had no effect on temporal parameters of knee joint angle (Godfrey et al, 1977). Patterns of electromyographic activity were not investigated.

Conclusions

The patterns of quadriceps and hamstring muscle activity of the contralateral (non-amputated) limb of the amputee subjects were similar to normal and were not influenced by changing the terminal device.

Muscle activity patterns in the prosthetic limb differed from those observed in the contralateral limb and the patterns were influenced by the terminal device used. There appeared to be more co-contraction of the quadriceps and hamstrings during the mid stance phase of the gait cycle when the SACH foot component was attached to the prosthesis. The additional muscle activity may be necessary compensation for lack of ankle movement in this device.

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