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Below-knee amputee gait with dynamic elastic response prosthetic feet: A pilot study

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Abstract—The purpose of this pilot study was to investigate some of the new dynamic elastic response (DER) prosthetic feet compared to the SACH foot and determine if any demonstrated trends of producing the most optimum gait. We investigated the gait of five below-knee amputees while wearing four different DER feet (Flex-Foot, Carbon Copy II, SEATTLE, STEN) and a standard SACH foot. Each subject used each foot for 1 month prior to in-depth gait analysis and energy expenditure testing at the Pathokinesiology Laboratory. Minimal differences in either free or fast walking were noted between the five feet. The Flex-Foot resulted in significantly different gait kinematics at the "ankle" compared to the other four feet, however this foot did not produce an increased velocity nor an improved energy cost. The results of this pilot study indicated that during free or fast-paced walking on level ground there were no clinically significant advantages of any one of the feet tested. Based on this pilot data, recommendations are made for future studies including appropriate sample size.

Key words: *amputees, gait, energy cost, prosthetics.*

INTRODUCTION

In the past several years, there have been numerous advances in the development of prosthetic feet for the amputee. The desire of amputees to participate in sports, and the high demands of athletics, have resulted in the development of prosthetic feet with more dynamic action than the conventional SACH foot which for years has been the industry standard. Many of the new designs are reported to store energy during stance and release energy as body weight progresses forward, thus helping to passively propel the limb (2,7,16). These newly designed feet are referred to as dynamic elastic response (DER) or energy storing feet (33).

The gait of amputees has been well documented, and has been shown to deviate from normal gait (4-6,9,10,27,32). In the past, there have been few options for prosthetic foot choices, and therefore, the foot-type was not a concern to the investigators. Several of the new feet have been well described in the literature (2,7,24,33). The differences in construction of these feet may well influence amputee gait.

Lower extremity amputees also have been shown to use a higher than normal oxygen consumption during walking, with energy cost increasing with higher level amputations (11,20,21,32). In the commercial literature for the DER feet, it is implied that the energy cost of walking is reduced when wearing these feet (8,13,17,30).

There is little objective data to support the use of DER feet by amputees for everyday walking. These new feet tend to be expensive, and some require special expertise for proper alignment (7,24,33).

The purposes of this pilot study were: 1) to compare some of the new DER feet with the SACH foot and determine if any trends were demonstrated for producing a more optimal gait; and, 2) to identify the appropriate variables to measure and sample size to use in a larger, comprehensive study. We hypothesized that the DER feet would improve the gait of below-knee amputees over that of the SACH foot.

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METHODS

Subjects

Five male below-knee amputees (three traumatic and two dysvascular) from the Long Beach Veterans Affairs Medical Center (VAMC) STAMP clinic participated in the study (**Table 1**). All five participants were independent community ambulators; none used assistive devices. Each subject had displayed volume stability of his residual limb for at least 30 months. All subjects consented to participate following explanation of the procedure and review of the informed consent form (as approved by the Institute Review Board) after which they signed the Rights of Human Subjects form. Following completion of the study, the subjects were able to choose one of the five feet tested to retain on a permanent basis.

Socket Design

Each subject was fitted with a new prosthetic socket by an experienced prosthetist at the Long Beach VAMC STAMP clinic. This socket was used throughout the study. None of the subjects required additional socks or significant socket modifications during the study.

Foot Selection

Five different prosthetic feet were tested in a random order: Flex-Foot,[1] Carbon Copy II,[2] SEATTLE,[3] STEN,[4] and SACH.[4] In order to insure the fitting of appropriate foot components and keel, each manufacturer was provided with the subject's age, weight, height, contralateral shoe size, activity level, and amputation level and length. The selection of the SACH foot heel wedge was based on the subject's weight according to the developer's guidelines (19). The appropriateness of each foot component/keel selection was then confirmed or modified at the time of prosthetic fitting. In every case, the manufacturer's guidelines for the selection of each of the different feet provided a biomechanical function which satisfied each subject and the prosthetist.

Alignment

Alignment of the first foot, randomly selected from the five foot-types, followed established prosthetic principles. In an effort to maintain control of as many variables as possible, the prosthetic alignment was not altered with each successive foot. The Vertical Fabrication Jig[5] was used to duplicate each alignment precisely when more than the interchange of a foot-bolt and foot was required.

Every alignment and fitting was made with a minimum of two board-eligible prosthetists and at least one certified prosthetist in attendance. Each foot fitting and alignment met with the approval of the subject, prosthetist, and clinic team.

Table 1.
Subject information.

Subject	Age (yrs)	Years post-amp	Cause of amputation	Stump length (in)	Weight (lbs)	Height (ft/in)
RF	57	2.5	dysvas	8.0	234	5′11″
EJ	43	20	trauma	7.5	158	5′9″
JL	39	19	trauma	4.0	247	5′9″
MR	58	3	dysvas	6.0	208	5′8″
RZ	45	12	trauma	4.0	259	5′11″

Each subject was given an accommodation period of approximately one month to adjust to each prosthetic foot. After this period of familiarization, the prosthesis and alignment was rechecked and the subject then went to the Pathokinesiology Laboratory at Rancho Los Amigos Medical Center for instrumented gait analysis. Thus, each subject was tested once a month over a 5-month period. The testing procedure was identical during each session.

Instrumentation

Gait analysis was done during self-selected free and fast-paced walking over a 10 meter level walkway with the middle 6 meters used for data collection. Thus, the acceleration and deceleration of walking was not included in the recorded data. A Stride Analyzer[6] with compression-closing footswitches taped to the soles of the subject's shoes calculated stride characteristics and foot-floor contact pattern.

Electromyographic (EMG) activity of the vastus lateralis, long head of the biceps femoris, and the gluteus maximus was recorded with 50 micron wire electrodes inserted into each muscle with a 25 gauge needle using Basmajian's technique (3). Electrode placement was confirmed by electrical stimulation of the muscle through the indwelling electrode, and by voluntary muscle contraction. The EMG signal was telemetered from the subject to the data collection computer by means of an FM/FM telemetry system.[7] The system bandwidth was 150 to 1000 Hz with an overall gain of 1000.

Sagittal plane motion of the pelvis, thigh, knee, and "ankle" was measured with an in-house two-dimensional video motion analysis system utilizing a Sony camera[8] and Apple II+[9] microcomputer. Film speed was 60 frames per second. Reflective markers were placed at the sacrum, anterior superior iliac spine, posterior superior iliac spine, greater trochanter, lateral femoral condyle, mid-tibia in line with the knee and ankle joint axis, fifth metatarsal head, and the lateral calcaneus. The latter four markers were placed on the prosthesis with the bony landmarks estimated from the intact side.

Sagittal plane torque data of the hip, knee, and "ankle" of the amputated limb were determined. Reflective markers were taped on the greater trochanter as a reference of the hip joint, lateral femoral condyle for the knee joint, and tip of the lateral malleolus to represent the ankle joint. The latter two markers were placed on the prosthesis with the location of the ankle joint estimated from the sound limb. The subject walked across a Kistler piezoelectric force plate[10] (41 × 61 cm) located in the middle of the walkway. Force plate position was not revealed to the subjects to eliminate targeting. Vertical, fore-aft, and medial-lateral ground reaction forces were recorded. A successful trial was one in which only the prosthetic foot landed fully on the force plate. The vertical and fore-aft force components indicated the direction and magnitude of the force vector. The reflective markers allowed determination of lever arm lengths at each joint. Using in-house software, the Apple II+ microcomputer was used to calculate sagittal plane demand torques from the force vector and lever arm data.

Footswitch and EMG data, and the three components of the ground reaction force recorded from the force plate were digitally acquired on a DEC PDP 11/23 computer[11] at a sampling rate of 2500 Hz. All data were printed out in analog form for visual analysis.

Energy expenditure at rest and during a 20-minute self-selected free-paced walk was monitored. The subject was fitted with three precordial electrocardiograph electrodes to monitor heart rate; a compression closing heel switch was taped to the bottom of one shoe to record stride frequency (converted to cadence); and a harness with a telemetry system was strapped to the subject. Extended from the harness was a mouthpiece with two one-way valves; air entered one side and expired air exited the opposite side. A hose connection from the mouthpiece to a modified Douglas bag allowed collection of expired air. A nose clip was placed on the subject to prevent nasal exhalation. A 60.5-meter outdoor level track was used for the walking test. Each meter of the track was marked for monitoring the distance travelled.

Respiration rate, heart rate, and cadence were recorded via telemetry on a strip chart recorder.[12] The gas samples were analyzed for carbon dioxide and oxygen content.[13] Temperature of the gas sample was monitored by a thermistor placed in the sample flow line. The volume of the collected expired air was measured by evacuating the collection bag through a gas flow meter.[14]

Procedures

Following intramuscular electrode insertions of the selected muscles, a maximal manual muscle test was performed for each muscle. Next, data were collected while the subjects walked at a self-selected free and fast pace. Footswitch, EMG, and motion data were collected simultaneously. After the walking trials, another maximal manual muscle test was recorded to ensure electrode placement. The indwelling electrodes and motion markers were removed and the subject again walked at a free and fast pace to collect the force plate and torque data.

An energy cost study was conducted either on the day prior to the above testing or the following day. This included a recording of energy expenditure at rest after the subject had been seated for 30 minutes, and fully instrumented at rest for 5 minutes. Following the rest period, a 20-minute free walk was completed. Individual gas samples were collected, and heart rate, respiration rate, and cadence (only during walking) were recorded during the last 2 minutes of the 5-minute rest period and at minutes 3 to 5, 9 to 10, 14 to 15, and 19 to 20 during walking. For the walking trial, distance travelled during each collection period and the total distance walked were monitored by the investigator.

Data Management

The digitally collected electromyographic data were rectified and integrated and reported as a percentage of the EMG recorded during a maximal manual muscle test (%MMT). The stance phase of each stride of data collected over the 6-meter walkway was normalized to 62 percent of the gait cycle in order to average data from multiple strides and different subjects. The EMG data were further analyzed by identifying duration and intensity of EMG during each of the subphases of the gait cycle (footswitch data indicated initial double-limb, single-limb, and terminal double-limb support which were used to indicate the phases)(22).

Foot-floor contact patterns recorded by the footswitches were hand measured from the printed record to determine the duration of heel-only contact and the end of foot-flat contact (heel-off).

The fore-aft shear (x) and vertical (z) ground reaction forces recorded from the force plate were used to calculate the center of pressure and ground reaction vector. The coordinates of the x and z forces on the force plate were monitored by the Apple II+ computer. Travel of the center of pressure was mapped from initial contact with the heel of the foot (0 percent foot length) to the end of stance (100 percent foot length) and graphed as percent of foot length versus percent of gait cycle.

The vector data and motion data from the reflective markers were used to calculate torque during walking. Torque data (ft-lbs) were reported in anatomical units (a.u.) which were derived by dividing the torque by the subject's leg length (feet) and weight (pounds). In this way, data from different subjects could be compared.

Joint torque data and motion data were analyzed to identify the maximum and minimum torque or degree-of-motion in each subphase and the point (as a percentage of the gait cycle) at which it occurred.

To calculate the energy expenditure at rest and for the 20-minute walk, the carbon dioxide and oxygen content, volume expired, and the temperature of the collected gas sample were used to calculate oxygen consumption (converted to STPD). Body weight (in kilograms) was used to convert the oxygen consumption to milliliters of O_2 consumed per kilogram-minute. For the walking data, body weight and velocity (meters/minute) were used to determine milliliters of O_2 per kilogram-meter.

Data Analysis

Statistical analyses were performed using BMDP statistical software.[15] All data were analyzed for normality of distribution using the Shapiro and Wilk's W statistic. Differences between the five prosthetic feet were determined either by repeated measures analysis of variance (ANOVA) with a single group (for normally distributed data) or by Friedman's two-way ANOVA for those data not normally distributed. A significance level of $p<0.05$ was used. A post-hoc Tukey test was used to find the significantly different comparisons.

The power of the tests for selected variables was determined using the mean square terms derived from the repeated measures ANOVA. When the power of the test was low, the mean square terms were also used to calculate the sample size needed in future studies to improve the power. The following equation was used (14):

$$\Phi = \frac{\sqrt{\sum_{j=1}^{P} \alpha_j^2 / P}}{\sigma_\varepsilon/\sqrt{n}}$$

where Φ is the parameter used to determine power on charts of Power Function for Analysis of Variance;

$\sum_{j=1}^{P} \alpha_j^2$ is the sum of squared treatment effects; n is the number of subjects; σ_ε^2 is the error variance.

Table 2.
Stride characteristics, free-paced walk.

	Mean (standard deviation)		
	Velocity (m/min)	**Stride length (meters)**	**Cadence (steps/min)**
SACH	66.9 (15.5)	1.35 (0.14)	98.1 (12.6)
Flex-Foot	69.3 (17.6)	1.40 (0.16)	98.0 (13.8)
STEN	70.4 (12.3)	1.41 (0.09)	99.8 (11.6)
SEATTLE	72.0 (12.6)	1.42 (0.11)	100.7 (10.8)
CC II*	73.2 (13.0)	1.42 (0.09)	102.3 (11.4)
Normal@	81.6 (9.41)	1.51 (0.14)	108.2 (9.1)

*(CC II = Carbon Copy II)
@Waters RL, Lunsford BR, Perry J, Byrd R: Energy-speed relationship of walking: Standard tables *J Orthop Res* 6:215-222, 1988.

RESULTS

Stride Characteristics

Of the 16 stride characteristics recorded—velocity, cadence, stride length, gait cycle duration; for both limbs: initial and terminal double-limb support as a percentage of the gait cycle (%GC), single-limb support %GC, stance %GC, duration of heel only contact (%GC), and time of heel-off (%GC)—only cadence and gait cycle duration showed a statistically significant difference between the five feet during free velocity walking (mean velocity = 70.3 ± 2.66 m/min). During 6 meters of indoor walking, cadence was greater (and gait cycle duration shorter) for the Carbon Copy II foot than for the SACH or Flex-Foot ($p = 0.02$) (102 steps/min versus 98 for the SACH and Flex) (**Table 2**).

There were no significant differences in any stride characteristics measured during fast paced walking (mean velocity = 87.4 ± 12.2 m/min).

With only five subjects, the power of our testing of velocity was 0.40; we would detect a difference only 40 percent of the time when one exists. The data from this pilot project indicated that we would need 14 subjects to increase the power to 82 percent. To detect a difference in stride length, the sample size should be 25 subjects. The power of testing to detect changes in cadence during free velocity walking was 0.65. Increasing our sample size by one more subject (for a total of six) would improve the power of our testing to 0.82.

Despite the different foot-types investigated, there was an asymmetry of stance duration under all prosthetic conditions. The duration of stance was significantly longer for the sound limb ($x = 66.3 \pm 2.18$ %GC) versus the amputated limb ($x = 63.1 \pm 1.88$ %GC) for all feet ($p < 0.0001$).

Joint motion

During both free and fast gait, the only difference detected in pelvic, thigh, knee, or ankle motion was the maximum dorsiflexion angle achieved during late stance. The Flex-Foot resulted in greater dorsiflexion ($\bar{x} = 19.8 \pm 3.3$ degrees) compared to all the other feet tested ($\bar{x} = 13 \pm 4.2$ degrees) ($p=0.003$). Although the Flex-Foot had less plantarflexion during loading response (2.4 ± 5.7 degrees) than the other feet (7.5 ± 3.7 degrees), this was not a statistically significant difference (**Figure 1**).

There was a small amount of knee flexion in loading response ($x = 6.3 \pm 10.4$ degrees) under all conditions. Mean peak swing knee flexion was 63.8 ± 7.7 degrees during initial swing (**Figure 2**).

Joint torque in gait

During free walking, the maximum dorsiflexion torque occurring at the ankle joint (in anatomical units, a.u., normalized to body weight and leg length) during stance was greater for the Flex-Foot (19.9 ± 7.5 a.u.) than for the other four feet (10.4 ± 2.0 a.u.) ($p=0.002$). There were no differences between any of the other feet (**Figure 3**).

At the knee, the ground reaction force vector remained slightly anterior to the knee joint axis, thus maintaining a minimal magnitude torque throughout stance during free-paced walking. Only with the SACH foot was there a small flexion torque at the knee (2.3 ± 5.4 a.u.) during loading response. Using the other four feet, there tended to be an extension torque throughout loading. With all foot types, the knee torque approached zero by the end of loading response (17% GC) or in midstance (Flex-Foot), with an increase in extension torque in terminal stance. In preswing (50% to 62% GC) there was only a very small knee flexion torque ($x = 0.9 \pm 0.7$ a.u.) (**Figure 4**).

The hip torque demonstrated a fairly normal pattern (28) with a flexion torque in loading response progressing to a maximum extension torque in terminal stance. The Flex-Foot trial had a slightly increased hip flexion torque in loading compared to the other four feet, but the difference did not reach statistical significance (**Figure 5**).

Fast walking elicited greater dorsiflexion torque in preswing with the Flex-Foot (17.3 ± 2.3) compared to

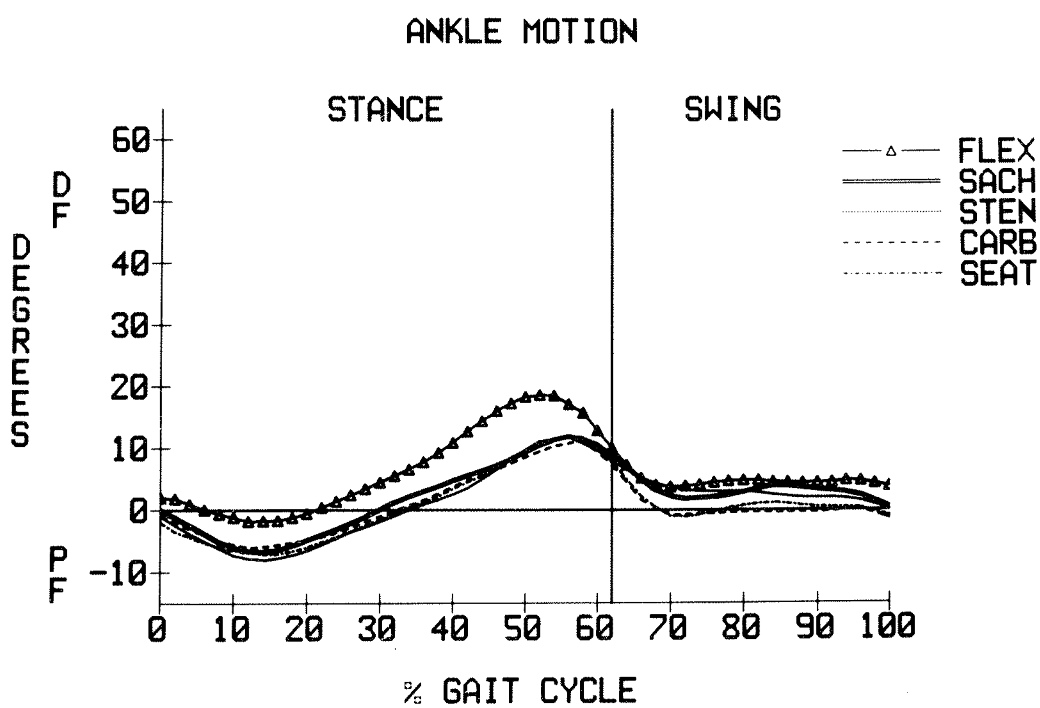

Figure 1.
Ankle motion during free-paced walking. DF = dorsiflexion; PF = plantarflexion. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

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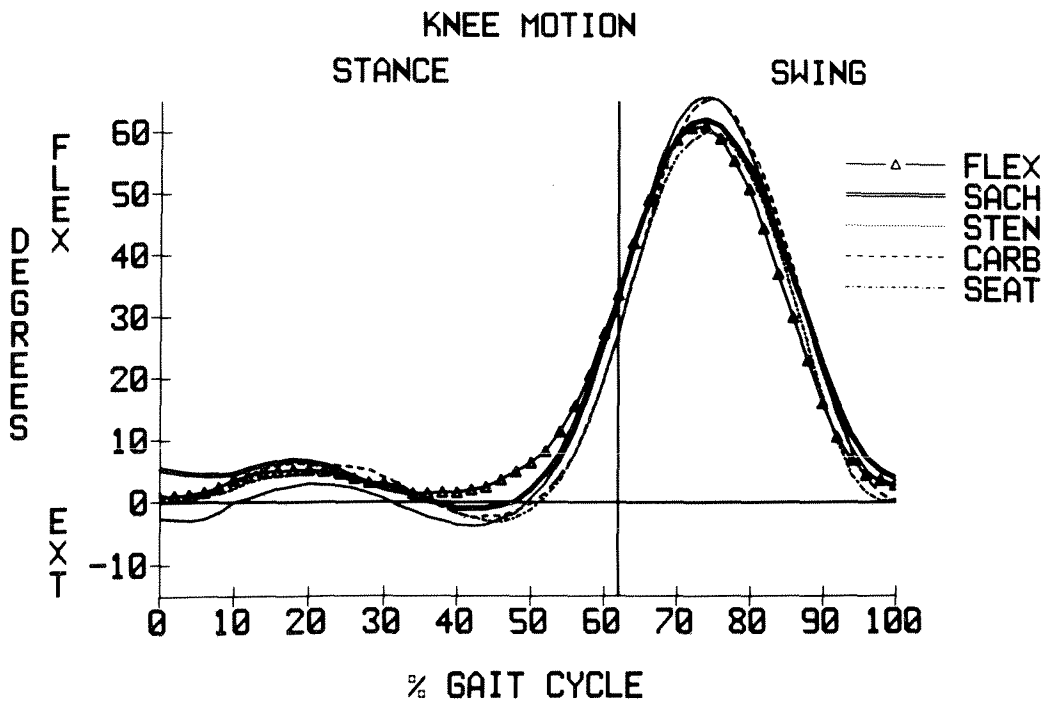

Figure 2.
Knee motion during free-paced walking. Y-axis: flex = flexion; ext = extension. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

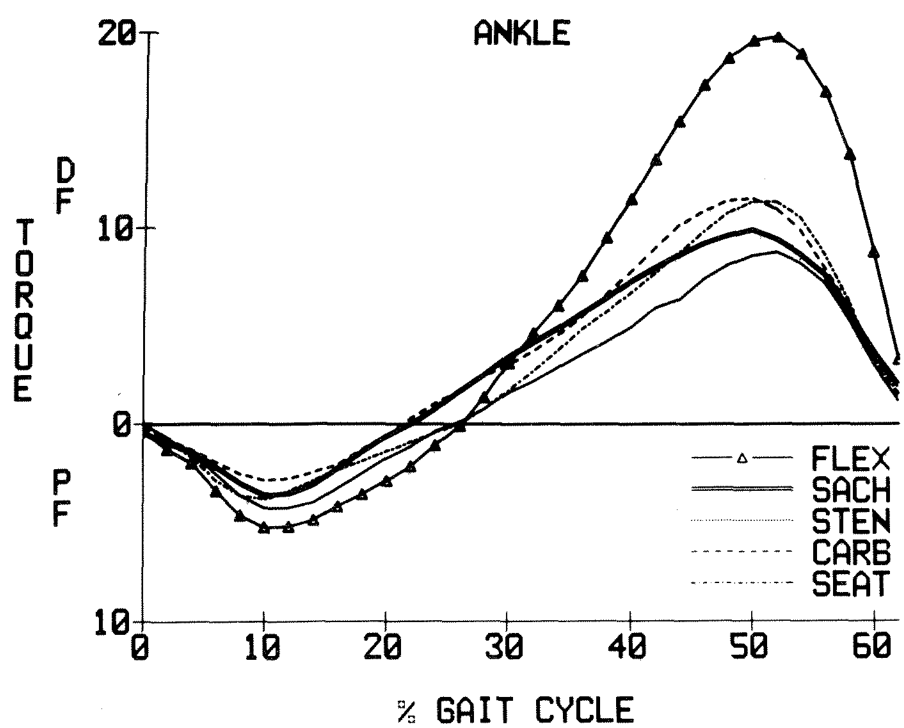

Figure 3.
Sagittal plane ankle torque (expressed in anatomical units, see text for explanation) during free-paced walking. DF = dorsiflexion, PF = plantarflexion. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

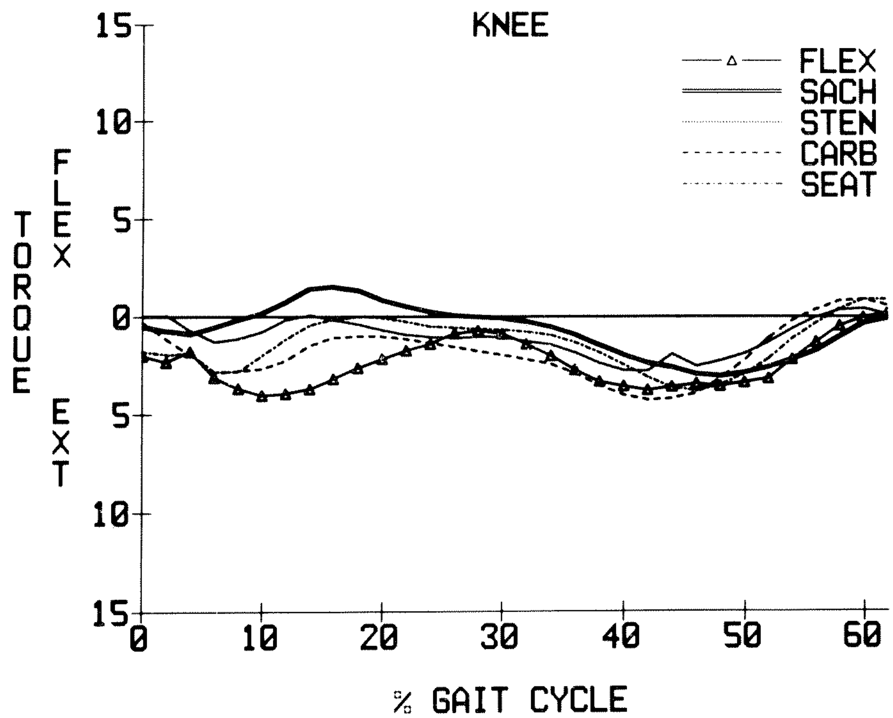

Figure 4.
Sagittal plane knee torque (in anatomical units) during free-paced walking. Y-axis: flex = flexion, ext = extension. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

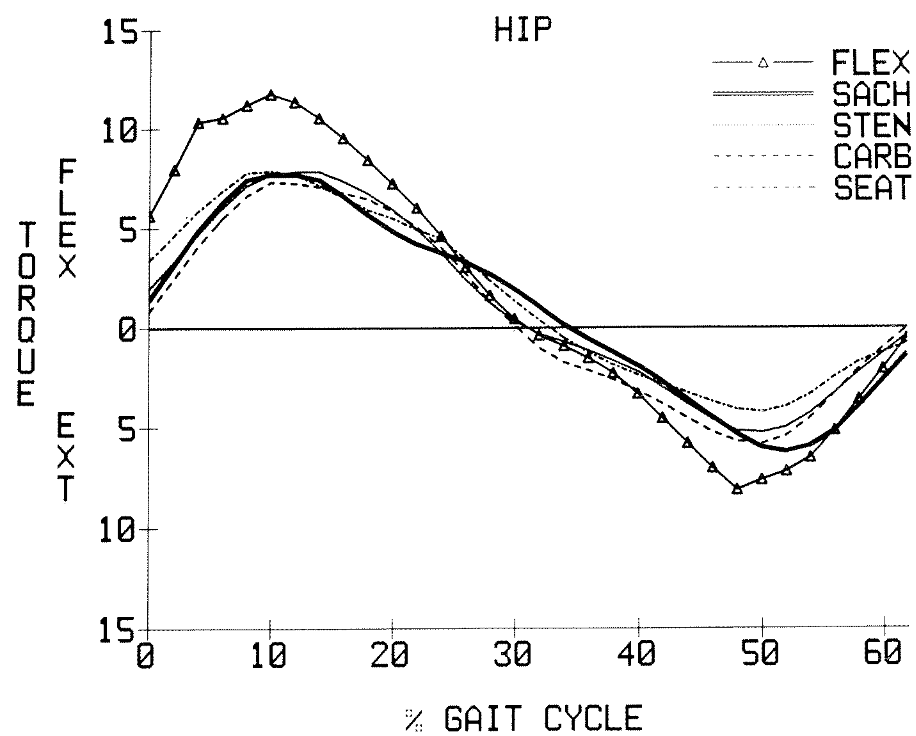

Figure 5.
Sagittal plane hip torque (in anatomical units) during free-paced walking. Y-axis: flex = flexion, ext = extension. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

only the SACH (8.5 ± 1.9) and STEN (9.1 ± 1.1) (p=0.006) (**Figure 6**). There was a small flexion torque at the knee during loading response with the STEN (3.3 ± 3.2 a.u.), SACH (2.0 ± 7.7 a.u.), and SEATTLE foot (1.0 ± 4.9), while there was an extension torque throughout loading response with the Carbon Copy II and Flex-Foot (**Figure 7**).

The hip torque pattern during fast walking was similar to that of free walking with slightly higher torque values for all feet except the SACH (**Figure 8**).

Force plate data

The maximum and minimum values of the medial-lateral shear force, fore-aft shear force, and vertical force during stance for each foot type were analyzed. For each subject, all of the feet resulted in similar force patterns. The second peak of vertical force in terminal stance ranged from 97.6 ± 8.3 percent body weight with the Flex-Foot to 99.5 ± 4.9 percent body weight with the SACH foot (**Table 3**).

Progression of the center of pressure

During single-limb support, the rate of progression of the center of pressure was more rapid with the Flex-Foot than for any of the other feet (p=0.0001). The other four feet tested had a more rapid progression during terminal double-limb support (preswing) than the Flex-Foot (p=0.0002) (**Figure 9**).

EMG of vastus lateralis, gluteus maximus, biceps femoris long head

For all five foot-types, there were no significant differ-

Table 3.
Vertical ground reaction force (as percent body weight) during free walking.

	Means (standard deviations)		
	Max., loading	Min., midstance	Max., terminal stance
SACH	110.4 (5.7)	78.0 (13.3)	99.5 (4.9)
Flex-Foot	104.2 (13.1)	73.9 (12.4)	97.6 (8.3)
STEN	104.8 (8.7)	76.9 (9.8)	97.8 (3.6)
SEATTLE	101.0 (9.0)	79.5 (5.9)	98.0 (9.5)
CC II	103.9 (8.7)	77.7 (13.4)	98.2 (6.5)
Normal*	111.0 (7.0)	74.0 (13.0)	112.0 (7.0)

*Chao EY, Laughman RK, Schneider E, Stauffer RN: Normative data of knee joint motion and ground reaction forces in adult level walking. *J Biomech* 16:219-233, 1983.

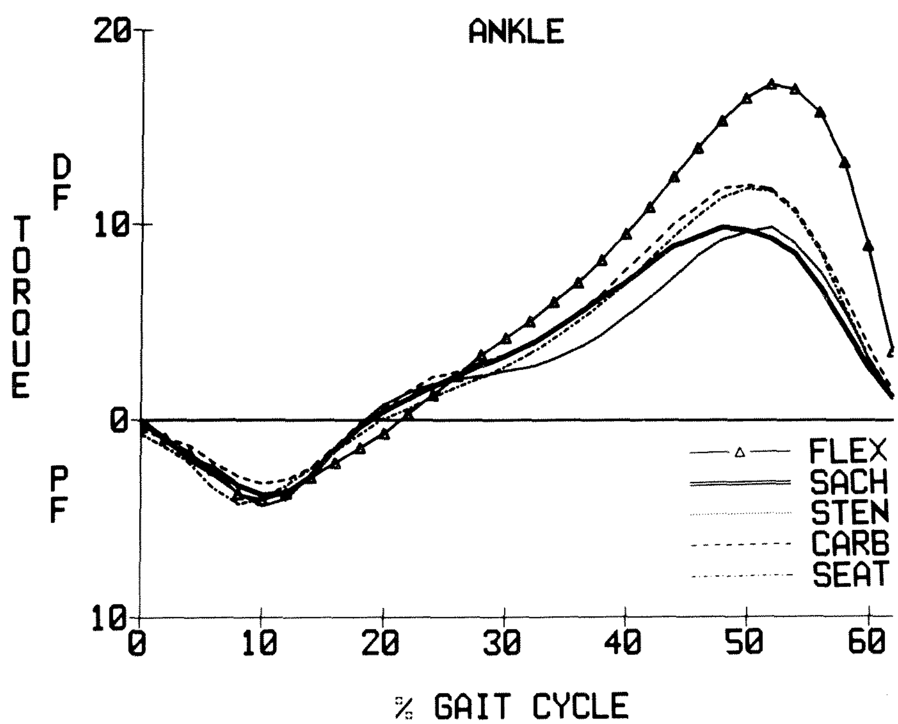

Figure 6.
Sagittal plane ankle torque (expressed in anatomical units) during fast-paced walking. DF = dorsiflexion, PF = plantarflexion. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

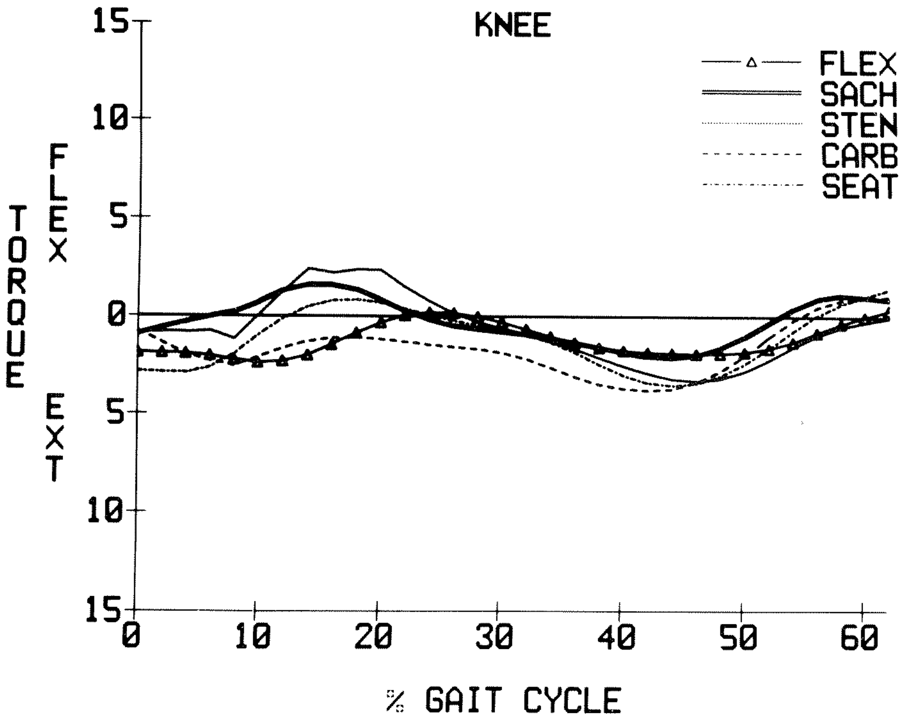

Figure 7.
Sagittal plane knee torque (in anatomical units) during fast-paced walking. Y-axis: flex = flexion, ext = extension. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

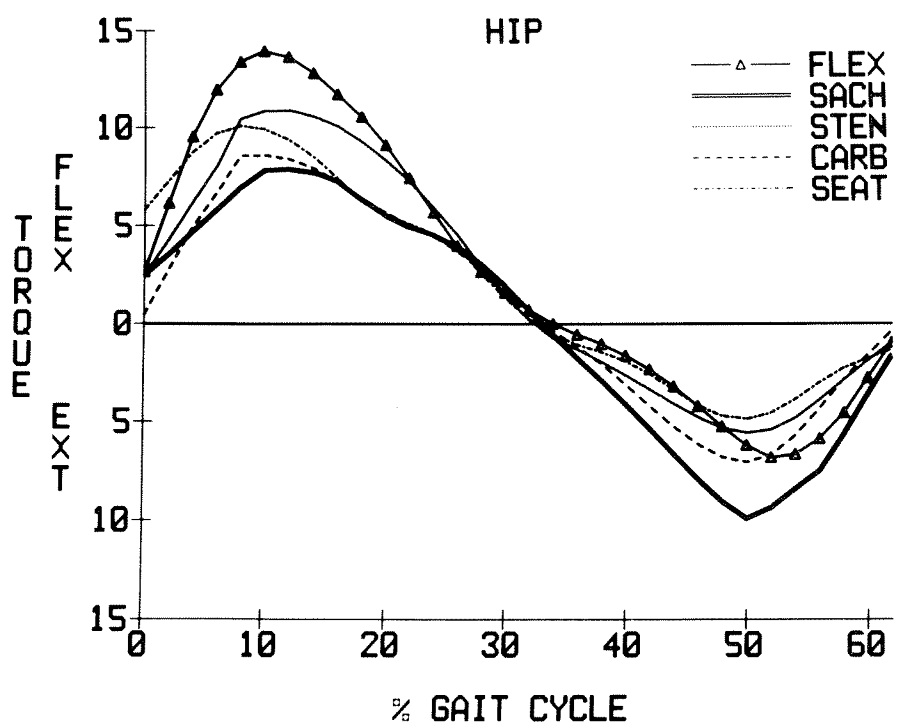

Figure 8.
Sagittal plane hip torque (in anatomical units) during fast-paced walking. Y-axis: flex = flexion, ext = extension. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

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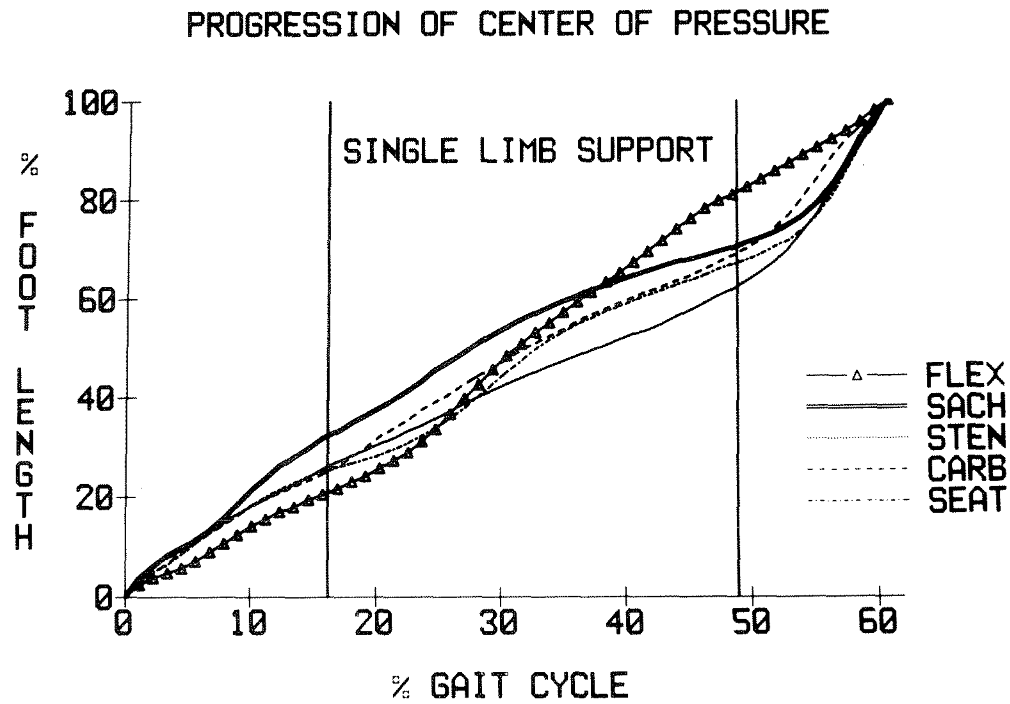

Figure 9.
Progression of the center of pressure during free-paced gait. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

ences elicited in intensity or phasing of activity of the muscles tested. However, all had prolonged activity in stance compared to normal (1,15). The long head of the biceps femoris had the greatest variability between foot-types of the three muscles examined (**Figures 10,11,12**).

Energy cost

Analysis of resting heart rate, respiration rate, and oxygen consumption revealed no significant differences between test days.

There were no differences between foot-types in energy cost (ml O_2/kg-min or ml O_2/kg-meter) during the 20-minute free walk. All foot-types resulted in oxygen consumption greater than normal (**Table 4**). Also, foot-type had no effect on the increase in oxygen consumption between the first (5-minute) and last (20-minute) sample.

The total distance walked during the 20 minutes was not affected by the type of prosthetic foot used.

Our pilot data indicate that a 5-minute walking test is not sufficient to elicit differences in energy expenditure between prosthetic feet. The within-subject variance (mean square error term) was greater than the variance due to foot-type (**Table 5**). Thus, the type of prosthetic foot worn had no effect on energy expenditure.

At 10 minutes of free walking, five subjects was not an adequate sample size to measure a difference between prosthetic feet. Twenty subjects would be needed to determine differences in energy efficiency (ml O_2/kg-meter) with a testing power of 0.86.

The minute-15 data for oxygen consumption per kg-meter had a power of 0.82. Thus, we can confidently say the type of foot had no influence on energy expenditure at 15 minutes of free walking.

Subject response

The five subjects all preferred the dynamic elastic response feet (Flex-Foot, STEN, SEATTLE, Carbon Copy II) over the SACH. The two dysvascular amputees (the two eldest) both chose the Carbon Copy II foot for their prosthesis following completion of the study. The three traumatic amputees all chose the SEATTLE foot (**Table 6**). Some of the subjects stated that the appearance of the foot influenced their choice; some preferred the realistic style of the SEATTLE foot, while others preferred the smooth foot. In addition, except for the one subject who had a SEATTLE foot prior to the study, each subject chose that foot which gave them the greatest velocity during free walking.

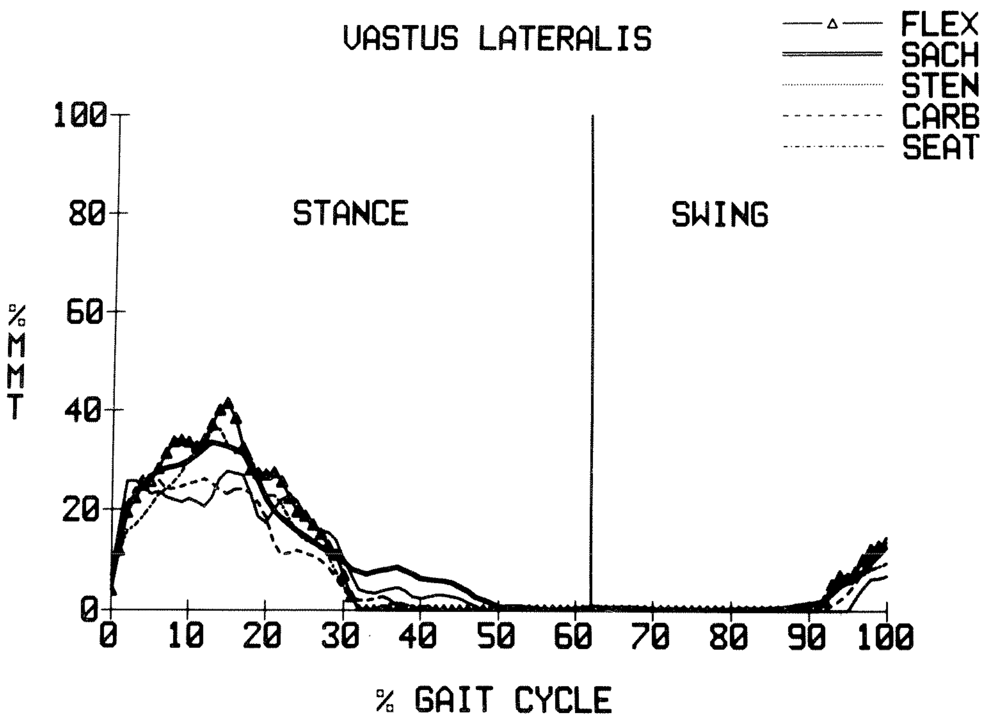

Figure 10.
Vastus lateralis electromyographic activity during free gait. %MMT = EMG as a percentage of the maximal manual muscle test level. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

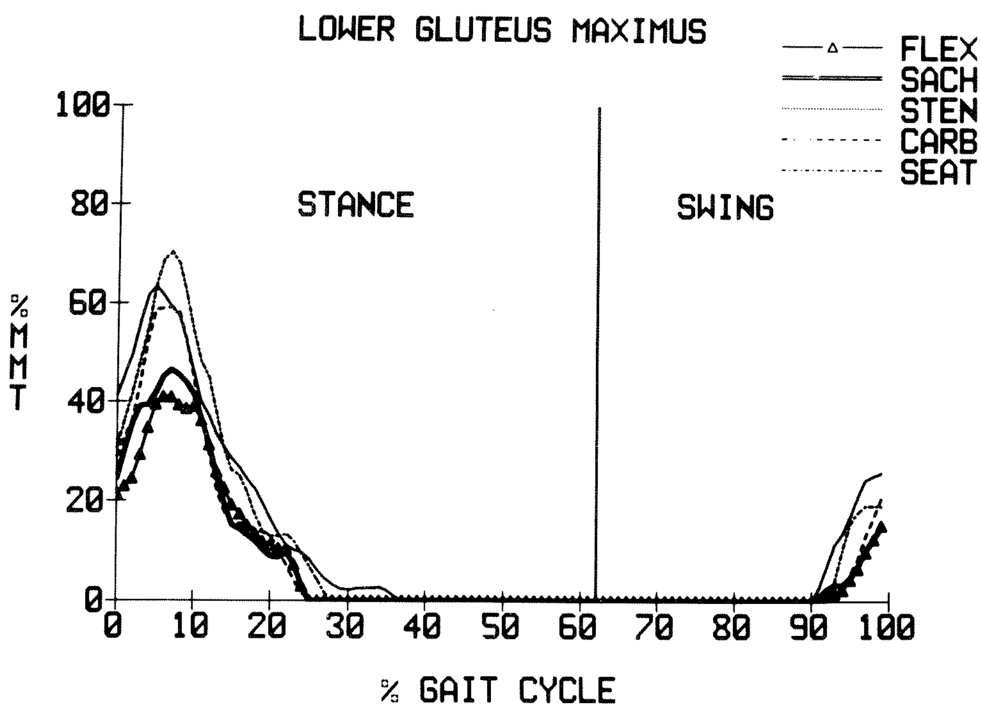

Figure 11.
Lower gluteus maximus electromyographic activity during free gait. %MMT = EMG as a percentage of the maximal manual muscle test level. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

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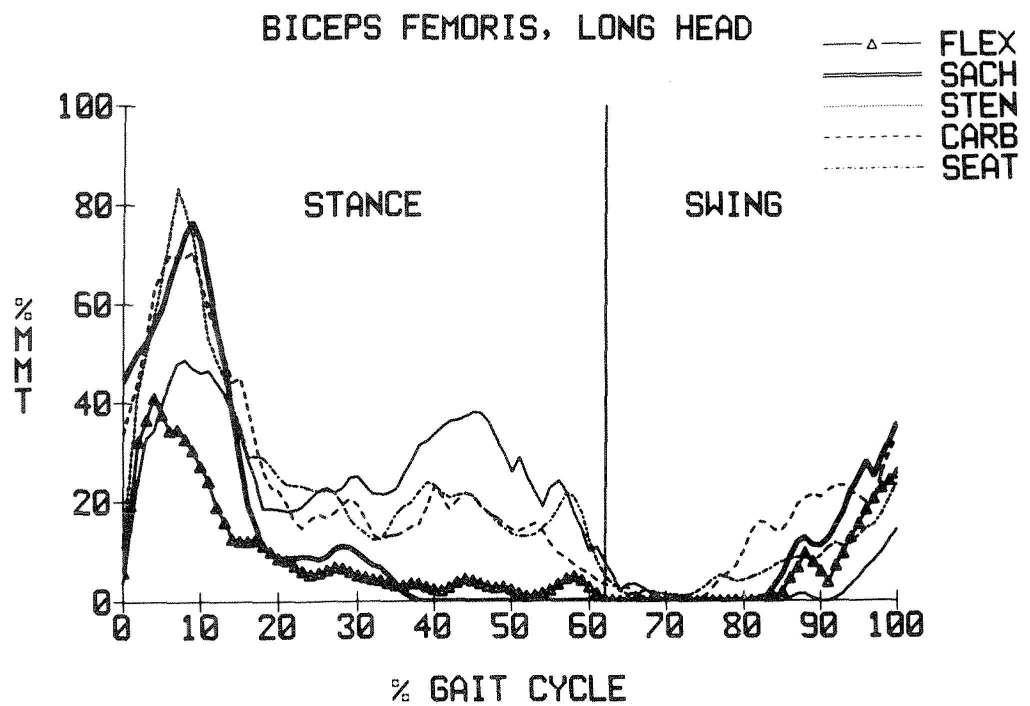

Figure 12.
Long head of the biceps femoris electromyographic activity during free gait. %MMT = EMG as a percentage of the maximal manual muscle test level. (SEAT = SEATTLE foot; CARB = Carbon Copy II.)

DISCUSSION

We anticipated the Flex-Foot, Carbon-Copy II, SEATTLE and, perhaps, the STEN demonstrating differences in gait parameters compared to the SACH. However, only the Flex-Foot stood apart from the others for a few parameters. This finding is not surprising because the construction of the Flex-Foot is much different from the other four feet tested in this study. The Flex-Foot has a graphite composite keel that extends to the prosthetic socket, whereas the other four feet were all attached to a rigid pylon at the ankle (7).

It is the authors' feeling that there exists a general opinion in prosthetic clinics that in optimum circumstances each of these DER prosthetic feet might receive their own individual alignment dictated by the unique physical design of each foot. However, no objective data have yet been obtained which support any such alignment changes for these DER feet. If alignment changes were allowed in this study, it would not be possible to identify whether the gait differences measured between the feet were the product of the prosthetic foot design, or the result of alignment changes.

The advertising literature of the dynamic elastic response feet claim an energy-storing mechanism to help propel the limb forward. This would be to replace the forefoot rocker (23). Other than the Flex-Foot, the "ankle" joints of the prosthetic feet tested were rigid. Plantarflexion was simulated during loading of the prosthetic limb as body weight compressed a cushion heel. As body weight progressed forward, ankle dorsiflexion was simulated by the mobility of the forefoot. It was primarily in the forefoot area that the prosthetic designs differed (7,33).

Stride characteristics of both free and fast gait did not change when different feet were worn. Only cadence of free-paced walking was slightly greater with the Carbon Copy II (102 steps/min) as compared to the Flex-Foot and SACH (both 98 steps/min) ($p<0.05$). This difference (4 percent) is similar to the day to day variability demonstrated in normal subjects (3 percent) and thus is not a clinically significant difference (25). Wagner *et al.*, in their investigation of the SACH foot versus the Flex-Foot using six subjects also found no difference in stride data between these two feet (31). Our pilot data indicated that in order to detect differences in velocity at least 14 subjects are required, and 25 subjects are needed to determine changes in stride length between five different foot types.

None of the feet used improved the symmetry of stance-swing ratios of the sound and amputated limb. The physical asymmetry of the below-knee amputee, resulting in asymmetrical gait, was not compensated for by any of the differently designed foot-types. As stated by Winter and Sienko, perhaps symmetry should not be a goal for the ambulatory amputee; "rather, a new non-symmetrical

Table 4.
Oxygen consumption during 20-minute free walk.

		Means (standard deviations) Min 5	Min 10	Min 15	Min 20
SACH	rate[1]	14.5 (3.9)	14.6 (3.7)	16.5 (3.1)	16.7 (3.2)
	net[2]	0.22 (0.03)	0.23 (0.04)	0.23 (0.06)	0.23 (0.04)
Flex-Foot	rate	14.3 (4.1)	14.3 (3.9)	15.9 (3.9)	16.8 (3.6)
	net	0.21 (0.03)	0.22 (0.04)	0.23 (0.06)	0.24 (0.06)
STEN	rate	15.6 (3.5)	15.6 (3.1)	17.3 (3.2)	18.8 (3.9)
	net	0.22 (0.04)	0.22 (0.04)	0.24 (0.05)	0.25 (0.07)
SEATTLE	rate	14.4 (3.0)	14.2 (2.6)	15.3 (2.3)	16.9 (1.3)
	net	0.21 (0.03)	0.20 (0.03)	0.21 (0.05)	0.23 (0.03)
CC II	rate	14.7 (2.9)	15.0 (2.3)	16.2 (1.7)	16.5 (1.8)
	net	0.21 (0.02)	0.21 (0.03)	0.22 (0.04)	0.22 (0.04)
Total	rate	14.7 (3.2)	14.8 (3.0)	15.4 (3.2)	17.2 (2.8)
	net	0.22 (0.03)	0.22 (0.04)	0.22 (0.05)	0.23 (0.05)
Normal[@]	rate	12.0 (2.0)			
	net	0.15 (0.02)			

1) rate = ml O_2/kg-min
2) net = ml O_2/kg-meter

[@]Waters RL, Lunsford BR, Perry J, Byrd R: Energy-speed relationship of walking: Standard tables. *J Orthop Res* 6:215-222, 1988.

optimal is probably being sought by the amputee within the constraints of his residual system and the mechanics of his prosthesis" (35).

Motion data revealed significantly greater dorsiflexion (i.e., shank versus foot angle) at the end of stance with the Flex-Foot compared to the four other feet (**Figure 1**). This agrees with the results reported by Wagner *et al.* (31). Most likely, the increased range was due to the flexibility of the Flex-Foot shank.

The increase in dorsiflexion range was accompanied by an increase in dorsiflexion torque in terminal stance (**Figure 3**). Fast-paced walking minimized the differences between the DER feet such that the Flex-Foot dorsiflexion torque was greater only than the SACH and STEN feet (**Figure 6**).

The increase in dorsiflexion torque may be analogous to the loading of a spring that is thought to provide the dynamic elastic response. However, there were no other indications of an elastic response from the Flex-Foot.

The peak vertical force in terminal stance is suggested to represent the push-off of the stored energy prosthetic foot (18,31,34,35). In this sample population, none of the feet, including the more flexible Flex-Foot (18,33), created a significantly greater terminal stance vertical force (**Table 3**).

As the body weight progressed forward, the Flex-Foot yielded with greater dorsiflexion and had a more rapid progression of body weight (center of pressure) over the foot during single-limb support (**Figures 1** and **9**). The other four feet had a more rapid progression of the center of pressure following contact of the sound limb (i.e., during preswing). Because none of our subjects chose the Flex-Foot at the end of the study, perhaps this rapid progression of body weight during single-limb support was perceived as instability by the amputee and not as an optimal characteristic.

The combination of excessive dorsiflexion and rapid forward progression of body weight while wearing the Flex-Foot is similar to that seen in patients with weak plantarflexors (26,29). It appears the flexible shank of the Flex-Foot does not provide adequate control of tibial motion. The excessive dorsiflexion was not accompanied by

Table 5.
Summary table of repeated measures ANOVA 5-minute energy expenditure.

A. $\dot{V}O_2$ ml/kg-minute

	Sum of squares	d.f.	Mean square	F	Tail prob.
Foot type	5.62229	4	1.40557	0.50	0.73
Error	44.73190	16	2.79574		

B. $\dot{V}O_2$ ml/kg-meter

	Sum of squares	d.f.	Mean square	F	Tail prob.
Foot type	0.00104	4	0.00026	0.57	0.69
Error	0.00722	16	0.00045		

Table 6.
Subject selection of foot type following completion of the study.

Subject	Cause of amputation	Foot type pre-study	Foot type post-study
RF	dysvas	SACH	Carbon Copy II
EJ	trauma	SEATTLE	SEATTLE
JL	trauma	SACH	SEATTLE
MR	dysvas	SACH	Carbon Copy II
RZ	trauma	SACH	SEATTLE

increased knee flexion as is generally seen with a weak calf. Thus, the mobility was absorbed in the Flex-Foot shank and was not translated to the intact knee joint of the below-knee amputee.

The Flex-Foot had a greater dorsiflexion torque and motion at the ankle joint. However, this did not result in any significant differences in torque or motion at the hip or knee joints. The knee torque was minimal throughout stance, indicating the amputee is attempting to create less demand at the knee. The heel rocker effect of the prosthetic foot tends to create a flexion torque at the knee (12). But, through a forward trunk lean and prolonged muscular activity, the ground reaction vector is directed anterior to the knee joint axis creating an extension torque.

The electromyographic activity of the muscles examined (vastus lateralis, gluteus maximus, and long head of the biceps femoris) was similar for all foot types. The absence of lower-leg musculature put increased demands on the upper leg muscles to control the limb in stance. Thus, there was prolonged duration of electromyographic activity in stance compared to normal (1,15) (**Figures 10, 11** and **12**).

The three muscles examined were active primarily during early stance to provide stability. If the DER feet truly assisted with propelling the limb forward, perhaps the muscles participating in early swing, such as the rectus femoris, iliacus, and short head of the biceps (23) would reveal a difference in activity in gait with the DER feet compared to nondynamic feet.

The Flex-Foot company now makes a Flex-Walk that uses a standard rigid pylon with their unique Flex-Foot (8). This rigid pylon structure is like that used with the other feet and the differences we observed in the Flex-Foot would most likely be less with the Flex-Walk.

Although there were slight differences in the mechanics of walking with the Flex-Foot as compared to the others, the oxygen consumption of free walking was not affected by foot type. The name given to these feet, energy storing, and the commercial literature provided by the companies implies or states improved endurance for walking (8,13,17,30).

The average energy expenditure for the first 5 minutes of walking, both as per unit of time (14.7 $\pm$ 3.2 ml O_2/kg-minute) and per unit of distance (0.21 $\pm$ 0.03 ml O_2/kg-m), of all five conditions agrees with that reported in the literature for below-knee amputees during 5-minute trials (11,20,21,32). In most of the studies, the authors did not specify foot-type and it must be assumed the standard SACH foot was used.

Nielson *et al.*, in their preliminary report state there is a decrease in energy cost during gait with the Flex-Foot compared to the SACH foot, although no statistical analyses were done (20). Their subjects were tested over a range of walking speeds from 1.0 mph (26.9 m/min) to 4.0 mph (107.5 m/min), for a 5-minute test and a velocity was found which was the most efficient in terms of oxygen uptake (20). Although the self-selected velocities were slower for our five subjects, our energy cost data are similar to those reported by Nielson. At 5 minutes, the oxygen consumption with the Flex-Foot was 0.21 ml O_2/kg-meter; the SACH foot had only a slightly higher VO_2 (0.22 ml O_2/kg-meter) (**Table 4**).

From our five subjects, it is apparent that these feet have no effect on energy cost during 20 minutes of free velocity walking. For the minute-15 sample, we had an 82 percent chance of detecting a difference between the feet. For the other samples (5-, 10-, 20-minute data), the within-subject variance was greater than the variance that could be attributed to foot-type. This indicated that foot-type had no effect on oxygen consumption during free walking.

Our laboratory testing revealed minimal differences between the five different prosthetic feet in this sample of five subjects. This pilot study investigated only walking on level ground, but the subjects used their prostheses under a variety of conditions. Comments included such things as an improvement in walking on ramps and stairs, and just a general feel that the swing limb was propelled forward.

All our subjects reported a preference in ambulatory function with the DER feet compared to the SACH foot. Following completion of the study, the subjects tended to choose that foot which gave them the greatest velocity even if statistically there were no differences. This suggests that very subtle changes in gait may be detected by the amputee, and these changes are perceived to be significant.

We included a diverse sample of amputees in this pilot project in order to identify variables which may reveal differences between the feet in a larger group of typical below-knee amputees. Further investigation is continuing with a sample of 20 below-knee amputees: 10 traumatic and 10 dysvascular. A larger population may bring out differences in some of those variables (e.g., velocity) where our sample size was too small to have sufficient power of testing. Also, we will look at the traumatic versus the dysvascular amputee to determine if these groups respond differently to the different prosthetic conditions.

Included in the continued study will be analysis of ambulation on stairs and ramps as well as level walking. Activity of the muscles contributing to early swing will be examined and the ground reaction forces and motion of the sound limb will be analyzed.

Since the subjects' final selection of a foot-type for long-term use was influenced by the physical appearance of the foot, the subjects in our continuing study will be kept blind to the foot-type being tested. All foot coverings will be identical to eliminate subjective bias.

CONCLUSIONS

Of the five prosthetic feet tested, only the Flex-Foot resulted in a change in gait dynamics during level walking. The Flex-Foot created greater dorsiflexion motion and torque at the end of stance as compared to the other feet. This difference, however, was not translated to an increased velocity nor an improved energy expenditure during free walking.

From our laboratory testing, no clinically significant changes in gait could be detected in below-knee amputees while wearing five different prosthetic feet. However, our subjects did notice a difference between the dynamic elastic response feet compared to the SACH. This leads us to believe we have yet to identify the specific variable that we should measure which will reveal the differences between the feet.

The results of this pilot study suggest there are no advantages of the dynamic elastic response feet for the amputee who is limited to level walking. Further investigation is needed.

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END NOTES

[1]Flex-Foot Inc., 27071 Cabot Rd, #106, Laguna Hills, CA 92653.
[2]The Ohio Willow Wood Company, PO Box 192, Mount Sterling, OH 43143.
[3]Model and Instrument Development, 861 Poplar Place South, Seattle, WA 98144.
[4]Kingsley Manufacturing Co., PO Box CSN 5010, Costa Mesa, CA 92628.
[5]VFJ-100, Hosmer Dorrance Corporation, 561 Division Street, PO Box 37, Campbell, CA 95008.
[6]Stride Analyzer: B & L Engineering, 8807 Pioneer Blvd., Unit C, Santa Fe Springs, CA 90670.
[7]Model 2600, BioSentry Telemetry, Inc., 20720G Earl Street, Torrance, CA 90503.
[8]Rotary Shutter Camera RSC-1050 and Video Motion Analyzer SVM-1010, Sony Corporation of America, 47-47 Van Dam Street, Long Island City, NY 11101.
[9]Apple Computer, Inc., 20525 Mariani Avenue, Cupertino, CA 95014.
[10]Kistler Instrumente AG, CH-8408 Winterhur/Switzerland; Subsidiary in the USA, Kistler Instruments Corp., 2475 Grand Island Blvd., Grand Island, NY 14072.
[11]Digital Equipment Corporation, 146 Main Street, Maynard, MA 01754-2571.
[12]Recorder model 302: Astro-Med, West Warwick, RI 02893.
[13]Oxygen Analyzer OM-11; Medical Gas Analyzer LB-2; Ventilation Measurement Module VMM series: SensorMedics Corporation, 1630 South State College Blvd., Anaheim, CA 92806.
[14]Collins motor #P-553-P, Warren E. Collins, Inc., 220 Wood Road, Braintree, MA 02184.
[15]BMDP Statistical Software, Inc., 1440 Sepulveda Blvd., Suite 316, Los Angeles, CA 90025.

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