Prosthetic Weight Acceptance Mechanics in Transtibial Amputees Wearing the Single Axis, Seattle Lightfoot, and Flex Foot

Jacquelin Perry, Lara A. Boyd, Sreesha S. Rao, and Sara J. Mulroy

Abstract—Loading response challenges the limb with the dual demands of accepting rapidly moving body weight to both absorb the shock of floor contact and create a stable limb over which the body can advance. Delay in achieving foot flat contact with the floor causes a prolonged period of heel only support and results in an unstable base of support for those persons with transtibial amputations. The purpose of this study was to identify mechanical causes of instability during weight acceptance with three different prosthetic foot designs, Single Axis, Seattle Lightfoot, and Flex Foot. Ten male individuals with transtibial amputations were tested on three separate occasions wearing each prosthetic foot. A comparison group of ten individuals without transtibial amputations was also examined. Mean free walking speed was significantly slower for those with transtibial amputations regardless of the prosthetic foot worn (p < 0.05). Contralateral toe off times were significantly later for each prosthetic foot (p < 0.01). The timing of peak knee flexion was found to be significantly later than normal for each prosthetic foot (p < 0.01). To minimize the impact of initial floor contact, persons with an intact limb used rapid plantar flexion, followed by a slower lowering of the foot to the floor. Dorsiflexion then stimulated knee flexion and foot flat. Two altered functions were found for all three prosthetic feet, reduced knee flexion and prolonged heel only support. Diminished knee flexion reflected delayed dorsiflexion and tibial advancement as a result of the cushioned heel. Lateness in reaching foot flat was also found. To improve the walking abilities of those persons with transtibial amputations, prosthetic foot designs need to incorporate mechanisms which promote early foot flat while preserving limb stability.

Index Terms—Gait, prosthetic foot, transtibial amputation.

I. INTRODUCTION

Each stride during walking begins with the transfer of body weight onto the forward limb. As the heel strikes the floor at initial contact, the body center of mass is far in front of the rear “supporting” limb. Consequently, weight transfer onto the accepting limb is very rapid and fairly abrupt [1], [2]. This situation creates the challenge of accepting rapidly moving body weight in a manner that both absorbs the shock of floor contact and creates a stable limb over which the body can advance.

Previous prosthetic foot research met this challenge by including a cushioned heel to absorb the shock of heel strike in the SACH foot design [3], [4]. Deformation of the cushion and the shape of the internal keel were both purported to allow the foot to gradually roll forward until it became flat on the floor and a stable weight bearing position was reached. In those people with a transtibial (TT) amputation limb stability during this period of heel only support is provided by the knee and hip extensor musculature. Any delay in advancing to foot flat subsequently increases this demand on the extensor muscles. This may be one cause of the higher energy cost of walking experienced by those with TT amputations [5].

Effectiveness of the cushioned heel as a shock absorber led to its continued use in most prosthetic foot designs. Recent attempts to reduce the high energy cost of walking by decreasing muscular demand have focused on an improved “push-off” through the design of dynamic elastic response feet [4], [6]–[9]. None of these new “energy recovering” foot designs, however, have reduced the measured energy cost of walking [5], [10]. The energy cost of walking for those with TT amputations continues to be as high as 25% greater than normal [5]. Those with TT amputations resulting from trauma are healthy and able to expend high amounts of energy. Yet they walk at speeds that are only 85% of normal [5], [11], [12]. The individual who has had a TT amputation as a result of dysvascular disease, and is limited by reduced strength and energy, only attains a walking speed from 40 to 65% of normal [5], [12]–[14]. These slowed walking rates represent a balance between energy expenditure and speed of progression [15].

Electromyographic recordings have identified the muscular activity of weight acceptance as a major contributor to the high energy cost of ambulation for those with TT amputations [17]–[19]. Active throughout this task are the knee extensors (quadriceps) and the hip extensors (gluteals and hamstrings). The gait of those with TT amputations demands a prolonged period of action by these muscles in order to meet the dual needs of stability and forward progression. In addition, the intensity of the muscular activity of those with TT amputations, as result of either traumatic and dysvascular origin, equals or exceeds normal levels despite the reduced walking speed of both groups.

To more distinctly identify the factors causing instability throughout the loading response phase and contributing to the excess energy cost in those with TT amputations a detailed study of prosthetic foot mechanics was undertaken. The purpose of this study was to identify mechanical causes leading to an unstable base of support during weight acceptance with three different prosthetic foot designs, the Single Axis foot (SA), the Seattle Lightfoot (SL), and the Flex Foot (FF).
II. METHODS

A. Subjects

Ten male individuals with TT amputations participated in this study (mean age = 62.4 y, range 49–72). All had a completely healed, unilateral TT amputation and displayed residual stump volume stability. Each was capable of independent ambulation. Subjects were recruited from the Long Beach Veterans Administration Medical Center’s prosthetic service (Long Beach, CA). Each subject was diagnosed with vascular disease as a result of diabetes mellitus prior to inclusion in this study.

All subjects were fitted with a prosthesis that allowed interchange of the foot components. In a random fashion, subjects were tested on separate occasions wearing each prosthetic foot. Each foot was worn for approximately one month prior to testing in order to allow the subject to accommodate to its design. Prosthetic fabrication and alignment were performed by the same certified prosthetist at the Long Beach Veterans Administration Medical Center.

A comparison group of ten individuals (five male and five female, mean age = 51.1 y, range 34–67) without TT amputation also were tested. An informed consent statement was signed by each participant prior to inclusion in this study.

B. Procedure

All testing was done at the Pathokinesiology Laboratory, Rancho Los Amigos Medical Center (Downey, CA). Test procedures were identical during each session for all groups, and each test was videotaped. All gait analyses were done at a self-selected (free) velocity over a level 10 m walkway with the middle six meters marked for data collection by photoelectric cells. This eliminated the effects of acceleration and deceleration. Two walking trials were gathered for each subject. All data were collected simultaneously.

Stride characteristics were obtained with the Stride Analyzer System. Footswitches were taped to the bottoms of the shoes of each participant. These contained compression closing switches under the heel, first and fifth metatarsal, and great toe areas. FM–FM telemetry was used to transmit signals from the sensors to the footswitch stride analyzer.

Motion analysis was performed with a six camera VICON system. Using an infrared strobe light and 14 retroreflective markers this system recorded motion about each joint in the lower extremity. Markers were taped to the subjects’ skin at the anterior superior iliac spine (bilaterally), greater trochanter, anterior thigh, medial and lateral femoral condyles, anterior shank, medial and lateral malleoli, dorsum of the foot, the first and fifth metatarsal heads, and the heel. The landmarks on the prosthetic shank and foot were estimated using the intact limb. Data were sampled at a rate of 50 Hz, filtered at 6 Hz and recorded digitally on a DEC PDP 11/73 computer. After recording the position of each VICON marker during quiet standing and a practice walk, two trials of self selected free walking were recorded. VICON motion data were processed to provide joint angles for each percent of the gait cycle.

Angular velocity of the foot, shank and thigh segments was determined by the screw displacement (helical) axis method described by Kinzel et al. (1972) [20]. Ankle and knee joint velocities were similarly computed.

A Kistler forceplate (41 × 61 cm), concealed in the center of the walkway was used to collect ground reaction forces. Only those trials in which an isolated and complete recording of the designated foot occurred were collected. Forceplate location was not revealed to the subjects in order to eliminate targeting. Consequently, multiple walking trials were occasionally necessary in order to successfully record ground reaction forces.

C. Data Analyzes

All data were averaged across the two walking trials. The stride analyzer system calculated gait velocity, stride length and cadence. Footswitch data were used to define the foot-floor contact patterns. Stance was initiated at 0% of the gait cycle (GC) and 100% marked the end of swing. To permit intersubject comparison and to allow the two walking trials to be averaged, the stance phase data of each subject was normalized to 62% GC. Time to achieve foot flat was calculated as the percent of the gait cycle when both the heel and one of the metatarsal areas contacted the ground. The period of weight acceptance was defined as the interval between initial contact and the onset of foot flat.

Joint motion in the sagittal plane at the ankle and knee was analyzed for the magnitude and timing of peak values during loading response. The rate of angular motion of each joint was calculated and analyzed for its peak velocity and the timing of this peak during the loading response phase. Joint moments normalized to body weight and limb length were determined by standard three dimensional segmental dynamics equations as described by Meglan et al. (1995) [21]. Ground reaction forces for computing joint moments were collected and used in conjunction with angular displacement data to calculate joint compliance. Joint compliance during weight acceptance was defined by the slope of the linear least squares regression between the moment and angular displacement as described Davis et al. (1996) [22].

Descriptive statistics for each prosthetic foot and the nonamputee subjects were calculated. All data were initially screened for normality of distribution. Two separate statistical comparisons were made. First, differences between each prosthetic foot group and the nonamputee participants were found. Dunnnett’s test was used to compare the nonamputee group with each of the three different prosthetic foot types. Following this, statistical comparisons were made between the three prosthetic groups. The significance of differences between each foot type was determined using a repeated measures ANOVA. When the ANOVA identified a difference between groups, post hoc Tukey’s tests were utilized to demonstrate where the statistical differences had occurred. A significance level of $p < 0.05$
was used for all statistical tests with corrections for multiple comparisons. All statistical calculations were performed on BMDP statistical software.\textsuperscript{3}

III. RESULTS

All results are first presented as comparisons to normal and followed by prosthetic group differences.

A. Stride Characteristics

When wearing each prosthetic foot tested those with TT amputations walked significantly slower than normal (SA foot 68% (\(p < 0.01\)), SL 69% (\(p < 0.01\)), and FF 71% of normal \(p < 0.05\)). This reduced gait velocity was primarily the result of significantly decreased stride length from normal for each prosthetic foot studied (\(p < 0.01\)). Cadence did not differ statistically between the three feet or versus normal (see Table I). Time to reach a foot flat position for the SL (21.0% GC) and FF (19.4% GC) were significantly later than normal (11.6% GC). Foot flat time for the SA foot (16.6% GC) did not differ statistically from normal. Contralateral toe off times were significantly later for all three prosthetic feet (SA 16.9% GC, SL 16.3% GC, FF 16.0% GC and normal 12.4% GC, \(p < 0.01\)) (see Table II).

B. Joint Motion

Peak plantar flexion during the period of loading response was found to be significantly greater for the SA foot (11.9°) than normal (7.7°) (\(p < 0.05\)). The timing of peak plantar flexion for each of the three prosthetic feet tested was not found to differ from that of the normal group (see Table III, Fig. 1). When wearing the SA foot, subjects walked with significantly less knee flexion (9.1°) than normal (18.2°) (\(p < 0.05\)). Although when wearing both the SL (12.2°) and the FF (9.7°) less knee flexion was recorded, this was not statistically different from normal. The time of peak knee flexion was found to be significantly later than normal (13.4% GC) for the SA (18.8% GC), the SL (20.0% GC) and the FF (18.8% GC) (\(p < 0.05\)). In addition, both the SL foot (4.4°) and the FF (5.7°) demonstrated significantly less plantar flexion than the SA foot (\(p < 0.01\)).

C. Angular Joint Velocity

Ankle joint velocity in the direction of plantar flexion was found to be significantly less than normal (2.4 rads/s) for the SL (1.0 rads/s) and FF (1.2 rads/s) (\(p < 0.01\)). In the direction of ankle dorsiflexion, the SL foot (0.8 rads/s) was significantly slower than normal (1.4 rads/s), while the SA (1.6 rads/s) and FF (1.1 rads/s) did not differ statistically from

\begin{table}[h]
\centering
\begin{tabular}{|l|c|c|c|}
\hline
\textbf{Group} & \textbf{Velocity} & \textbf{Stride Length} & \textbf{Cadence} \\
\hline
SA & 67.5\* & 75.6\** & 91.2 \\
 & (13.3) & (14.1) & (8.5) \\
SL & 68.5\* & 75.7\** & 93.1 \\
 & (21.2) & (7.7) & (9.4) \\
FF & 70.6 & 76.3\** & 94.0 \\
 & (11.1) & (8.8) & (7.7) \\
Normal & 88.0 & 91.1 & 97.3 \\
 & (14.0) & (6.4) & (9.3) \\
\hline
\end{tabular}
\caption{Stride Characteristics for SA, SL, FF, and Normal Groups (All Data Represented as a Percentage of Normal) Mean (STD Dev)}
\end{table}

\begin{table}[h]
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\begin{tabular}{|l|c|c|}
\hline
\textbf{Group} & \textbf{Foot Flat} & \textbf{Contralateral Toe Off} \\
 & (% GC) & (% GC) \\
\hline
SA & 16.6 & 16.9\** \\
 & (8.3) & (2.9) \\
SL & 21.0\* & 16.3\** \\
 & (7.7) & (2.5) \\
FF & 19.4\* & 16.0\** \\
 & (7.0) & (2.0) \\
Normal & 11.6 & 12.4 \\
 & (3.3) & (2.0) \\
\hline
\end{tabular}
\caption{Time to Foot Flat and Contralateral Toe Off for SA, SL, FF, and Normal Groups Mean (STD Dev)}
\end{table}
normal (see Table IV). The time of peak plantar flexion and dorsiflexion velocities were not different from normal for any of the prosthetic feet (Fig. 3). The SA (2.6 rads/s) was most similar to normal and was significantly faster than the SL and FF designs ($p < 0.01$) (see Table IV). Additionally, the SA foot was significantly faster than the SL and FF prosthetic feet ($p < 0.05$).

Knee joint velocity was significantly slower than normal (2.8 rads/s) for all three prosthetic feet (SA 1.6 rads/s, SL 1.4 rads/s, and FF 1.5 rads/s) ($p < 0.01$) (see Table IV). However, the timing of this peak did not differ among the groups (see Fig. 4).

**D. Joint Compliance**

Joint compliance during weight acceptance was defined by the slope of the linear least squares regression between the moment and angular displacement [22]. This was calculated for the knee and ankle for nine of those with TT amputations.

One subject was identified as an outlier and for this measure was eliminated from our analysis. During the loading response phase at the ankle, the SA foot (61.2°/kgm/Nm)
was significantly more compliant than normal (30.1°/kgm/Nm) (p < 0.05). At the knee, the SA foot (84.0°/kgm/Nm) was significantly more compliant than normal (16.6°/kgm/Nm) (p < 0.01). In addition, the SL (16.9°/kgm/Nm) and the FF (23.5°/kgm/Nm) were significantly less compliant than the SA foot (p < 0.01). Both the SL (31.1°/kgm/Nm) and FF (31.0°/kgm/Nm) differed statistically from the SA design (p < 0.05) (see Table V).

### IV. DISCUSSION

Detailed analysis of normal function has revealed that the motions that occurred during the period of weight acceptance provided a system to facilitate shock absorption. At this time, limb stability was preserved by selective muscle control at each joint. To minimize the impact of initial floor contact, persons with an intact limb utilized two modes of foot motion following the initiation of stance with a heel strike. The first reaction was rapid plantar flexion which reached a peak velocity of 2.4 rads/s by 2.3% of the gait cycle. Following this, the ankle moved at a slower rate to complete an arc of 7.7° of plantar flexion by 7% of the gait cycle. Then the ankle reversed its motion, toward dorsiflexion, reaching foot flat at 11.6% GC. Throughout this series of motions the heel remained the only area of foot support.

This sequence of actions suggests that the initial rapid arc of plantar flexion was a free fall of body weight on a virtually unrestrained ankle joint. Early slowing of the foot drop by the pretibial muscles (anterior tibialis and long toe extensors) provided the first shock absorbing system. Murray (1966), by tracking the path of the toe, had previously shown a pattern of rapid foot drop followed by a slower arc of motion [23]. These dual patterns of motion within the weight acceptance phase indicates two distinct demands at this time. The foot must reach a position of stable contact with the floor as rapidly as possible in order to allow weight transfer to the forward limb. However, momentum must not be lost in order to continue moving forward in an efficient manner. Shock absorbing capability of the system was demonstrated by the tissue flexibility or compliance calculations. For normal function the ratio of displacement to the controlling energy or moment, was found to be 30.1°/kgm/Nm (Table V).

The subsequent arc of controlled ankle motion toward plantar flexion preserved the heel as a rocker. By 5.5% of the gait cycle peak knee flexion velocity had been reached (2.8 rads/s). This corresponds to the time at which peak plantar flexion occurred (7.0% of the gait cycle). Progression of the body across the heel led to anterior advancement of the tibia and a second shock absorbing system, knee flexion. With inertia delaying advancement of the femur, and quadriceps modulation allowing the knee to flex, the heel rocker effect resulted in 18.1° of knee flexion by 13.4% of the gait cycle. The relative arcs of motion, 7.7° of plantar flexion, and 18.1° of knee flexion, indicate that knee flexion is the major shock absorbing system. In addition, such knee flexion allows the foot to reach a stable flat posture while the thigh is still in a trailing position.

The gait of individuals with a TT amputation revealed insufficiencies in prosthetic design which disrupted limb mechanics at both the knee and foot. Similar to previous findings, the gait velocity of those with TT amputations was slower than normal regardless of the type of prosthetic foot worn. Two altered functions also common to all of the prosthetic feet studied were reduced knee flexion and prolonged heel only support. All other deviations reflected the specific prosthetic foot design.

Prior studies have shown that loading response knee flexion of those with TT amputations is less than normal. This also was true for the subjects in the current study. With all three prosthetic feet examined, the average knee flexion arc during weight acceptance was 9–12° compared to the normal 18° (see Table III). In addition, the rate of knee motion was significantly slower and the timing of peak knee flexion was later (p < 0.01)(see Table IV). While not specifically examined, the reduced knee flexion response in those with TT amputations may reflect a delay in tibial advancement. This may be caused by the cushioned heel, which was used in all three foot designs. With heel strike applying the compressive force on the broad posterior margin of the wedge shaped cushion, there would be greater height loss posteriorly than at the narrow anterior apex. A second possibility could be the passive inhibition of rapid knee flexion by the proximal brim of the prosthesis.

Structurally, the SA prosthetic foot is a freely moving hinge fitted with rubber bumpers to limit the functional arc. There is a wedge shaped cushion heel for primary shock absorption.
Gait recordings showed that ankle motion began with an arc of plantar flexion not statistically dissimilar to the normal initial foot drop. Foot fall angular velocity for the SA prosthesis reached 2.6 rad/s by 3.3% of the gait cycle. This resulted in a total plantar flexion arc of 11.9° by 8.4% of the gait cycle. The much greater joint flexibility, as identified by its high compliance of 61.2 kgm/Nm indicates that the total plantar flexion arc was an uncontrolled fall until bumper restraint occurred. This strongly contrasts with the normal early deceleration provided by the pretibial muscles.

With the SA foot the combined effects of bumper restraint and progression over the heel resulted in rapid reversal of ankle motion toward dorsiflexion. There was sufficient ankle plantar flexion, however, to allow foot flat at 16.6% of the gait cycle. The posterior ankle bumper provided a less effective heel rocker and resulted in significantly slowed (1.6 rad/s) and late knee joint angular velocity (9.8% of the gait cycle, \( p < 0.01 \)) (see Table IV). This was a notable contrast to the angular velocity of knee flexion of 2.8 rad/s at 5.5% of the gait cycle demonstrated by those without amputations.

Instability imposed by free motion at an insensitive joint is another factor leading to reduced knee flexion. In an effort to control this unstrained motion at the ankle those wearing the SA foot restricted and delayed motion at the knee. Peak knee flexion was only 9.1° for those wearing the SA foot (versus 18.2° for normal). Additionally, this motion peaked at 18.8% of the gait cycle, a point well after foot flat contact had been established. Willingness to delay knee flexion and subsequent forward progression until a stable foot position had been achieved demonstrates the difficulty created by a functional disassociation between the foot and shank segments in the SA design.

The SL foot has a much more rigid construction. Its “ankle joint” is a “C” shaped polypropylene mass, open anteriorly. The heel is a cushioned wedge. In response to heel strike, only a minimal arc of ankle plantar flexion occurred (4.4°). However, the cushioned heel allowed a foot drop, with a significantly slowed (\( p < 0.01 \)) peak angular velocity in the direction of plantar flexion of 1.0 rad/s at 2.5% of the gait cycle. The limited flexibility implied by these data was confirmed by a compliance of only 16.9 kgm/Nm. Again, the heel rocker mechanics used to advance the tibia were limited, with peak angular velocity at the knee reaching only 1.4 rad/s by 6.7% of the gait cycle. These limitations of an overly stiff prosthetic foot significantly prolonged the period of heel only support during weight acceptance. This was demonstrated by the delayed time of foot flat until 21% of the gait cycle.

Design of the FF provides two areas of mobility within the hindfoot, a plantar blade and a cushioned heel. In addition, there also was a flexible shaft component. These design differences resulted in a compliance value that was most similar to normal (23.5 kgm/Nm). The rate of initial ankle angular velocity toward plantar flexion was significantly less than normal (1.2 rad/s at 2.7% of the gait cycle, \( p < 0.01 \)). This resulted in a smaller arc of motion, 5.7°. Initiation of tibial advancement and resultant angular velocity at the knee showed the greatest limitation (1.5 rad/s at 8.0% of the gait cycle). While its knee angular velocity was most similar to the SL foot, the effect was slowed. The time in the gait cycle when foot flat was reached also was late (19.4% of the gait cycle). Similar to the SL foot a large delay was noted between contralateral toe off and foot flat. Prolongation of the period of heel only support caused instability during weight acceptance and led to reduced and delayed knee flexion.

In summary, the SA foot provided the most timely foot flat posture but its free arc of plantar flexion motion created instability as it had no intrinsic restraint to decelerate the rate of foot fall prior to bumper contact. Also, the SA foot has no material to moderate the subsequent rate of dorsiflexion prior to terminal contact with the anterior bumper, a second source of instability. Both the SL and FF include dorsiflexion control but were stiff, and shock absorbing knee flexion was reduced.

V. CONCLUSIONS

The motions used during weight acceptance are designed to provide shock absorption and preserve forward momentum at the expense of passive stability. In response dynamic stability is provided by compensatory actions of the hip and knee extensor muscles. The mechanics of each prosthetic foot design dictated the stability available at the knee. Despite posturing to decrease knee moments when wearing the Single Axis foot the rapid arc of motion experienced generated tibial instability such that knee joint compliance values were five times higher than normal. Lack of coupling between the foot and tibial segments of the Single Axis foot created more rapid plantar flexion and dorsiflexion while these individuals were balanced on the heel. The stiffer Seattle Lightfoot and Flex Foot designs had the converse effect of markedly prolonging heel only support. This unstable posture both delayed forefoot contact and resulted in reduced forward progression during the weight acceptance period of stance.

To allow persons with a transtibial amputation to attain the stability of timely foot flat support with limited knee flexion, a greater arc of functionally restrained plantar flexion would be required. Subsequently, there must also be a means of stimulating shank advancement to preserve forward momentum.

REFERENCES

Jacquelin Perry received the Bachelor’s degree in physical education from the University of California at Los Angeles (UCLA), a certificate in physical therapy with five years experience in the Army, and the M.D. degree from the University of California at San Francisco, followed by residency training in orthopaedic surgery. In 1955, she joined the staff of Rancho Los Amigos Medical Center, Downey, CA, and has remained until present. Her practice began with reconstructive surgery of polio patients with particular emphasis on correction of the deformed spine. From this, she progressed to rehabilitation of the severely disabled with emphasis on the stroke patient. Her clinical programs also included spinal injury, arthritis, and head trauma. After medical complications removed her from the surgical theater, she emphasized rehabilitation and gait analysis. She developed a systematic approach for analyzing gait which has become an established system that is accepted worldwide. Further efforts toward gait analysis led to the development of an instrumented laboratory system with emphasis on dynamic electromyography to determine the precise timing and intensity of lower extremity muscle activity during walking. She also has employed dynamic EMG to quantify glenohumeral and scapular function both in sports and spinal cord injury. She has published extensively on gait and has written many book chapters on upper extremity muscle function. In 1992, she published a book titled, *Gait Analysis*, which is a comprehensive guide to normal and pathologic muscle activity during walking. Her research interests include the study of muscle function in spastic, arthritic, amputee, stroke, spinal cord injury, and polio patients.

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