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Stair ambulation in persons with transtibial amputation: An analysis of the Seattle LightFoot™

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Abstract—The purpose of this study was to document gait patterns in a group of individuals with transtibial amputations (TTA) during stair ambulation, and to identify the functional limitations associated with this task. Ten persons with TTA fitted with a Seattle LightFoot™ prosthetic component, and 14 nondisabled subjects participated in this study. Electromyographic activity (EMG) of the vastus lateralis (VL), rectus femoris (RF), gluteus maximus (GMAX), semimembranosus (SMEMB), biceps femoris long head (BFLH), and biceps femoris short head (BFSH) was assessed using indwelling wire electrodes during ascending and descending stairs. Lower limb kinematics (VICON) and stride characteristics (Footswitch Stride Analyzer System) also were collected. Stride characteristics revealed that those with TTA had a significantly slower rate of stair ambulation and demonstrated stance phase asymmetry between limbs compared to the nondisabled. Kinematic analysis determined significant limitations in prosthetic ankle motion, which necessitated compensatory functions at the hip and knee to accomplish stair ascent and descent and resulted in significantly greater muscular effort (increased EMG intensity and duration) compared to nondisabled.

Key words: *amputation, gait, prosthetics.*

INTRODUCTION

The increasing emphasis on salvaging the knee joint in lower limb amputations has resulted in a greater potential for this population to achieve independent ambulation. A common activity of daily living for both nondisabled individuals and those with transtibial amputations (TTA) is ascending and descending stairs, which imposes significantly increased demands on the lower limbs. During this task, the body is carried in a net vertical direction with forward translation, which results in phasic muscle action and range of motion at the joints that differ from those of level walking (1,2).

Of particular concern for the person with amputation is the inherent lack of ankle mobility common in most prostheses. As previously reported for level walking, reduced ankle dorsiflexion results in gait compensations to augment forward progression, which in turn increases the muscular effort and energy expenditure of ambulation (3,4). It would appear that limited prosthetic foot mobility would have a profound effect in negotiating stairs, as the ankle is normally subjected to relatively large moments and arcs of motion (1). One commonly used prosthetic device is the Seattle LightFoot™[1]. It, along with other dynamic elastic response feet, has been promoted as "energy storing," a characteristic that has been suggested as reducing the demands of ambulation.

While the gait of the person with a TTA during level walking is well known (5–7), only the stride characteristics for stair ambulation have been reported in this population (8). To date, no kinematic or electromyographic information is available to assess the limitations associated with this common functional task. The purpose of this study was to document the gait patterns and compensatory mechanisms characteristic of persons with TTA during stair ambulation. Since stair ambulation challenges strength and range of motion, as well as balance and muscular coordination, it is an ideal activity to explore the

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higher level functional abilities of the individual with a TTA. In addition, an understanding of gait on stairs in persons with amputation is a critical step toward improving patient management and eventual ambulatory ability. Information concerning stair ambulation also will be beneficial in designing rehabilitation programs for those individuals with TTA, providing specific information regarding the muscle action and kinematics necessary to perform this complex, balance-threatening task. Furthermore, this knowledge is essential for the design of new prosthetic components aimed at promoting increased functional independence for those with TTA. Comparisons between the muscle action, stride characteristics, and kinematics in those with and without TTA should provide a means by which to analyze the performance of those with amputations and to make suggestions designed to improve overall ambulation ability.

METHODS

Subjects

Ten males with unilateral TTA constituted the patient group (**Table 1**). Amputations in eight of these subjects were secondary to trauma, while two were the result of vascular disease. All subjects were independent community ambulators and none used assistive devices. In addition, each subject had displayed volume stability of the residual limb for at least 2 years prior to participating in this study.

Fourteen nondisabled individuals (8 male and 6 female) comprised the control group (**Table 1**). These subjects were screened for musculoskeletal or neurological impairments and were free from any conditions that would affect gait.

Table 1.
Subject characteristics-mean (SD).

	TTA (n=10)	Control (n=14)
Age (yrs)	51 (15)	34 (11)
Height (cm)	178 (7)	175 (9)
Weight (kg)	79 (11)	76 (13)
Prosthetic use (yrs)	15 (15)	—

TTA = subject with transtibial amputation.

Instrumentation

Prior to testing, all persons with amputation were fitted with a Seattle LightFoot prosthesis. Prosthetic alignment followed established prosthetic principles, with each fitting being clinically optimized and reviewed by a team of three certified prosthetists. Subjects were then given a 1-month accommodation period in which to adjust to the prosthesis. All of the testing procedures outlined were performed on the residual limb.

Muscle Action

After the adjustment period, gait analysis was performed at the Pathokinesiology Laboratory at Rancho Los Amigos Medical Center. In order to determine the relative demands of stair ambulation, the major extensor muscles of the lower limb were studied, as it is these muscles that have been previously demonstrated to show abnormalities during level walking (4). To record the timing and intensity of muscle activity of the residual limb of those with TTA, indwelling, bipolar wire electrodes were inserted into the muscle bellies of the vastus lateralis (VL), rectus femoris (RF), lower portion of gluteus maximus (GMAX), semimembranosus (SMEMB), long head of biceps femoris (BFLH), and short head of biceps femoris (BFSH) using Basmajian's technique (9). Confirmation of electrode placement was determined by mild electrical stimulation and voluntary muscle contraction. An FM/FM telemetry system[2] was used to transmit the EMG signal from the subject to a DEC 11/23 computer[3]. The system bandwidth was 150 to 1000 Hz with an overall gain of 1000. Data acquisition rate for each channel was 2500 Hz.

Digitally acquired EMG data for all gait conditions were full wave rectified and integrated over 0.01-second intervals. Intensities were reported as a percentage of the maximal, isometric manual muscle test (%MMT). Assessment of EMG timing was accomplished through the EMG Analyzer software[4]. The EMG analyzer determined the onsets and cessations for all packets of EMG that exceeded an amplitude of 5 %MMT. Packets of EMG separated by an interval of less than 5 percent of the gait cycle were combined. Mean onset and cessation times were then calculated and a time-adjusted mean profile for each muscle was obtained. All EMG onsets and cessations were reported as a percentage of the gait cycle (%GC). A gait cycle was defined as the period of time from ipsilateral initial contact to the next ipsilateral initial contact.

To allow for comparison of EMG intensity between subjects and muscles, and to control for the variability of electrode placement, EMG data were normalized to that acquired during a maximal isometric effort. The maximal

manual muscle tests were done in the standard positions as described by Kendall and McCreary (10) and were performed prior to the gait trials.

Motion

Three-dimensional (3-D) motion analysis was performed using a six-camera, VICON motion system[5]. Reflective markers placed on the sacrum, anterior superior iliac spine (bilaterally), greater trochanter, anterior thigh, medial and lateral femoral condyles, medial and lateral malleoli, anterior tibia, dorsum of the foot, fifth metatarsal head, and the posterior heel were used to determine sagittal plane motion of the lower limb. Marker placement on the prosthesis was estimated from the bony landmarks on the sound limb. All motion data were sampled at 50 Hz, filtered at 6 Hz and recorded digitally on a DEC PDP 11/73 computer[5].

In order to allow averaging of data acquired from multiple strides and subjects, motion data were processed, digitized, and normalized to a stance phase that represented 62 %GC. Joint motion in the sagittal plane at the ankle, knee, hip, and pelvis was analyzed for minimum and maximum values at each phase of the gait cycle.

Stride Characteristics

Stride characteristics and phasing of EMG activity were delineated by use of a Stride Analyzer[4]. This system consisted of insoles containing compression-closing footswitches. Sensors at the heel, first and fifth metatarsal heads, and the great toe responded to compression equal to or greater than 3 psi. Analog footswitch data were collected by the same DEC 11/23 computer[3] used to acquire the EMG signals. Stride characteristics that were measured included velocity, cadence, stride length, stance, single limb support, and initial and double limb support times.

A four-step staircase with a step height of 15 cm and a tread depth of 27 cm was used for stair ambulation. Subjects were instructed to place only one foot on each step and were allowed to practice as necessary. Individuals began with their involved limb for both stair ascent and descent, and use of the hand rail was allowed if necessary. Two trials of ascending and descending stairs at a self-selected velocity were recorded with EMG, footswitch, and motion data being collected simultaneously. All data from multiple strides and trials were averaged for each individual subject. Following the gait trials, the maximal isometric muscle tests were repeated to ensure that the intramuscular electrodes had not been displaced during testing.

All subjects with amputation were able to complete the gait testing satisfactorily. Motion and stride data were collected on all 14 of the control subjects, while EMG data were obtained for 7 of these subjects.

Statistical Analysis

Shapiro and Wilk's W statistic was used to screen all data for normality of distribution. Independent t-tests were used to compare motion and stride characteristics between groups, while paired t-tests were employed to test for differences between the sound and residual limbs of the subjects with amputation.

Comparison of EMG data between groups was made with a 2-way analysis of variance (group x muscle) with repeated measures on one variable (muscle). This analysis was repeated for EMG onset, cessation, and mean intensity. A significance level of $p<0.05$ was used for all comparisons. All data were analyzed using BMDP Statistical Analysis software[6].

RESULTS

Ascending Stairs: Stride Characteristics

The group with amputations had a significantly slower velocity during stair ascent compared to the controls (29.6 m/min vs. 33.4 m/min; $p<0.05$). In addition, the subjects with amputation demonstrated stance asymmetry between the sound and residual limbs. The mean stance time for the sound limb was 69.4 %GC, which was significantly greater than the stance time of the prosthetic limb (59.6 %GC; $p<0.01$) as shown in **Table 2.** This difference was reflected in all three of the stance subdivisions, with the greatest asymmetry occurring during single limb support (sound: 40.4 %GC vs. residual: 30.6 %GC; $p<0.01$). When compared to the sound limb, the initial double limb support time of the residual limb was significantly increased, while the terminal double limb support time of the residual limb was decreased ($p<0.05$). In contrast, the control subjects demonstrated symmetrical stance times between limbs (63 %GC), with no significant differences in single or double limb support times (**Table 2**).

Ascending Stairs: Motion

Significant differences in ankle motion were evident at the two ends of stance. Compared to intact limbs, the prosthetic foot complex allowed less dorsiflexion during initial stance (7° vs. 13°; $p<0.01$), and reduced plantar-

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Table 2.
Stride characteristics: ascending stairs—mean (SD).

	TTA Residual Limb N=10	TTA Sound Limb	p-value*	Control N=14	p-value**
Velocity (m/min)	29.6 (3.6)		—	33.4 (3.6)	<0.05
Stance (%GC)	59.6 (2.1)	69.4 (2.1)	<0.01	63.3 (2.7)	<0.01
SLS (%GC)	30.6 (2.1)	40.4 (2.1)	<0.01	36.7 (2.7)	<0.01
IDLS (%GC)	14.8 (1.2)	13.6 (1.5)	<0.05	13.0 (2.3)	<0.05
TDLS (%GC)	13.6 (1.5)	14.8 (1.2)	<0.05	117. (2.5)	<0.05

TTA = subject with transtibial amputation; *Significance between sound and residual limb of TTA; **Significance between the Control group's reference limb and TTA's residual limb; SLS = single limb support; IDLS = initial double limb support; TDLS = terminal double limb support; %GC = percentage of the gait cycle.

flexion at the end of stance with the peak difference occurring during early swing (13° vs. 5°; p<0.001) as shown in **Figure 1.** Despite these differences at the ankle, the knee motion between the groups was similar (**Figure 2**).

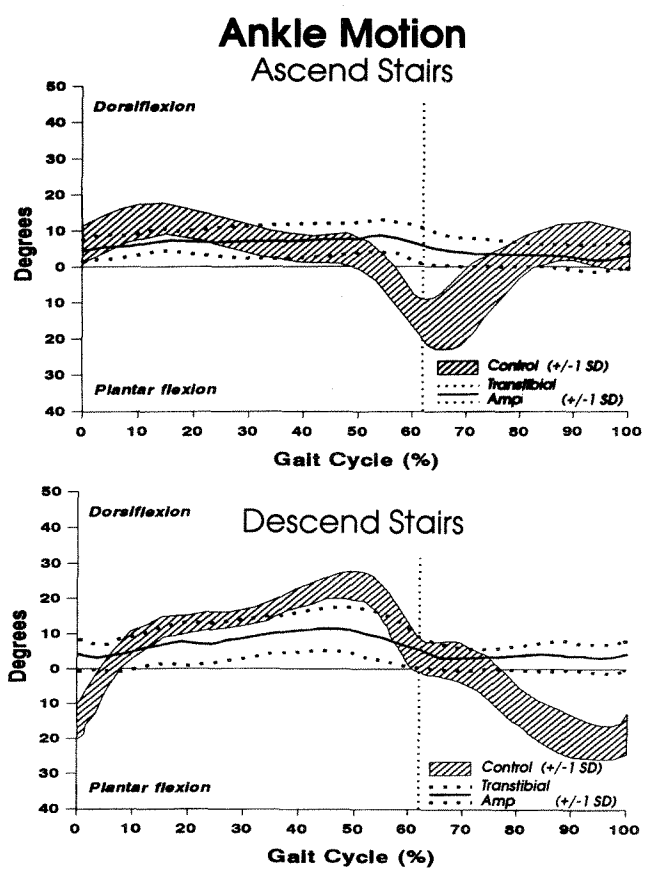

Figure 1.
Mean joint motion (±1 standard deviation) of the ankle for controls (n = 14) and those with transtibial amputation (n = 10) during stair ascent and descent.

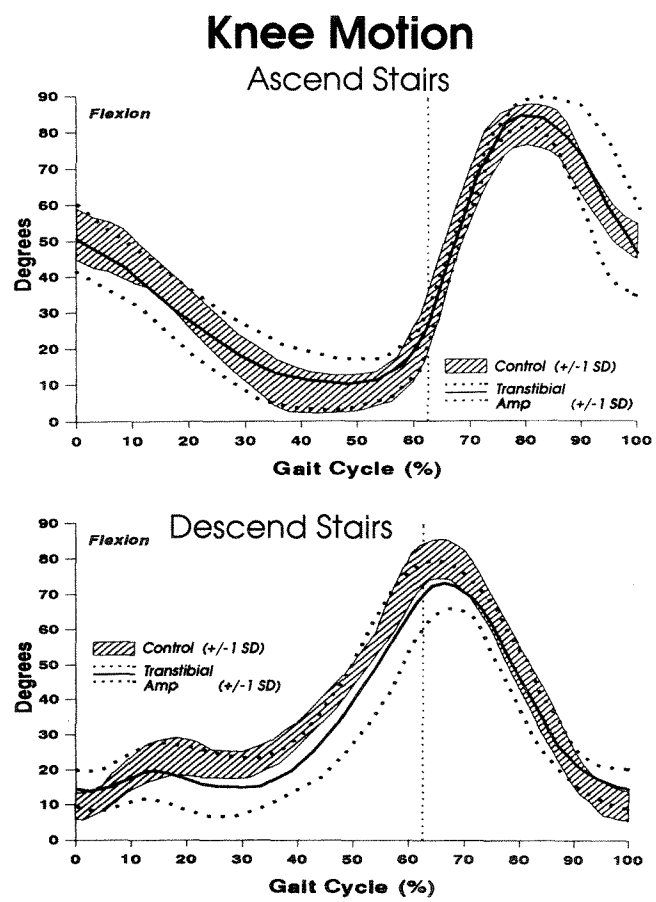

Figure 2.
Mean joint motion (±1 standard deviation) of the knee for controls (n = 14) and those with transtibial amputation (n = 10) during stair ascent and descent.

At the hip, the group with amputation had significantly greater flexion throughout stance, with the largest difference occurring at approximately 50 %GC (22° vs.

9°, $p<0.01$) as shown in **Figure 3**. In addition, the subjects with amputation demonstrated greater anterior

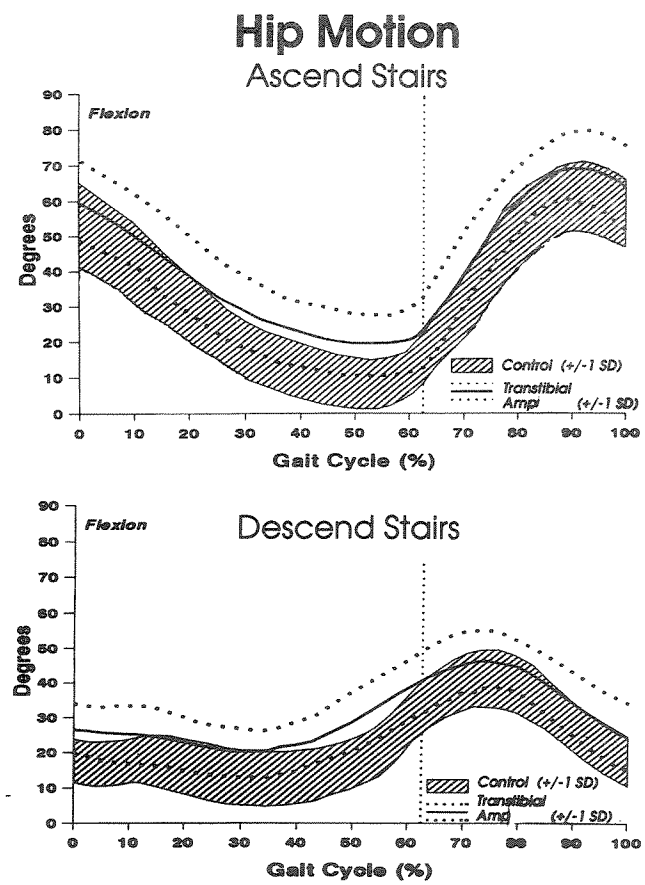

Figure 3.
Mean joint motion (±1 standard deviation) of the hip for controls (n = 14) and for those with transtibial amputations (n = 10) during stair ascent and descent.2

pelvic tilt compared to the controls throughout the gait cycle ($p<0.05$; **Figure 4**).

Ascending Stairs: Muscle Action (EMG)

The group with amputation demonstrated significantly greater EMG intensity for all muscles tested (**Figures 5–10**). When averaged across all muscles, the mean intensity for the group with amputation was 32.4 %MMT compared to 22.8 %MMT for the control subjects. Among the primary hip and knee extensor muscles (SM, BFLH, VL, and GM), the dominant activity occurred during stance for both groups of subjects (**Figures 5–8**).

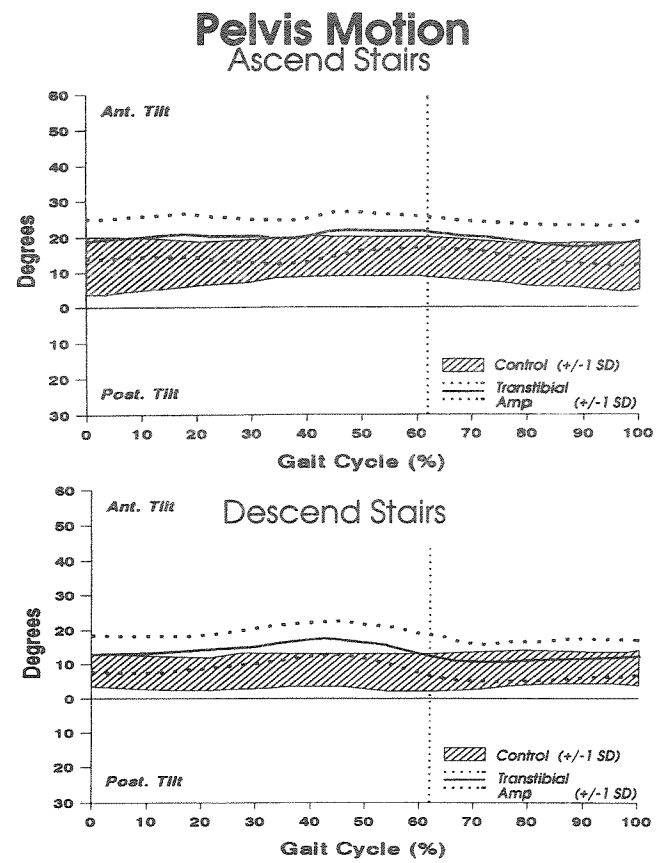

Figure 4.
Mean joint motion (±1 standard deviation) of the pelvis for controls (n = 14) and for those with transtibial amputation during stair ascent and descent.

Timing also was similar, but the mean intensity of the three hip extensors averaged 20 percent greater for the subjects dependent on a prosthetic foot. For the VL, the increased intensity among the subjects with amputation was 40 percent higher than the control effort.

Two muscles, RF and BFSH, showed significant activity in both swing and stance (**Figures 9** and **10**). RF was seen to be biphasic in both groups, however this differed among subjects. For example, only 5 out of 10 subjects with amputation had activity during the late stance-swing portion of the gait cycle, as opposed to 5 out of 7 of the controls. Similarly, 6 out of 10 subjects with amputation demonstrated stance phase activity, compared to only 2 out of 7 controls (**Figure 9**). The BFSH was nearly continuous for the subjects with amputation (38 %GC to 25 %GC), with activity being significantly longer in

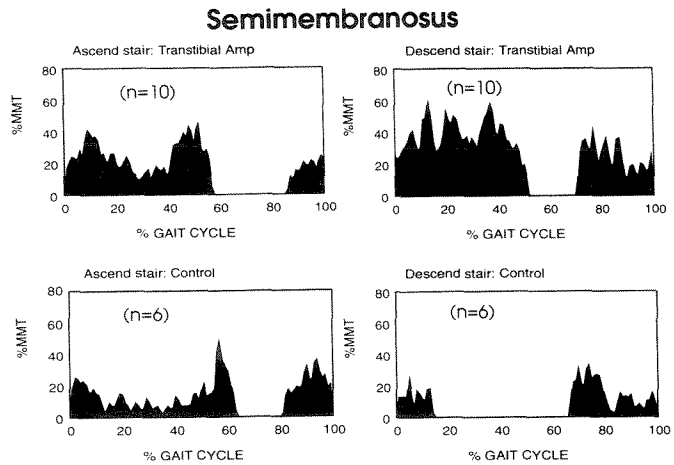

Figure 5.
EMG profile for semimembranosus while ascending and descending stairs (n = number of subjects; %MMT = percent of maximal muscle test). Amp = amputation.

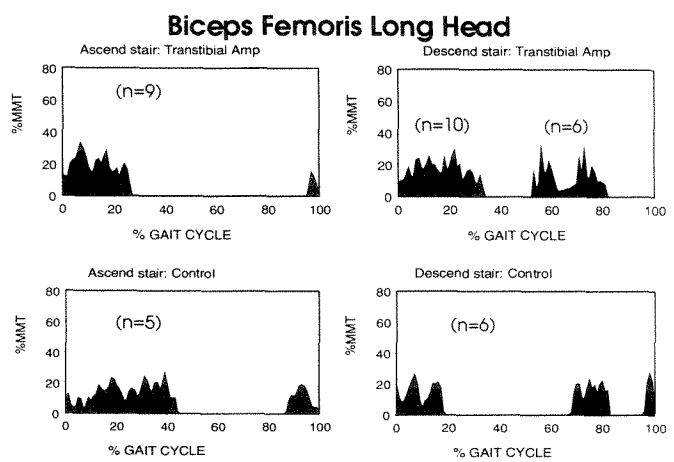

Figure 6.
EMG profile for biceps femoris long head while ascending and descending stairs (n = number of subjects; %MMT = percent of maximal muscle test). Amp = amputation.

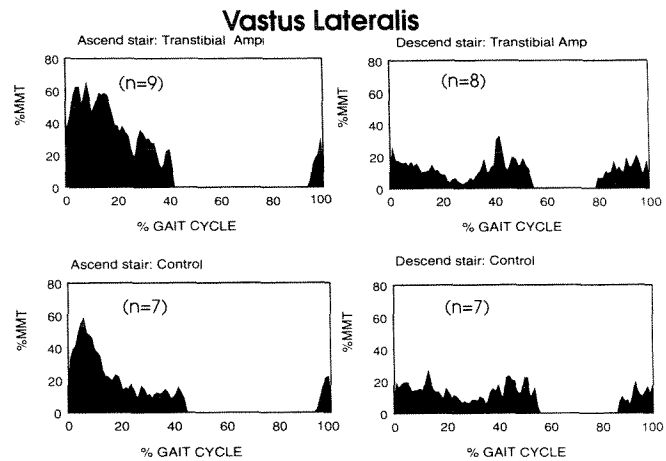

Figure 7.
EMG profile for vastus lateralis while ascending and descending stairs (n = number of subjects; %MMT = percent of maximal muscle test). Amp = amputation.

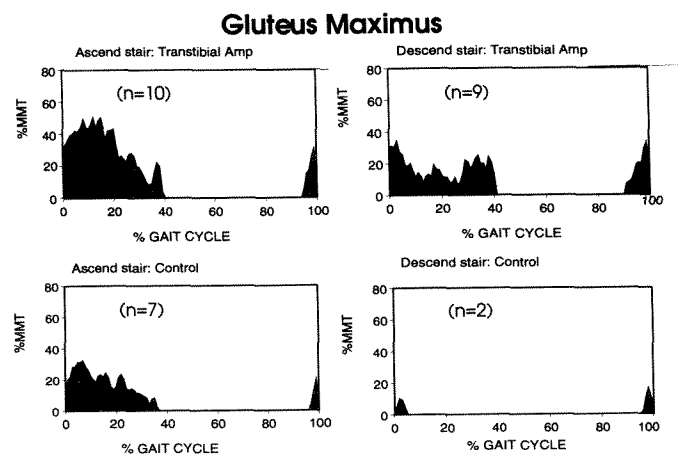

Figure 8.
EMG profile for gluteus maximus while ascending and descending stairs (n = number of subjects; %MMT = percent of maximal muscle test). Amp = amputation.

duration compared to the control phasing (45 %GC to 93 %GC) as shown in **Figure 10.** No other timing differences (onset or cessation) were found for the remaining muscles for ascending stairs.

Descending Stairs: Stride Characteristics

As with ascending stairs, the group with amputation demonstrated a significantly slower velocity during stair descent compared to the controls (29.6 m/min vs. 35.2 m/min; $p<0.05$). In addition, stance asymmetry between the sound and residual limbs of the subjects with amputation was observed. The mean stance time for the sound limb was 65.8 %GC, which was significantly greater than the stance time of the prosthetic limb (58.3 %GC; $p<0.01$). This difference between limbs was reflected in all three of the stance subdivisions; however, only the single limb support time was statistically significant (residual: 30.6 %GC vs. sound: 40.4 %GC; $p<0.01$). The con-

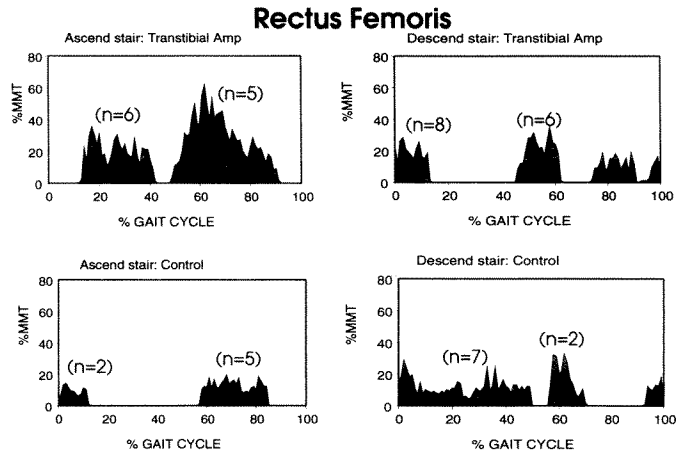

Figure 9.
EMG profile for rectus femoris while ascending and descending stairs (n = number of subjects; %MMT = percent of maximal muscle test). Amp = amputation.

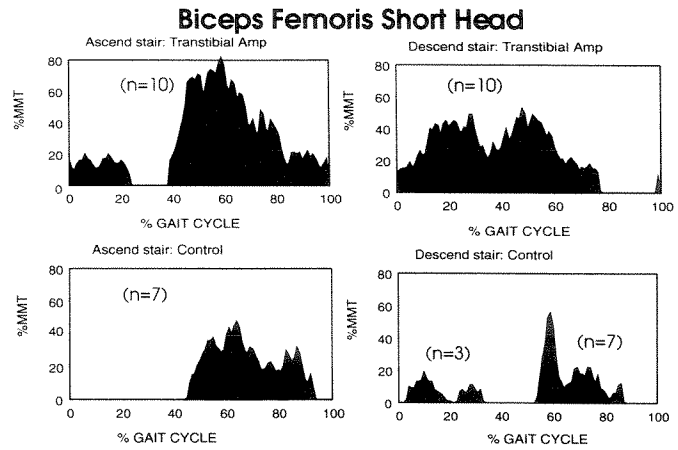

Figure 10.
EMG profile for biceps femoris short head while ascending and descending stairs (n = number of subjects; %MMT = percent of maximal muscle test). Amp = amputation.

trol subjects demonstrated symmetrical stance times between limbs (61.7 %GC), with no significant differences in single or double limb support times (**Table 3**).

Descending Stairs: Motion

During stair descent, ankle motion of the subjects with amputation demonstrated significant stance and swing phase differences compared to controls (**Figure 1**). At initial contact, the prosthetic "ankle" was in 3° of dorsiflexion compared to 14° of plantarflexion as observed in the control subjects ($p<0.001$). During stance, the intact ankle progressed to a maximum of 23° dorsiflexion at 48 %GC, while the prosthesis only yielded to 10° of dorsiflexion at 46 %GC ($p<0.001$). During swing, the prosthetic ankle maintained approximately 4° of dorsiflexion, while the ankle of the control subjects gradually dropped to 19° of plantarflexion by terminal swing ($p<0.001$; **Figure 1**).

The control group had significantly more knee flexion during stance compared to the subjects with amputation, with the largest difference occurring in late stance (25° vs. 17°, $p<0.01$) as shown in **Figure 2**. In contrast, the subjects with amputation had significantly greater stance phase hip flexion, with the greatest difference occurring at 35 %GC (29° vs. 17°; $p<0.01$) as shown in **Figure 3**. Those with TTA also demonstrated greater anterior pelvic tilt compared to the controls throughout the gait cycle ($p<0.01$; **Figure 4**).

Table 3.
Stride characteristics: descending stairs—mean (SD).

	TTA Residual Limb N=10	TTA Sound Limb	p-value*	Control N=14	p-value**
Velocity (m/min)	29.6 (5.4)		—	35.2 (3.6)	<0.05
Stance (%GC)	58.3 (3.0)	65.8 (3.8)	<0.01	61.7 (2.4)	<0.01
SLS (%GC)	34.2 (4.5)	41.7 (3.0)	<0.01	38.3 (2.4)	<0.01
IDLS (%GC)	15.2 (6.1)	10.1 (5.6)	NS	12.2 (1.7)	NS
TDLS (%GC)	10.1 (5.6)	15.2 (6.1)	NS	12.6 (3.6)	NS

TTA = subject with transtibial amputation; *Significance between sound and residual limb of TTA; **Significance between the Control group's reference limb and TTA's residual limb; SLS = single limb support; IDLS = initial double limb support; TDLS = terminal double limb support; %GC = percentage of the gait cycle; NS = nonsignificant at the $p<0.05$ level.

Descending Stairs: Muscle Action (EMG)

The greatest difference between the two groups was seen in the gluteus maximus. Nine out of 10 of those with TTA had significant stance phase activity (average of 23 %MMT), which lasted until 40 %GC (**Figure 8**), while only two of the control subjects had EMG action (97 %GC to 4 %GC). The EMG of during stair descent for those with TTA was much more variable with three muscles (BFLH, RF, and BFSH) having biphasic activity. The subjects with amputation also demonstrated longer stance phase EMG of the BFLH and SMEMB compared to controls (no intensity differences), with EMG continuing until 34 %GC and 52 %GC, respectively (**Figures 5** and **6**). In addition, 6 out of 7 controls and 6 out of 10 subjects with amputation demonstrated swing phase BFLH activity (**Figure 6**).

Similar to the ascend stair condition, the group with amputation had nearly continuous, high-level BFSH activity (average of 35 %MMT) that began at initial contact, lasted throughout stance, and continued to the middle of swing (77 %GC) as shown in **Figure 10**. In contrast, the primary EMG in the control group was of short duration, occurring from 54 %GC to 87 %GC. Only 3 out of 7 controls demonstrated BFSH stance phase activity (**Figure 10**).

Although there was no difference in VL timing or intensity between groups (**Figure 7**), the RF did show a different pattern of activation. Compared to the control group, the subjects with amputation had a significantly earlier onset (76 %GC vs. 94 %GC) and significantly earlier cessation (16 %GC vs. 47 %GC). Five out of 10 subjects with amputation and 4 out of 7 controls also had RF activity in late stance-early swing as shown in **Figure 9**.

DISCUSSION

The results of this study indicate that a prosthetic foot significantly alters stair ascent and descent and presents an increased challenge for the person with a TTA. In general, the difficulty of stair ambulation for those with a TTA was reflected by a significantly slower velocity, an asymmetrical gait pattern, and altered muscular activity compared to the controls. Such findings are indicative of reduced functional ability, and illustrate the limitations of current prosthetic componentry in replicating nondisabled gait patterns.

Stride Characteristics

On average, velocity of subjects with amputation during stair ascent was 80 percent of the controls, while the velocity during stair descent was 84 percent of the controls. These differences imply limited ability to elevate body mass and reflect a diminished rate of forward progression. The observed reduction in gait velocity during stair ambulation is consistent with data previously reported for this population during level walking (11–13). The 17-year age discrepancy between our control and experimental groups may suggest that this is one cause of the significant difference in walking velocities. Other gait data from our laboratory, however, which include 247 nondisabled individuals, indicate that age does not significantly influence adult walking speed until after age 70 (14).

The subjects with TTA also demonstrated significant differences in stride parameters between the sound and residual limbs. The largest deviation was evident in the stance time of the sound limb, which was 14 percent longer than that of the residual limb for ascending stairs, and 13 percent longer than the residual limb for descending stairs. Our finding of stance phase asymmetry between the two limbs was greater than that reported for level walking by Menard et al. (15) and Lewalen et al. (16). These authors observed that subjects with TTA spent only 4 percent more time on the sound limb compared to the amputated limb. This discrepancy between stair and level walking indicates that the increased demand of stair ambulation imposes greater gait asymmetry in this population.

The differences in stance duration, for persons with TTA during both stair ascent and descent were reflected by diminished single limb support times on the involved side. Compared to the controls, 20 percent less time was spent in single limb support during stair descent, and 12 percent less time in single limb support while ascending stairs. According to Eke-Okoro (17), diminished single limb support time is an indication of instability and difficulty in controlling balance, which suggests that persons with TTA are less stable during stair ambulation, and may require postural compensations to assist them during this task.

Individuals with TTA had a lengthened initial double limb support (IDLS) interval on the amputated side for both conditions. As this phase of the gait cycle is considered to be the period of weight acceptance, such a finding indicates a delay in transferring weight onto the prosthetic limb (14). This temporal asymmetry can be explained

by the limited ankle kinematics of the Seattle LightFoot prosthetic device during initial stance.

Motion

Normal loading response motion at the ankle was characterized by 13° of dorsiflexion, while the subjects with amputation demonstrated only 7° of dorsiflexion. Similarly, during stair descent, the control subjects demonstrated approximately 24° of dorsiflexion motion (14° of plantar flexion at initial contact to 10° of dorsiflexion by the end of loading response), while the subjects with amputation remained in 7° of dorsiflexion throughout weight acceptance. Decreased mobility of the prosthesis during loading response limited the ability to advance body weight over the foot and consequently delayed forward progression.

Restricted ankle motion in the persons with TTA also was evident during terminal stance. During stair ascent, the intact ankle displayed two stages of progressive plantarflexion. With the onset of single limb support, the ankle gradually reduced its previously dorsiflexed posture toward a neutral alignment. Then, during terminal double limb support (TDLS), there was an additional arc of plantarflexion as the limb prepared for swing. In contrast, the prosthetic feet of the group with amputation remained in approximately 10° of dorsiflexion. During stair descent, the primary motion deficit for those with TTA was decreased dorsiflexion in single limb support (10° vs. 23°). The restricted mobility evident with the Seattle LightFoot during this phase of the gait cycle is consistent with that reported by Torburn et al. (3) during level walking, and further demonstrates the disadvantage of limited ankle mechanics.

Muscle Action

Compensations for the inability of the prosthetic foot to simulate normal motion were revealed in altered muscular activity. In general, persons with TTA demonstrated more intense and prolonged EMG for all six muscles when compared to the controls. This finding of increased muscular effort is consistent with previous reports of increased energy expenditure that has been documented in this population during gait (3,11).

Andriacchi also noted that during stair ascent both the motion and moment at the ankle are in dorsiflexion and that the soleus balanced this action (1). This allowed the subjects to advance body weight over the supporting foot while the quadriceps activity extended the flexed knee. Then, during the period of double limb support the ankle significantly plantarflexed to preserve balance. The prosthetic foot lacked both the dorsiflexion arc for center of gravity progression and free plantarflexion to assist balance as the other limb began its elevation task.

Increased hip flexion and anterior pelvic tilt facilitated advancement of the body's center of gravity over the noncompliant ankle. Postural stability was preserved by a 20 percent greater effort of the hip extensor muscles and a 40 percent increase in VL muscle activity. Further demand was placed on the hip extensors; as a result, a forward trunk lean, which was employed to augment progression over the stiff prosthetic foot, was evident by the increased hip flexion and anterior pelvic tilt during stance.

The high quadriceps (VL) demand implies that the prosthetic foot did not provide optimal stability. The increased demand at the knee in the group with TTA was further illustrated by the observation of stance phase activity of the RF, which was recorded in 6 of the 10 subjects. This muscle was recruited to assist the vasti in the elevation of body weight. In contrast, only two control subjects demonstrated stance phase activity of RF. The primary RF activity for both groups, however, occurred during pre-swing, serving to flex the hip and assist in advancing the limb.

The increased hip flexion was controlled by the hip extensors (SMEMB, BFLH, GMAX) as shown by the prolonged activity of this musculature into mid and terminal stance. The largest EMG difference between the two groups during stair descent was observed in the GMAX. In the control group, only two subjects demonstrated brief activity of this muscle during terminal swing and loading response. Among those with TTA, however, all demonstrated high-level GMAX activity. Increased recruitment of the GMAX by the person with amputation would be logical, as this muscle would assist in lowering of body weight and controlling excessive hip flexion without inducing further knee flexion by hamstring activity.

Andriacchi (1) found that during stair decent, the greatest demand occurs at the knee (knee flexion moment). This is controlled by eccentric quadriceps activity throughout stance, which serves to lower the body center of mass to the next step. The lack of a significant difference in VL intensity or timing between the two groups during this activity indicates that the prosthetic foot did not alter this demand. However, the high demand of this task for both groups was illustrated by stance phase activity of the RF, which was recruited to assist in the lowering of body weight. Earlier than normal cessation of RF activity in those with TTA is consistent with

the decreased knee flexion and the accompanying increase in hip flexion occurring in late stance. At the knee, the need for extensor muscle action is reduced, while RF activity would add an undesirable demand on the hip extensors (gluteus and hamstrings) as they actively stabilize the hip's flexed posture.

An interesting finding in this study was the nearly continuous activity of BFSH in those with TTA. As demonstrated by the control group, this muscle functioned primarily as a swing phase knee flexor, which was necessary to assist in clearing the foot for limb advancement. In the persons with TTA, however, there also was substantial stance phase BFSH activity, particularly during descending stairs. This increased EMG may have been a protective response to prevent excessive pressure of the distal anterior tibia against the prosthesis. Such action would also counteract the anterior drawer effect of the quadriceps, which were also active during stance.

CONCLUSION

During both stair ascent and descent, the lack of normal ankle dorsiflexion reduced the ability of those persons with TTA to progress over the prosthetic foot. This necessitated compensatory gait strategies, which resulted in a slower walking velocity, asymmetrical gait pattern, and increased muscular activity. The requirements of stair ambulation, as well as other functional tasks, challenge designers of prosthetic feet to provide the mobility needed for optimum progression, while also preserving the stability required for level surface gait. Additionally, stair ambulation should be recognized by rehabilitation professionals as a separate, highly complex task for those with transtibial amputation.

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