FINAL REPORT

The impact of step targeting during normal gait for persons wearing either a SACH or a dynamic-response foot

Presented by Project Principal Investigator:

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INTRODUCTION

This study compared dynamic-response feet to SACH feet with respect to factors influencing unilateral trans-tibial amputee balance and ability to adapt to variable terrains. This was done by measuring ground reaction forces for 30% perturbations of step length during level walking. These perturbations resulted in either a lengthening or a shortening of one step length by 30% of the normal step length. Subjects walked along a 12 metre walkway and across two flush mounter force platforms while forces were recorded for both feet. Three experimental conditions were completed with each foot type: normal step length, short step length (reduced by 30% from normal), and long step length (increased by 30% from normal).

Specific Aims: The specific aims of this project were to investigate the extent to which persons with a unilateral trans-tibial amputation (TTA) modulated vertical and horizontal forces as a means to negotiate changes in terrain and maintain balance and how this was affected by wearing a SACH foot versus wearing a dynamic-response foot. It was hypothesised that:

- Long and short steps will show significantly altered peak forces during the first half of the stance phase for the leading foot and the second half of the stance phase for the trailing foot,
- These differences will be augmented for the SACH foot as compared with the energy storing foot and
- Timing of gait characteristics will be preserved as participants modulate gait to maintain stability.

Background/Significance: There has been much research activity focused on themes relating to human locomotion and how such factors as terrain, aging, and associated changes in kinesthetic sensitivity impact natural gait. Other related research themes include investigations of these issues in different populations, such as those with Parkinsonism, stroke, and persons with lower-limb amputation.

An individual who has had part or all of one or two legs amputated faces many struggles and challenges in the search for return to full mobility. These challenges include, but are not limited to, the level of amputation and the degree of fit of the prosthesis, their overall health and state of mind, cause of amputation, and the quality or degree of sophistication of their prosthetic. Treatment of the individual by the specialist must consider all of these aspects as they provide support for the individual.

Research activities to date have focussed typically on single issues, often in a laboratory environment, partially because of research constraints, funding support, and related factors. For example, there is a considerable body of knowledge regarding the decline of CNS functions associated with aging but the consequences of the additional complication of lower-limb amputation have not been adequately covered (e.g. McFayden et al, 2002). Similarly, much research has been conducted on slips and falls and yet the main focus of the research is on persons with both intact legs (e.g. Feldman et al, 2007, Liu-Ambrose et al, 2008). The incidence of falls and fear of falling are common in individuals with lower-limb amputations. Environmental barriers can increase the frequency of falling (Uelger et al, 2010). Walking on uneven surfaces, in particular, may be difficult for persons with lower-limb amputations due to limited ankle-joint mobility on the prosthetic side, and lack of distal musculature and sensory feedback from the lower limb. Individuals with and without lower-limb amputation differ in the way they respond to challenging surfaces like stairs, ramps, or uneven ground. Additionally, an individual with a lower-limb amputation may make different adaptations on their prosthetic and intact limbs. Another factor that may affect persons with lower-limb amputations is the degree of sophistication of their prosthesis. Balance, in particular, is potentially impacted by the mechanical properties of a prosthetic foot – how stiff or flexible it is and in what planes it is flexible/stiff. Design may have an effect on how easily a person can recover from a balance perturbation. Confounding these factors is the impact aging or progression of a condition such as Parkinsonism can have on the central and peripheral systems, systems that determine balance control, recovery from perturbations and general mobility.

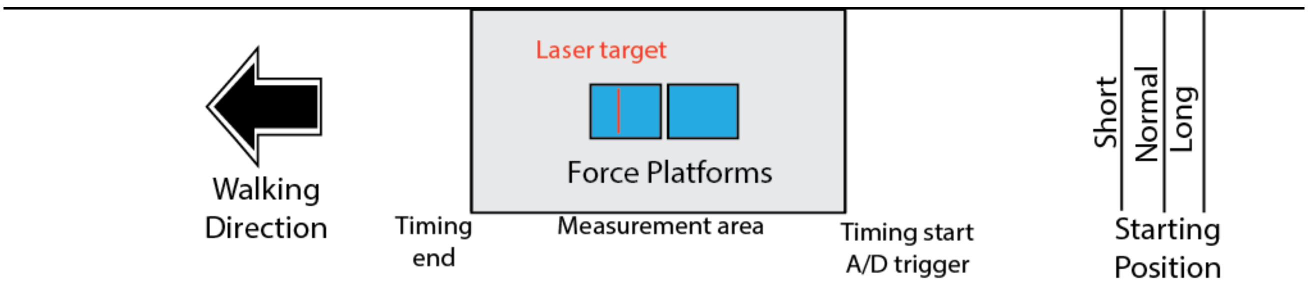
These are just some of the many factors worth exploring when considering how persons with amputations deal with their environment. This is an area that deserves research attention because of the anticipated rise in the number of persons needing an amputation of a lower limb. Diabetic and traumatic amputations are increasing the number of individuals with limb loss. Dysvascular amputee care alone costs more than $4.3B in the US. This proposal aims to conduct a pilot study to explore how people with TTA respond to sudden changes in the walking environment, and how choice of prosthetic componentry may affect their responses. In every-day life, people walk over a variety of terrains with different surface characteristics such as cement or grass, slippery surfaces like puddles or ice, and uneven ground that include steps and curbs. Variations in terrain can lead to falls if individuals do not or cannot adapt their gait appropriately. This is acutely true for persons who have had one or both legs amputated or who have decreased sensitivity due to peripheral neuropathy.

In this project, participants will be required to target a landing zone during normal walking. By adjusting the starting positions we can manipulate whether participants can land on the target without altering their stride or we can force them to either shorten or lengthen the final step to achieve the target. This has relevance because, for example, persons crossing a road need to anticipate whether they can easily negotiate the curb. Or perhaps an individual comes to a point in the sidewalk where it is broken and they need to make a rapid adjustment to foot placement to continue walking. Central to this is the effect prosthetic componentry has on the amputees' ability to make these adjustments. To investigate this, we will include a condition that explores the extent to which participants can be more or less successful at the task depending on which type of foot they use. We anticipate differences in reaction forces as the SACH foot provides less flexibility than a dynamic response foot to terrain variations. We believe these differences in flexibility will translate into force variations with step-length perturbations.

Methods

Participants: A group of five individuals volunteered for participation in this pilot study. Inclusion criteria for the participants included males over the age of 19, without neurological conditions which influence balance, having a unilateral trans-tibial amputation and having an exoskeletal type prosthetic limb (to aid in foot changeover). The personal foot type for these individuals was a dynamic-response (energy storing) foot type.

Experimental setup: The experimental protocol required that participants, from different starting positions, walk through a measurement area over two adjacent flush-mounted force platforms (Figure 1a and 1b). Data collection was triggered by breaking a photo beam at the onset of the measurement area. A second photo beam provided the time required to walk through the measurement area.



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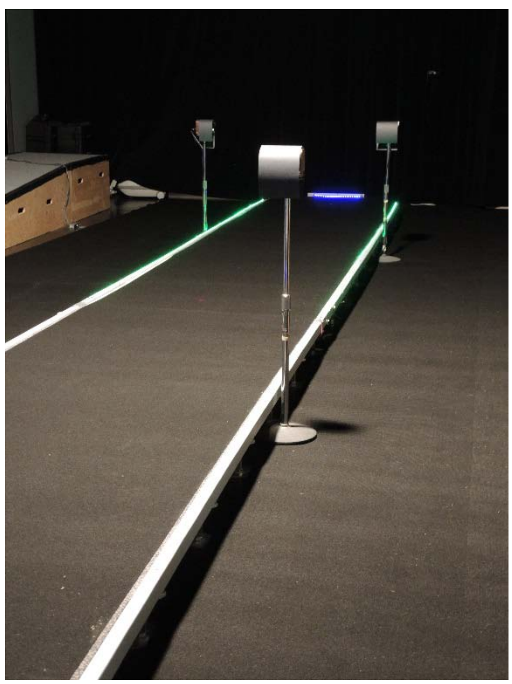

Figure 1. a. Schematic of recording layout and b. photograph of walkway

To facilitate targeting, heel strike on the first platform triggered a red laser beam that projected a red line onto the second platform. To vary the step length onto the second platform, the starting position for each trial was varied. The starting positions were also marked by one of three red laser beams projected onto the walking surface under control of the research team. The selection of the starting position was determined randomly and was not known to the subject. For this reason, lights in the motion capture laboratory were dimmed to a level that the participants were unable to see the platforms. To ensure that the participants walk along the desired path a ribbon of green led lights was placed along each side of the walkway.

The data collection system consisted of two force platforms (AMTI model OR-6, Bertec model 4060) integrated with a Vicon motion capture system (Nexus). Sample rate was set at 960 Hz. On either side of the platforms was placed a pair of photo cells. Breaking the first beam initiated a timer and the signal was used to trigger the A/D on the Nexus system. The signal from the vertical force channel on the first platform was used to trigger the target laser beam aimed at the centre of the second platform. Breaking the second photo-cell beam stopped the timer and this was used to verify that the average walking speed through the measurement area was within the +/- 7% range.

Protocol: Participants visited the motion capture laboratory on a single occasion. The session consisted of a number of steps, including explanation of protocol and completing informed consent, recording anthropometric details, determination of self-selected walking speed and step length, orientation to experimental setup and practice walking in a dimmed room, recording of calibration data, recording of experimental data on own foot, exchange with SACH foot and recording experimental data on SACH foot.

Anthropometric data: These data included height, weight, age, years since amputation, discussion of activity level (K), and identification of personal foot type.

Determination of self-selected walking speed: In a long corridor a 10-metre section was identified. Participants walked through this section a minimum of ten times while time to cover the ten metres and the number of steps taken were recorded. These data were averaged and the average step length and average speed were computed. From these data the 30% perturbation of step length was determined in addition to the self-selected walking speed.

Orientation: After completion of the walking speed determination the participants entered the motion capture laboratory and were given sufficient time to look around and acquaint themselves with the environment. With the lights on participants were provided with the opportunity to walk along the measurement area. When comfortable with that we dimmed the lights, in stages, so that they could accommodate the darkened room comfortably. This was also an opportunity for the subjects to walk at their self-selected speed while we measured the time through the recording area to ensure they could maintain that speed. This period was also used to determine the optimal starting location that ensured both feet landing on the appropriate force platform. This practice period was continued until the participant lands on the platforms with each foot, without any apparent adjustments to his gait style.

Calibration data collection: To ensure that the level of lighting did not affect the ground-reaction force and walking speed data, a series trials were recorded, five with the lights on and five in the dimmed-light environment. Participants had accommodated the dimmed-light environment before starting these five trials. Walking speed was controlled and the ground-reaction force data from both platforms were collected.

Protocol for data collection: During the orientation session participants practiced walking over the force platforms at their self-selected pace and the normal start position was determined. The short and long start positions were computed as 30% of the normal step length determined during hall walking session. Starting on the long or short positions resulted in the participant either over-shooting or under-shooting the target on the second platform. In these conditions an adjustment to their gait was necessary to ensure that they landed on the platform.

There were two categories of foot condition, one wearing their own dynamic response foot and one wearing the provided SACH type of foot. For each foot, the walking trials were divided into two groups. The **lead group** included trials where the prosthetic leg landed on the second platform while the intact leg landed on the first platform. The **trail group** included trials where the intact leg landed on the second platform while the prosthetic leg landed on the first. There were 15 trials in each of these groups. The 15 trial block consisted of five trials with no step-length perturbation, five trials with a lengthened step length and five trials with a shortened step length. The data-collection matrix is shown in the table below.

Own- Dynamic						SACH					
Lead			Trail			Lead			Trail		
Long	None	Short	Long	None	Short	Long	None	Short	Long	None	Short
5	5	5	5	5	5	5	5	5	5	5	5

Table 1. Schematic of recording layout and number of trials for each condition.

Each data-collection session proceeded with one experimenter walking with the subject to provide increased confidence while the other investigator operated the data-collection system. With the subject not looking, the appropriate start beam was selected. The participant positioned himself at this line and when ready walked the length of the measurement area. While walking, the next start beam was selected ensuring the participant remain unaware of the change. If the speed was within range and the feet landed cleanly on each platform the file was saved and the next trial was initiated. The order of all trials was randomly determined.

Data were always collected on the participant's own foot in the first set of trials. Once the first set of data had been collected a prosthetist swapped out the dynamic foot for a SACH. The prosthetist ensured that the fit and alignment was optimal and the participants were given enough time to become comfortable walking with this foot. When the participant felt comfortable the second set of data were collected. The total time commitment was about 2.5 hours.

Analysis: Using the procedures of Sanderson et al (1993), variables such as peak and time-to-peak force, the time of crossover between braking and propulsion, horizontal and vertical impulses were for each trial. The raw data from the force platform was time normalized to 100% of stance duration and amplitude normalized to newtons of force per kilogram of body mass. The five trials for each condition were averaged for each participant and then a group average for each condition was computed. From the two components of the ground reaction forces a number of variables were identified based on their use in conventional gait analysis as indicators of the gait style. A comparative analysis of the stability of each person as it reflects the different properties of each foot type was determined. It should be noted a sample size of n=5 has been chosen based on budgetary constraints rather than statistical sample size calculations. As such, rather than looking for statistically significant results in this study the analysis examined trends to support further research in this area.

RESULTS

1. **Schematic representation of file naming protocol.** The two force platforms were oriented as shown in Figure 2. The second platform, the Bertec, was always the target platform. The target shown as a red line in the centre of the platform. Lead condition refers to the specified leg landing on the target platform. Trail refers to the specified leg landing on the AMTI platform. When the intact leg was the lead leg it landed on the Bertec while the prosthetic leg pushed off from the AMTI. The opposite was the case when the prosthetic leg was the lead leg.

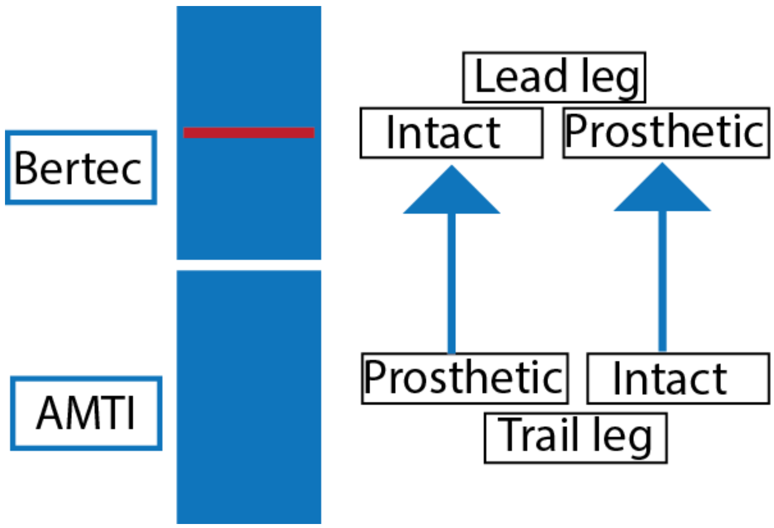

Figure 2. Schematic of file naming convention

2. **Anthropometry.** Basic data were recorded from each participant including, mass, age, time since amputation, level of activity, and details of their prosthetic leg and foot.

Subject	Age (Years)	Mass (kg)	Years since surgery	K level	Personal Prosthetic type
1	76	77.3	41	3	Dynamic
2	51	72.6	32	4	Dynamic
3	41	109.7	31	3	Dynamic
4	48	59.1	46	4	Dynamic
5	57	72.2	4	3	Dynamic

Table 2. Anthropometric data for each participant.

3. **Step length/speed.** After the practice trials, the step length for each foot was determined by having the participants walk along a 10-metre section of carpeted hallway. The mean values for walking speed and step length was determined from 10 trials.

Participant ID	Walking speed(m/s)	Step length (m)
1	1.07	0.63
2	1.90	0.96
3	1.32	0.77
4	1.57	0.81
5	1.03	0.65

Table 3. Mean self-selected walking speed

4. **Walking speed for lead/trail.** Average walking speed for all trials in the leading and trailing condition were determined. There was no significant difference among any walking conditions, which indicated we were successful at controlling walking speed. Leading referred to the prosthetic leg targeting the second platform whereas trailing referred to the prosthetic leg on the first platform.

Participant ID	Walking speed when leading (m/s)	Walking speed when trailing (m/s)	Walking speed when leading (m/s)	Walking speed when trailing (m/s)
	Foot - own		**Foot - SACH**	
1	1.08	1.11	1.10	1.11
2	1.79	1.80	1.83	1.79
3	1.33	1.32	1.33	1.36
4	1.62	1.61	1.58	1.62
5	1.05	1.12	1.14	1.06
Mean	**1.37**	**1.39**	**1.40**	**1.39**
SD	0.33	0.31	0.31	0.32

Table 4. Mean and standard deviation (SD) walking speed for each participant for leading and trailing conditions.

5. **Assessment of effect of dimmed lights on gait parameters.** Because the walking conditions were to be completed in a room with dimmed lights it was necessary to determine whether these dimmed-light conditions would affect the walking characteristics. While the pilot experimentation had indicated there would be no effect we did collect some trial data to verify this assumption for the amputee population group. The table below presents key gait indicators. The modest differences in these variables supported the assumption that dimmed lights had little or no effect on the gait characteristics. Indeed, self-reports from the participants indicated their comfort with the experimental environment.

	Lights dimmed	Lights on
Mean of Trail Crossover (%)	54.6	54.5
SD of Trail Crossover (%)	7.5	5.9
Mean of Lead Crossover (%)	61.7	58.2
SD of Lead Crossover (%)	15.8	9.4
Mean of Trail Max Fy (N/n)	0.19	0.20
SD of Trail Max Fy (N/n)	0.07	0.06
Mean of Trail Min Fy (N/n)	-0.19	-0.18
SD of Trail Min Fy (N/n)	0.07	0.06

Table 5. Mean and standard deviation (SD) crossover, maximum and minimum forces during calibration trials with lights on and dimmed.

6. **Force-time curves.** The figures below present the three components of the ground-reaction forces for the two legs in each light condition. These plots show there were only minor differences in the normalized ground-reaction force patterns. As shown in Table 3 there was no change in walking speed. Subjects reported that they were comfortable walking in the dimmed light situation. We concluded that it would be valid to use the dim-light condition as part of the experimental protocol to mask the foot target.

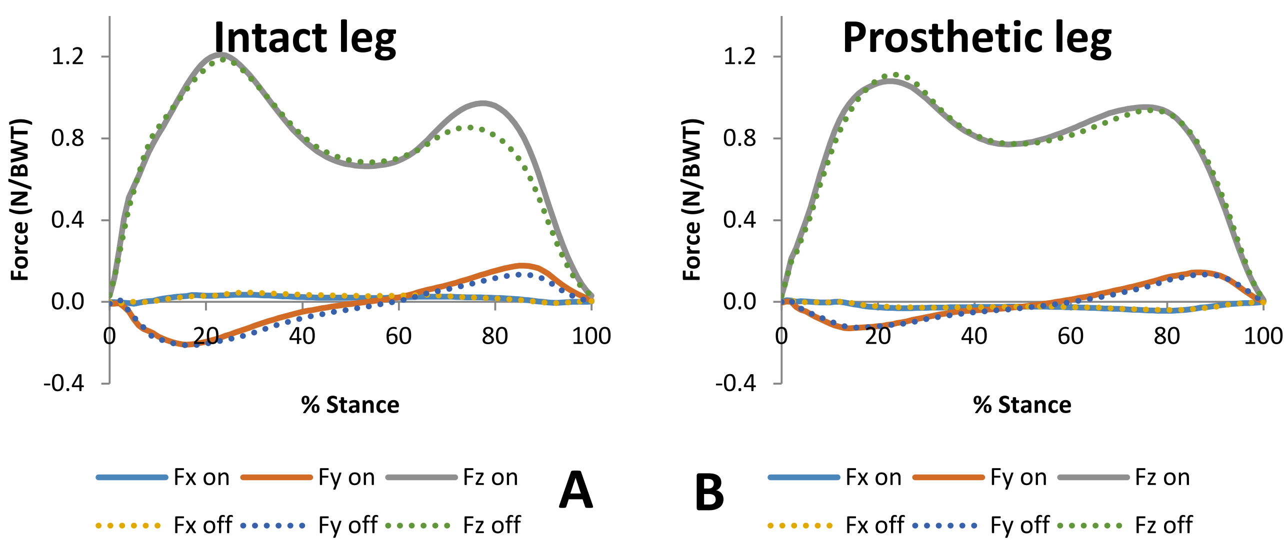

Figure 3. Mean vertical and anterior-posterior force components for the dimmed-light condition and lighted conditions for the intact leg, A, and the prosthetic leg, B.

7. **Cross-over.** The point at which the whole-body centre of mass is over the base of support is a useful indicator of the transition from braking and propulsion. In these data, there was essentially no effect on the % stance at which cross-over occurs in either step-length condition or type of prosthetic foot. This was not surprising given that speed was constrained. In frame D there was an interaction between the type of prosthetic and the short-step condition that might warrant further investigation.

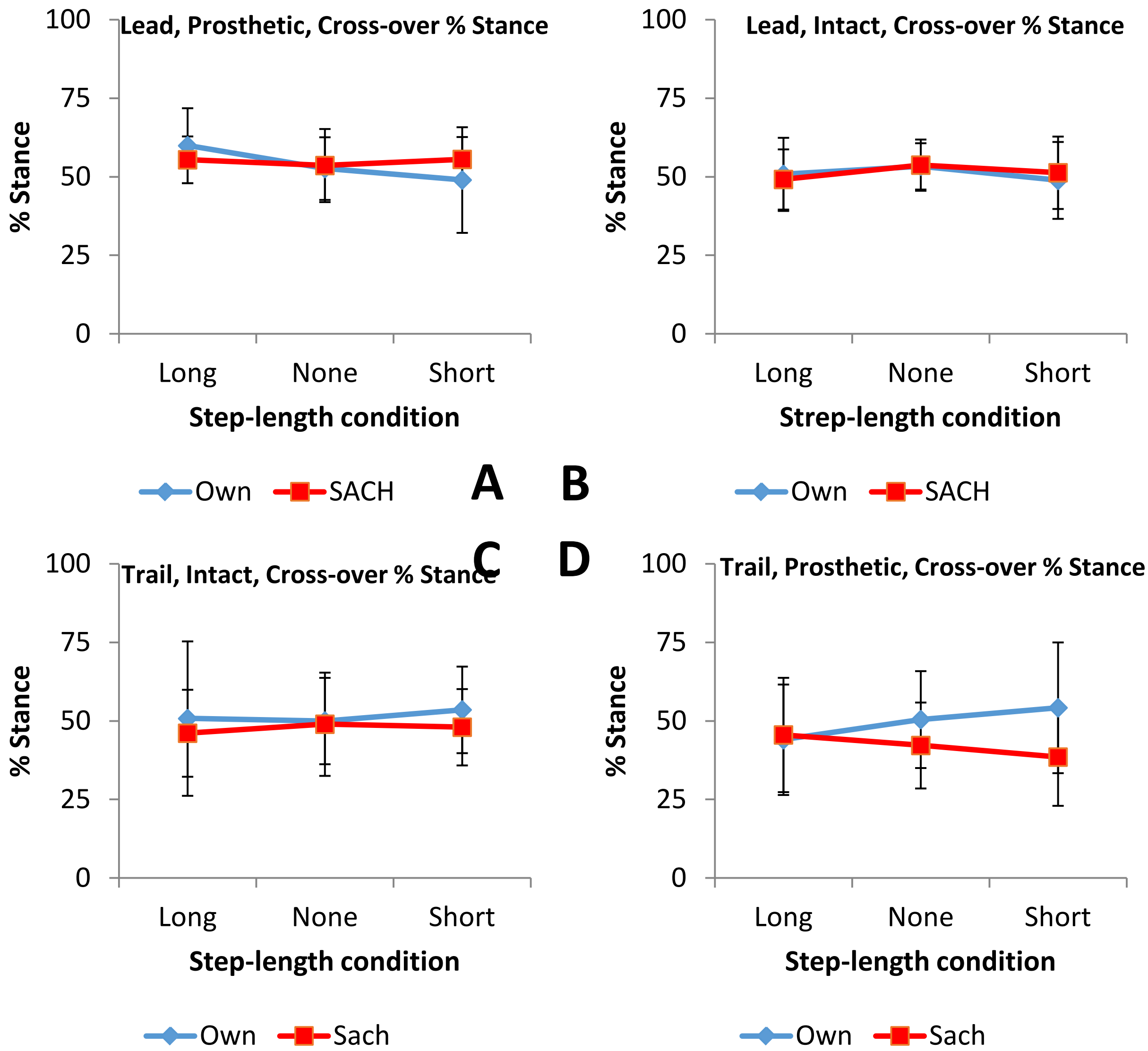

Figure 4. Means and standard deviations for the cross-over point during stance phase. The cross-over point is the point during stance phase when the whole body centre of mass was over the base of support.

8. **Maximum and Minimum anterior-posterior forces.** One of the hypotheses involved the maximum forces during propulsion and braking. Previously published data indicated that the push-off peak force (positive force) would be attenuated at the short step length and accentuated at the long step length. The opposite would be true for the braking peak force (negative force). There is evidence of consistency in frames A and B however, the effect is very weak. This raises the questions of whether the lack of effect is a consequence of a slower walking speed, the small n or increased complexity of amputee gait.

	Long		None		Short	
Condition	Own	SACH	Own	SACH	Own	SACH
Lead (target) leg						
Maximum Fy (N/n)	0.154	0.185	0.173	0.169	0.171	0.154
% Stance to Maximum Fy	82%	82%	78%	81%	77%	84%
Minimum Fy (N/n)	-0.231	-0.219	-0.180	-0.197	-0.136	0.189
% Stance to minimum Fy	13%	13%	19%	18%	21%	16%
Trail leg						
Maximum Fy (N/n)	0.151	0.135	0.153	0.153	0.106	0.160
% Stance to Maximum Fy	80%	76%	84%	77%	68%	76%
Minimum Fy (N/n)	-0.111	-0.146	-0.137	-0.122	-0.140	0.133
% Stance to minimum Fy	17%	14%	18%	16%	20%	29%

Table 6. Mean peak and time-to-peak anterior-posterior force all conditions.

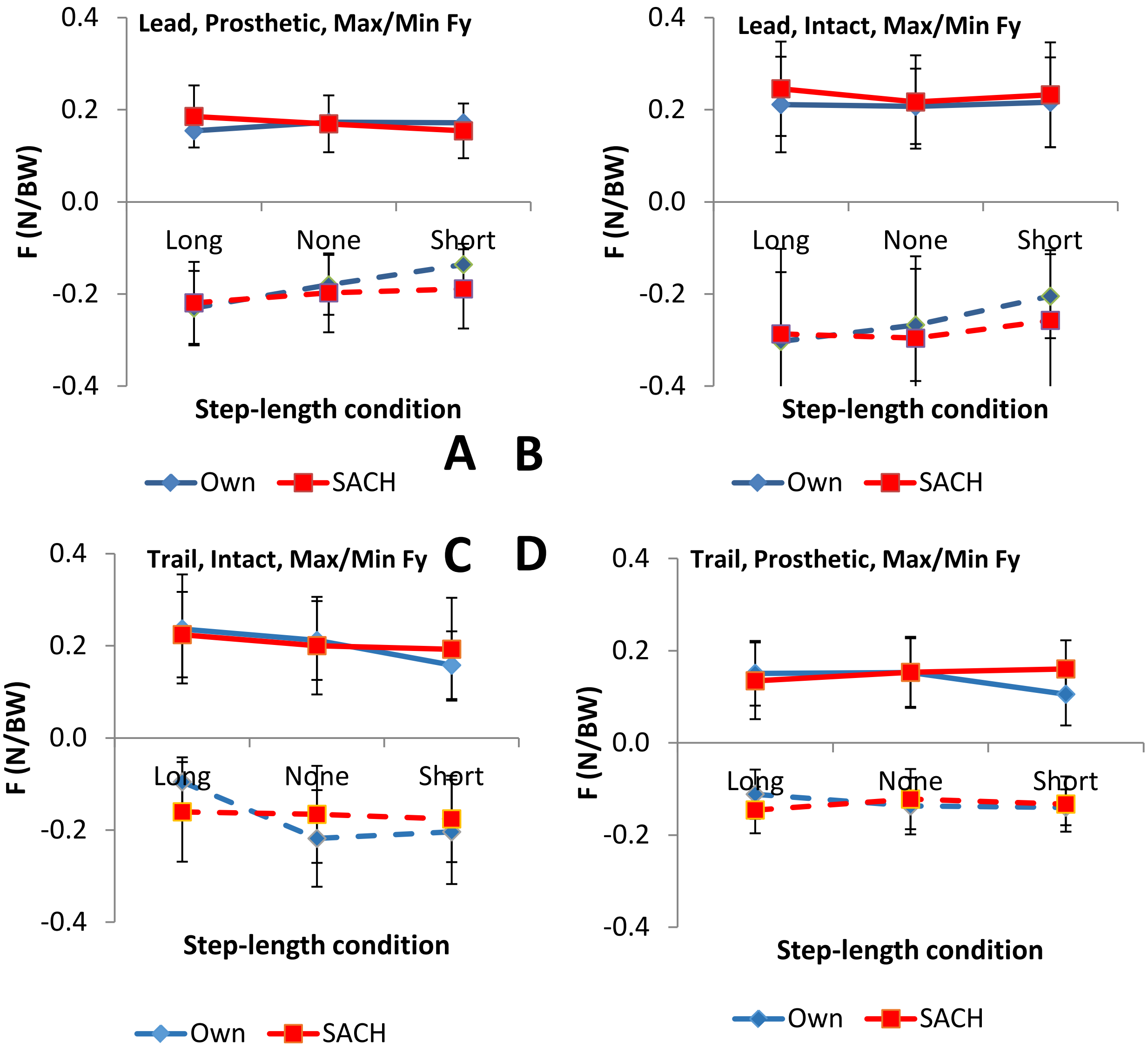

Figure 5. The mean maximum and minimum values for the anterior-posterior force component for each variable.

8. **Force-time curves.** On the following pages are the force-time curves for the anterior-posterior (Fy) ground forces. Examining these plots provides further clues as the repsonses of the participants to the step-length and foot-type manipulations. It was hypothesized that for the long-step condition the propulsion force at push-off (trailing leg) would be increased and consequently the braking force on the target foot would be increased. The opposite would be true for the short-step length condition.

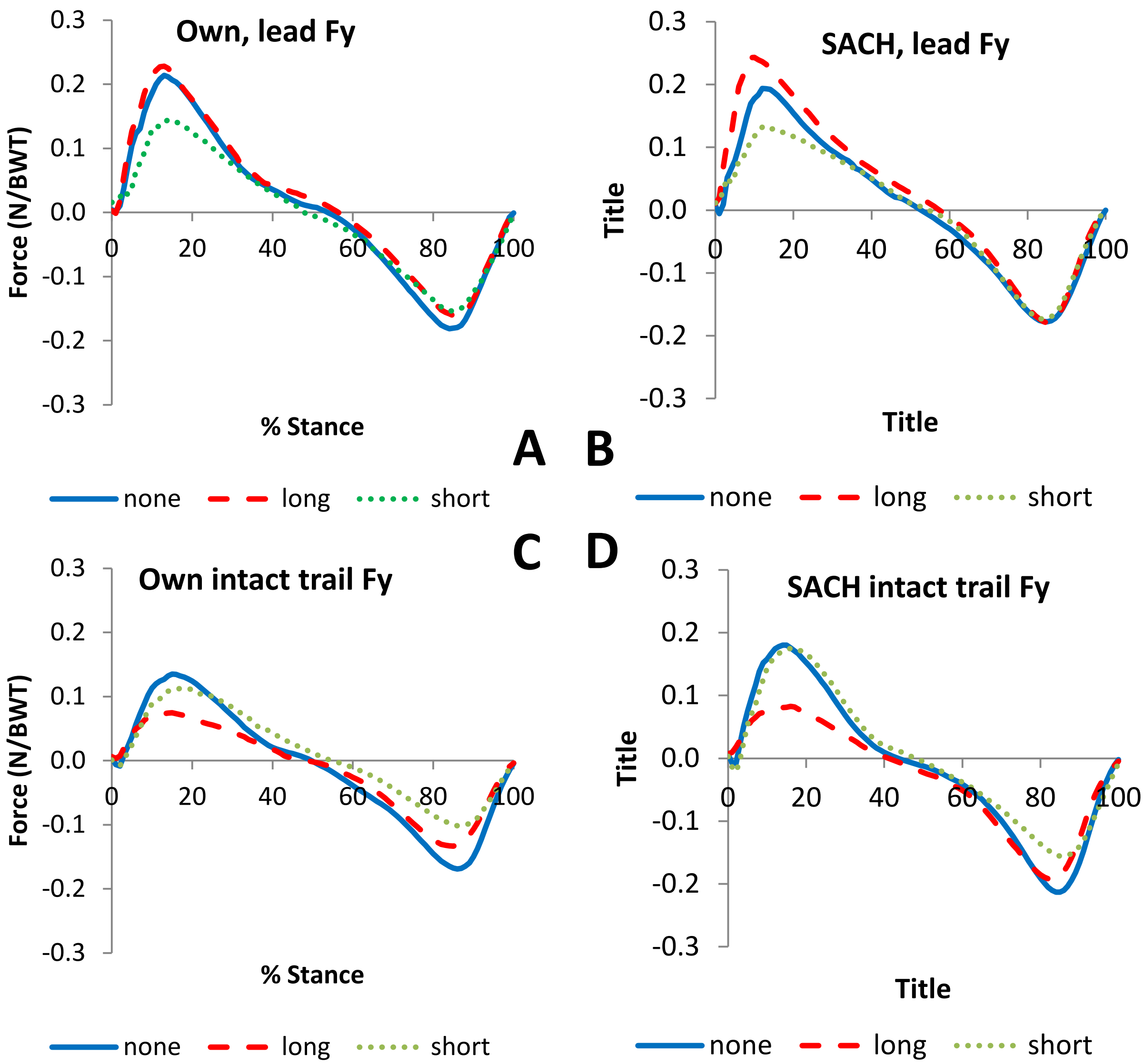

Figure 6. The data in the plots below are partially consistent with the hypothesis. The landing peaks are accentuated in the long step-length condition and reduced during the short step-length condition. As expected, the second half of the propulsion phase shows the forces essentially identical. This is a consequence of the participants maintaining the constant speed and that the next step was not perturbed.

It is interesting to note that the initial phase of support for the trailing leg shows some variations with step-length. It would seem that the participants were anticipating a perturbation and this had an impact of these force-time curves.

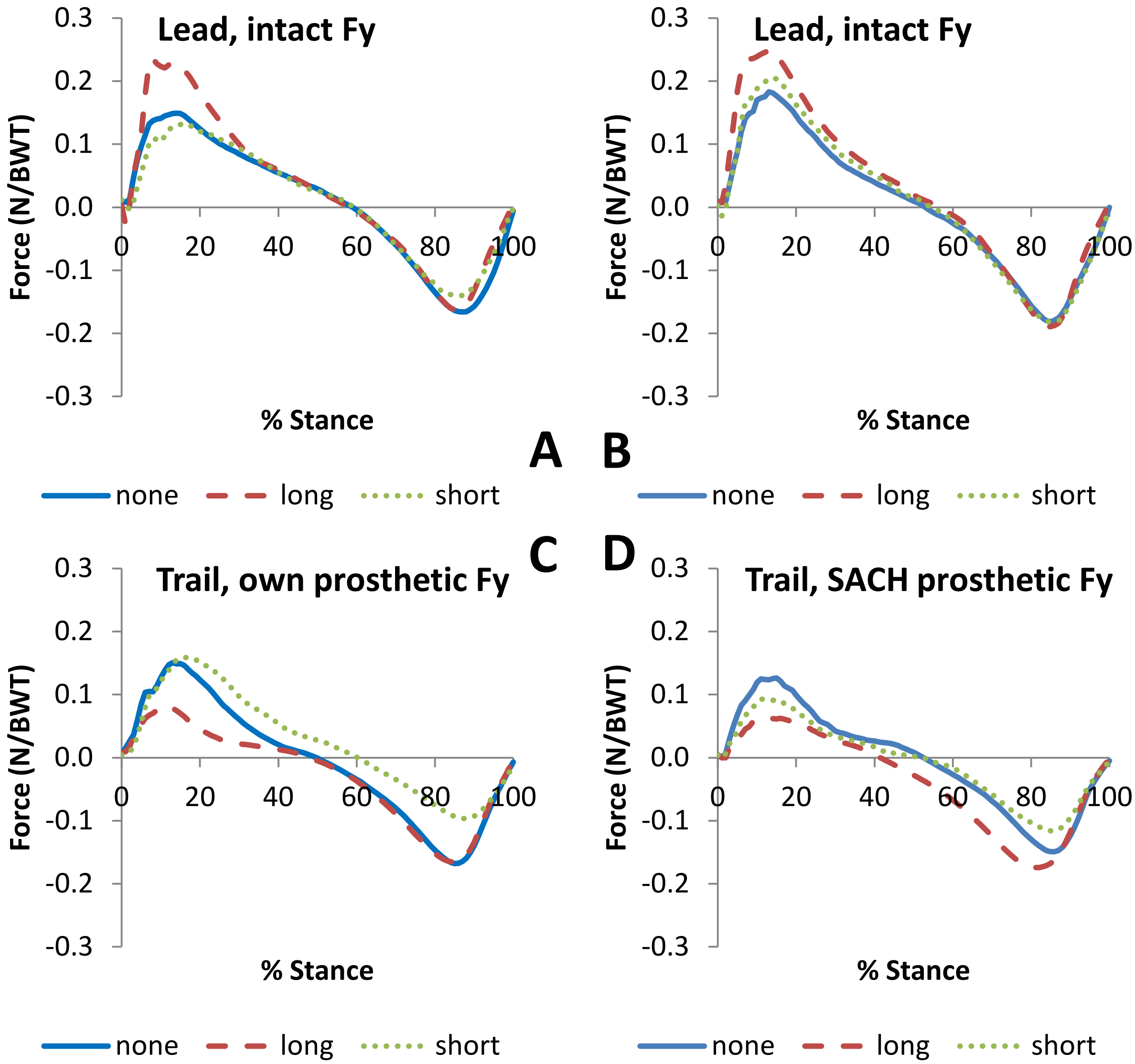

Figure 7. The plots on this page show the conditions where then target foot was intact leg and trailing leg was the prosthetic leg. Propulsion peak on the prosthetic leg is low, as hypothesized, but the corresponding braking peak on the target leg was not. Note that the target leg in the long-step condiiton was large but not the propulsion peak from the trailing leg.

9. **Comparison with publically available data**. These plots compare the A/P and vertical GRF from the ISB website (http:// http://isbweb.org/), normative data from David Winter, for a sample of our data, specifically the intact leg, trailing when the participant was wearing his own prosthesis. The patterns are similar in relative timing and peaks, although not identical. This confirms that our data were consistent with other published data. Kinematic data were downloaded from a public source and using the trunk marker I computed the average velocity over the gait cycle as 1.4 m/s. Table 3 shows that the walking speed of our amputees varied between 1.03 m/s and 1.9 m/s. This explains why our data are not exactly like the control data from Winter (and, of course, our individuals were persons with a trans-tibial amputation and Winter's were individuals with intact legs).

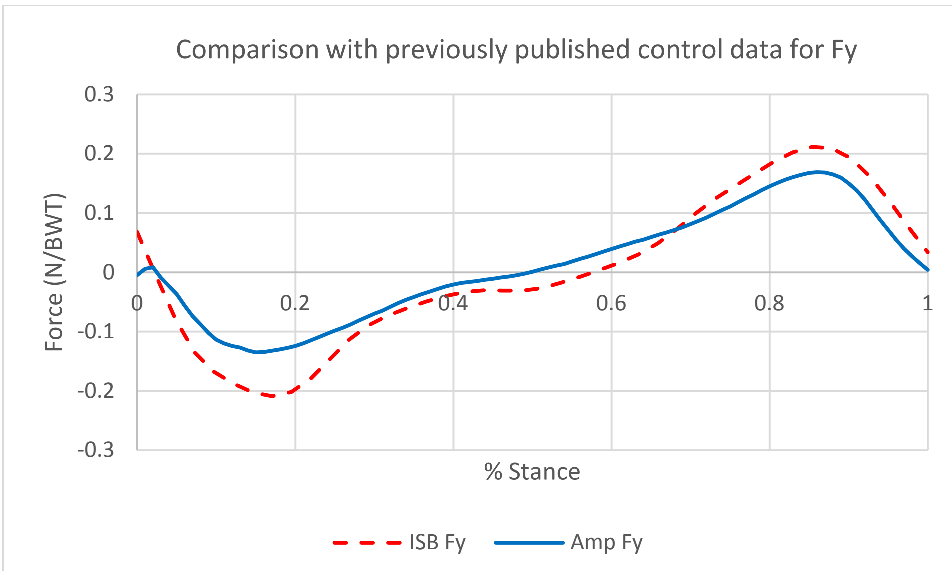

Figure 8. Comparison of mean intact leg anterior-posterior force and publically available data from a non-amputee group walking at a similar speed.

DISCUSSION

The specific aims of this project were to investigate the extent to which persons with a unilateral trans-tibial amputation (TTA) modulated vertical and horizontal forces as a means to negotiate changes in terrain and maintain balance and how this was affected by wearing a SACH foot versus wearing a dynamic-response foot. The results gathered in this study provided some clues regarding this response but clear conclusions were difficult because of the small sample size confounded by the complexity of the gait patterns of persons with a unilateral trans-tibial amputation. Nonetheless, there were a number of indicators regarding how these individuals accommodated the challenges.

The persons who volunteered for participation in the study were representative of an active group. All were at a level of K3 or higher, frequently walked and exercised and consequently had no difficulty with the walking task presented. Little time was required for them to accommodate to the dim lighting in the motion capture laboratory. The level of lighting was sufficiently dim to ensure that the participants could not identify the location of either force platform but the room was not completely without some ambient light, such as that shown on the computer monitors. Anecdotally, the participants commented that the perturbations came as a surprise and they felt that they had to make significant gait adjustments.

The degree of lighting during the data recording sessions was a potential confounding factor. Using the laser lights to indicate starting and target positions and to mask the location of the force platforms mandated that the lighting in the laboratory be dimmed. To assess the impact or potential impact of walking in a dimly-lit room a series of five control trials with lights on and five control trials with dimmed lights were conducted. In each case the participants were requested to walk along the walkway at their self-selected speed. Table 5 presents some key results. For both the lead and trail legs it was clear that there was essentially no effect of the dimmed lights on the gait variables.

There are differences in the gait patterns recorded from the non-amputee person compared to persons with either a trans-tibial and trans-femoral amputation. Forward velocity of walking was significantly lower in the amputee and is lower in the AK than in the BK subjects (Skinner et al, 1985). Traumatic trans-tibial amputees ambulate with time-distance parameters of velocity, cadence, stride length and gait cycle which are all two standard deviations below normal. The same parameters for the traumatic trans-femoral amputee are only one standard deviation below normal. The symmetry of walking seen in the normal subject is not present in the lower extremity amputee (Skinner et al, 1985). The mean walking speed of the current group was 1.39 m/s, a higher average than seen in other publications. This higher average was consistent with this being an active and mobile group.

Data presented in tables 3 and 4 illustrate consistent walking speed for all conditions. This shows that we were able to control walking speed thus eliminating it as a contributing factor to any target related changes. Clearly from the data it was evident that there was no significant difference among the conditions.

In Figure 3 are presented the force-time plots for each leg in the different lit conditions. Visually, there appears to be little difference in the plots with the exception of the intact leg in the dimmed-light condition. It was not clear where these differences arose. Examination of the individual plots did show some difference between and among the subject responses. At this point, having such a small group did not facilitate further exploration. We concluded that the lights had little or no effect on the overall gait variables. However, the differences in Figure 3, panel A, suggest further investigation is warranted.

The hypothesis for the current study was that the anterior-posterior landing forces on the target or lead leg would be greater than normal with the long step and smaller than normal with the short step. Further, a similar relationship would hold for the second half of the stance phase for the trailing leg. That is, there would be an interplay between the landing force on the target leg and the propulsion forces on the trailing leg. We had also hypothesized there would be no change in the timing of these events. A corollary of this interplay would be that the landing force on the trailing leg and the propulsion force on the target or lead leg would be insensitive to the step-length changes. The data presented in Table 6 and in Figure 5 summarize these key variables for each condition.

It was clear from the timing data that there was little or no effect of targeting on the relative timing of the peak forces, either in the braking or the propulsion phases. These data were consistent with the observations that walking speed was tightly controlled and consistent across all conditions. This was also consistent with the data reported by Sanderson et al (1993) regarding the overwhelming power of timing and walking speed. It is likely that in a non-laboratory situation an individual might chose to slow their gait when targeting, for example when stepping up on a curb.

Data presented in Figure 5 regarding the peak forces indicate a mixed and non-consistent response to the experimental perturbation. We predicted that the propulsion peak force, positive force, on the lead leg would be insensitive to the perturbation because it is in preparation for the step out of the measurement area. The subsequent step would be a normal, non-perturbed step. As was discussed above, walking speed was constant and hence one would expect the ground reaction forces to be consistent and that was the case as shown in the positive peaks in panels A and B. Further, there was no clear separation as a consequence of foot type. The braking peak forces (negative peaks) in panels C and D were not as clear. When the prosthetic leg was the supporting leg, panel D, data were consistent with the hypothesis – not sensitive to the step-length changes. However, on the intact side there was much more variation across conditions and between foot types. The small participant number and the variability within the cells prevents further discussion but there is clearly room for further investigation.

The peak force data for the braking forces on the lead or target leg were consistent with the hypothesis. At the shortened step-length the peak braking force was less than the normal step length while the peak force in the long step-length condition was greater. Similarly, the propulsions forces on the trailing leg were consistent with the hypothesis and there was an evident interplay between these legs. Consistent between-feet differences were difficult to confirm. The peak forces on the SACH were typically higher than the on the own foot, which was a dynamic-response foot, on both the targeting and trailing foot. To gain further insights into these differences the plots of the normalized anterior-posterior force versus the percentage of stance duration are displayed in Figures 6 and 7.

Summary. While the small sample size of the experimental group made it difficult to make strong conclusions, it was nonetheless clear that there was an interaction between trailing and leading foot. This was not overly surprising as during the double-support phase both feet are in contact with the ground. The pattern of this interaction was not overly affected by the type of foot worn by the participants. Using a smooth walking surface like that in the motion capture lab provided a predictable surface for the participants and in so doing masked any potential differences that could be seen between the foot type. A stronger evaluation could be made with a surface that was irregular, similar to walking surfaces outside the laboratory. This masking effect was compounded by the strong speed control that was implemented. In a more realistic environment, participants would change their walking speed as they approached uncertain situations. However, anecdotally, the participants felt much less competent with the SACH-type of foot. This could have reflected a degree of discomfort with this foot since most wore a more effective dynamic-response foot. On the other hand, it was possible that the perturbation was not sufficiently strong to evoke substantial changes in the walking performance of this group. There was an increased degree of complexity in the gait of individuals with a uni-lateral trans-tibial amputation that serves to increase within- and between-subject variability on most biomechanical variables and thus reduce the ability of the investigators to illuminate differences in gait in response to environmental changes. Clearly, to expose the elemental differences in the gait patters adequately and in response to different foot types a more substantial study is warranted.

This pilot study has provided areas requiring further work. To elicit differences in the gait patterns in response to targeting or related perturbations a more complex design is required. For example, a walking surface that is similar to current sidewalks would be preferable. Allowing participants to vary their speed would also provide a better approximation of activities of daily life. Walking is a complex task that requires an interaction with the ground, sequenced muscle contractions and on-line continuous control by the central nervous system. The added impact of the lower-limb amputation, uni- or bi-lateral results in an even more complex system. The next studies will need to be more complex deal with the subtle differences that arise as one moves around the environment.

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