ANALYSIS OF DYNAMIC BALANCE CONTROL IN TRANSTIBIAL AMPUTEES

by

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A THESIS

Presented to the Department of Human Physiology
and the Robert D. Clark Honors College
in partial fulfillment of the requirements for the degree of
Bachelor of Science

June 2015

An Abstract of the Thesis of

Shaun Resseguie for the degree of Bachelor of Science
in the Department of Human Physiology to be taken June 2015

Title: Analysis of Dynamic Balance Control in Transtibial Amputees

Approved: [redacted]

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The powered prosthetic foot (PPF) is designed to provide transtibial amputees (TTA) with active propulsion and plantar flexion similar to that of the biological limb. Previous studies have demonstrated the PPF's ability to increase TTA walking speeds, while reducing the energetic costs, however, little is known about its effects on dynamic balance control. The purpose of this study was to assess dynamic balance control in a sample of TTA subjects during level-ground walking and obstacle crossing tasks. Control subjects (n=5) and TTA subjects (n=3) were instructed to complete a series of functional walking tasks during each lab visit. The TTA subjects completed the walking protocol twice, first in their passive energy-storing prosthetic foot (ESPF) and again in the prescribed PPF after two weeks of acclimation. Motion data were collected via a 10-camera system with a 53-marker and 15-segment body model. Center of mass (CoM) motion within the frontal plane was analyzed and used as a functional indicator of dynamic balance control. Preliminary findings from the study indicate that TTA that are less able to maintain dynamic balance control may benefit from the PPF in restricting their M-L CoM motion. However, robust walkers who demonstrate balance control with the ESPF may be adversely affected by the PPF.

Acknowledgements

I would like to thank Professor Hahn for taking the chance and providing me with this opportunity to work at the Bowerman Sports Science Clinic, despite my lack of knowledge in biomechanics at the time. I would also like to thank the graduate students for guiding me along the various aspects of research, the local prosthetist, Ryan Keele, for volunteering his time to ensure proper fitting of the prosthesis throughout the study, and undergraduate students for assisting me in my data collections. I am honored to have had the privilege of undergoing this strenuous but rewarding process as an individual. I have grown tremendously through this journey and gained invaluable lessons that will help my in my future endeavors.

Table of Contents

Introduction	1
Methods	4
Results	8
Discussion	13
References	17

Introduction

There are nearly two million people living with limb loss in the United States, nearly two thirds of which are lower limb amputees (Ziegler-Graham et al., 2008). A majority of these individuals are prescribed some form of a prosthetic foot to compensate for the lost limb and restore some functional independence. Prosthetic feet options have traditionally been limited to passive models of basic, articulated, or dynamic-response styles. The solid ankle cushion heeled (SACH) foot is an example of the basic model that is rigid without any moving parts. The foot is fitted with a rubber wedge that compresses under the amputee's weight to cushion the heel during walking. The foot is stable and does not provide mediolateral (side-to-side) movement. The articulated foot, on the other hand, is fitted with either a single axis or multiple axes. The single-axis foot simulates the ankle joint in allowing the foot to move up and down in an anteroposterior (forward and back) fashion to provide knee stability. Similarly, the mutli-axis foot provides anteroposterior motion along with mediolateral movement, which increases the wearer's ability to navigate uneven terrains. Lastly, there is the more advanced dynamic elastic response or energy-storing prosthetic foot (ESPF). The ESPF foot has a flexible keel in the forefoot that acts much like a spring, such that when an individual presses down on the foot, energy is stored elastically through the stance phase and released during toe-off to provide a sense of propulsion as the keel returns to its original shape. The energy return, however, is always less than the input as some of the energy is lost as heat. Numerous prosthetic foot models have been developed to fit the various needs of amputee's, ranging from walking, running, cycling, skiing, to dancing. Each foot is designed specifically to compensate for the loss of certain abilities

of the biological limb. However, no prosthetic foot is capable of restoring all of the functions.

Despite the advances in prosthetic feet, 54% of individuals with a unilateral lower-limb amputation reported having fallen within the previous year, 75% of which had fallen more than once, and 40% of the fallers sustained an injury as a direct result (Miller et al., 2001). In an earlier study, 80% of the amputees attributed their falls to a loss of balance (Ülger et al., 2010). In the case of transtibial amputees (TTA), the loss of sensory input, muscle contraction, and joint manipulation below the knee that normally function together for balance control predisposes them to a greater risk of falling compared to their age-matched able-bodied counterparts (Miller et al., 2001). Recently within the past decade, the powered prosthetic foot (PPF) became an available option. The BiOM Ankle System (iWalk, Bedford. MA) is the first of such design that is capable of providing net positive power to propel the individual forward during locomotion. The PPF is battery powered and mimics the human ankle joint by providing active ankle plantar flexion, the act of pointing one's foot downward to push off during walking. Independent studies have found that the PPF is capable of increasing an individual's walking speed by 21% while decreasing their metabolic cost by 8% (Herr et al., 2012). However, little is known about how the reintroduction of active propulsion and plantar flexion affects amputee dynamic balance control, defined as one's ability to maintain full body equilibrium during motion.

Since dynamic balance control cannot be directly measured, we chose to utilize an individual's M-L center of mass (CoM) motion within the frontal plane (plane that dissects the body into a front and back half) during the task of obstacle-crossing as our

indicator dynamic balance control. The task of obstacle-crossing requires a complex set of motor skills because of the inherent asymmetry of motion between the lead- and trail-limb. As a result, any imbalance experienced by the individual would amplify their M-L CoM motion. For example, able-bodied young individuals that demonstrated successful control strategies were less likely to experience an increase in M-L CoM motion when negotiating obstacles of different heights, which did not hold true for individuals with compromised balance control (Chou et al., 2003). In the same study, it was identified that individuals at risk of falling demonstrated the greatest M-L CoM motion at obstacle heights of 2.5% of subject height (Chou et al., 2003). Therefore, a large increase in M-L CoM motion during obstacle-crossing could be used as an indicator of dynamic balance control and identify individuals at a greater risk of falling.

The PPF, BiOM, has been designed to mimic the dynamics of the biological limb in both its ability to actively plantar flex and propel the individual forward, both of which are critical for functional locomotion and balance control. The purpose of this study was to examine the compensatory mechanisms of TTA gait and balance control during level ground locomotion and obstacle-crossing while wearing the PPF, in order to determine whether the reintroduction of active plantar flexion and propulsion would lead to a more stable dynamic balance control via measurements of M-L CoM motion. We hypothesized that 1) as walking speeds decreased and obstacle heights increased, CoM motion would increase; 2) the amputee group would experience a greater CoM motion in all conditions compared to their able-bodied counterparts; 3) the TTA would experience an increased CoM motion when wearing the PPF compared to the ESPF.

Methods

Subjects

Three TTA subjects (2M/1F; 55.7 ± 8.8 years) and five able-bodied control subjects (4M/1F; 29.0 ± 12.0 years) were recruited for this study from the University of Oregon community and local prosthetic clinic. The experimental protocol was approved by the Institutional Review Board of the University of Oregon. Experimental procedures were verbally explained and informed consent was obtained prior to testing. The TTAs were recruited based on the following inclusion criteria: unilateral transtibial-level amputation, duration since amputation greater than six months, currently ambulating with a passive prosthetic foot, able to understand and follow verbal directions in English, and no more than 250lbs in weight. Subjects were excluded from the study if they had a history of rheumatic diseases, history of neurologic deficits or other musculoskeletal disorders that would affect gait, or in need of an upper extremity gait aid.

Protocol

The experimental protocol consisted of measurements of height, weight, and body segment anthropometrics, including knee width, ankle width, and functional leg length at the beginning of each session. Retro-reflective markers were placed on the subjects prior to testing. Each subject was then required to perform a series of functional movements, followed by unobstructed level ground and obstacle crossing tasks down a 20-meter walkway. The level ground task comprised of three different self-selected walking speeds of slow (SSS), normal (SSN), and fast (SSF). The

obstacle-crossing task comprised of two different obstacles heights of 3cm (Low) and 12.7cm (High), representing the height of doorway thresholds and precast parking lot concrete blocks, respectively. The obstacles consisted of two adjustable vertical supports and a padded horizontal cross bar. The cross bar was made with a 1cm diameter metal rod encased by a 5cm exterior diameter foam casing to protect the subject. The obstacle was created such that the horizontal foam rod would easily roll-off the vertical supports if it comes into contact with the foot in an effort to minimize tripping. The subjects were allowed to select their preferred crossing limb and to identify their starting location to ensure a comfortable crossing-stride when navigating the obstacle. The crossing-stride was defined as the heal-strike of the trailing-limb just before the obstacle to the subsequent heel-strike of the same foot.

All the subjects performed the conditions ten times in the following order: SSN, SSS, SSF, Low, High. The TTA performed each of the above-mentioned trials a second time at a later session. The first session was performed in their personal ESPF and the second session was performed in the provided PPF. Upon completion of the first session. a certified prosthetist fitted the TTA with the PPF and fine-tuned the settings to optimize the foot for each subject. The amputee subjects were given two weeks for acclimation in an effort to minimize noise in the data as a result of unfamiliarity of the prosthetic foot. At the end of the two weeks, the subjects returned to the lab and completed the same level-ground and obstacle-crossing tasks in the PPF. Breaks were provided as needed during data collection.

Kinematic Data

All subjects were tested with the same experimental protocol at the Bowerman Sports Science Clinic laboratory within the University of Oregon, Eugene, Oregon. Whole-body kinematic data were collected via a 10-camera system (Motion Analysis Corp., Santa Rosa, CA) with a static marker set of 63 retro-reflective markers and dynamic marker set of 53 retro-reflective markers. Three-dimensional marker trajectories were collected at 120 Hz and low-pass filtered using a 4th-order Butterworth filter with a cutoff frequency of 6 Hz. The retro-reflective markers were placed on bony landmarks to create unique 15-segment full body models of each subject (Figure 1) via Visual3D (C-Motion Inc., Germantown, MD). Virtual markers were created through functional movements to identify hip and knee joint centers. The whole body CoM location and motion were computed as the weighted sum of the 15 segments (head, trunk, pelvis, upper arms, lower arms, hands, thighs, shanks, and feet). The validity of this CoM estimation technique was demonstrated previously in 13-segment body models that excluded the hand segments (Chou et al., 2001). The M-L CoM range of motion (difference between the maximum and minimum displacement achieved during the crossing stride) within the frontal plane was then computed and used as functional indicators of dynamic balance control (Chou et al., 2003).



Figure 1: 15-segment full body model generated from the static mark set.

Statistical Analysis

Student's t-tests were used to calculate the significance of our findings. Analysis of the condition effect across the subject groups was tested via a two-tailed paired t-test with an α-value of 0.05 (Table 1). Analysis of the subject effect across the conditions utilized different types of t-tests (Table 2). For instance, comparisons within the TTA groups (ESPF and PPF) used a two-tailed paired t-test, while comparisons between the control group and TTA groups (ESPF and PPF) used a two-tailed unpaired t-test. As a result of a lack of matched control subject for the TTA, we assumed an unequal variance in our data. To adjust for the assumption, a Bonferroni correction was used, which resulted in an α-value of 0.017.

Results

Table 1 shows the averages and standard deviations of subject group walking speeds for each condition (SSF, SSN, SSS, Low, and High). The significance of the condition effect was tested across different self-selected walking speeds and obstacle heights. On average, the three groups demonstrated a convergence in their walking speeds over the different self-selected walking speeds. Combined walking speeds decreased significantly from conditions SSF to SSN ($p<0.001$), SSN to SSS ($p<0.001$), and SSF to SSS ($p<0.001$). The control group consistently walked faster than the TTA subjects in the three self-selected walking speeds. As for the obstacle height conditions, we saw a significant increase in all three groups' walking speeds as obstacle heights increased from SSN (no obstacle) to Low ($p=0.010$) and a significant decrease from Low to High ($p=0.012$). There was not a significant difference between group walking speeds when comparing between conditions SSN and High. While both the control group and ESPF group experienced a decrease in walking speeds from SSN to High, the PPF group experienced an increase in walking speeds instead.

	Condition										Walking Speed Effect			Obstacle Height Effect		
	SSF		SSN		SSS		Low		High		SSF:SSN	SSN:SSS	SSF:SSS	SSN:Low	Low:High	SSN:High
	Average	SD	Average	SD	Average	SD	Average	SD	Average	SD	P-value			P-value		
Contro Group	1.647	0.141	1.361	0.106	1.042	0.144	1.434	0.170	1.303	0.114	<0.001	<0.001	<0.001	0.121	0.015	0.019
ESPF Group	1.620	0.211	1.341	0.158	0.908	0.047	1.359	0.124	1.324	0.068	0.027	0.038	0.022	0.538	0.543	0.807
PPF Group	1.638	0.179	1.305	0.050	0.990	0.090	1.386	0.063	1.356	0.113	0.069	0.053	0.052	0.032	0.505	0.298
Combined	1.635	0.014	1.336	0.028	0.980	0.068	1.393	0.038	1.328	0.027	<0.001	<0.001	<0.001	0.010	0.012	0.466

Table 1. Averages and standard deviations of subject group walking speeds (m/s) for each condition (α=0.05).

	Control Group		ESPF Group		PPF Group		Passive v. Powered	Control v. Powered	Control v. Passive
	Average	SD	Average	SD	Average	SD	P-Value		
SSF	0.026	0.007	0.030	0.004	0.035	0.001	0.204	0.059	0.341
SSN	0.030	0.010	0.043	0.005	0.044	0.012	0.926	0.130	0.091
SSS	0.039	0.010	0.057	0.008	0.053	0.006	0.092	0.060	0.037
Low	0.026	0.011	0.041	0.010	0.042	0.013	0.674	0.103	0.088
High	0.036	0.006	0.055	0.013	0.054	0.017	0.897	0.059	0.024

Table 2. Averages and standard deviation of subject group M-L CoM motion (m) for each condition (α=0.017).

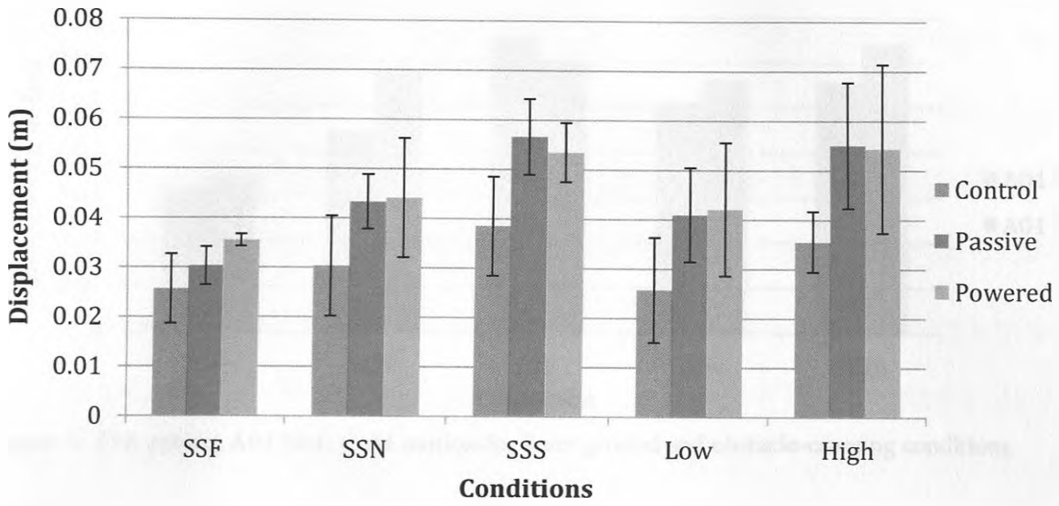


Figure 2. M-L CoM motion for level ground and obstacle-crossing conditions.

Figure 2 shows the averages and standard deviations of the M-L CoM motion in the frontal plane for each condition across the subject groups. All three groups experienced an increased M-L CoM motion when walking speeds decreased and obstacle heights increased from Low to High. The TTA showed a greater M-L CoM motion compared to the controls across all conditions regardless of the type of prosthetic foot worn. These differences were magnified as the walking speeds decreased and obstacle height increased. None of the observed differences, however, were significant (Table 2). This was the result of the fact that each TTA responded uniquely to the PPF in terms of their M-L CoM motion. When wearing the PPF, TTA subject A01 experienced a greater CoM motion for all the conditions except for SSS (Figure 3). TTA subject A02, on the other hand, only experienced an increased M-L CoM motion for conditions SSN and Low while wearing the PPF (Figure 4). Additionally, TTA subject A03 only experienced an increased M-L CoM motion in conditions SSF and SSN (Figure 5). As a result, there was a large amount of variability within the data.

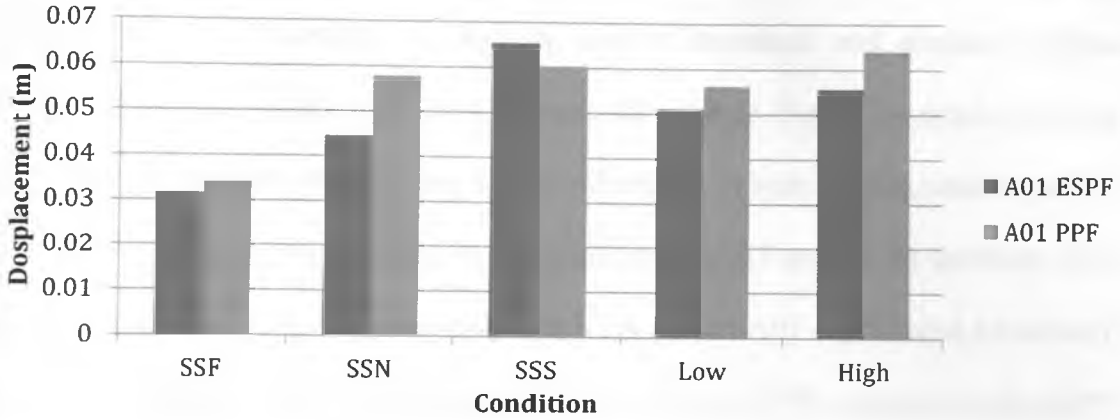


Figure 3. TTA subject A01 M-L CoM motion for level ground and obstacle-crossing conditions.

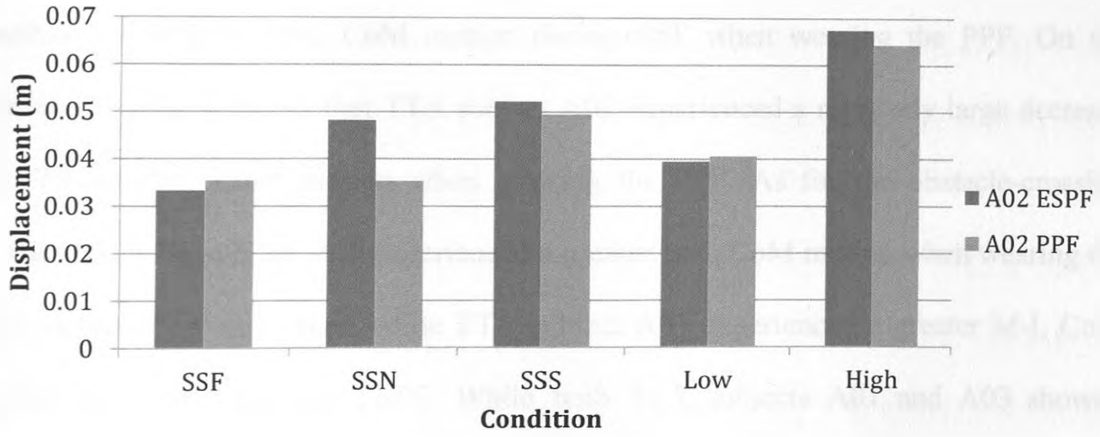


Figure 4. TTA subject A02 M-L CoM motion for level ground and obstacle-crossing conditions.

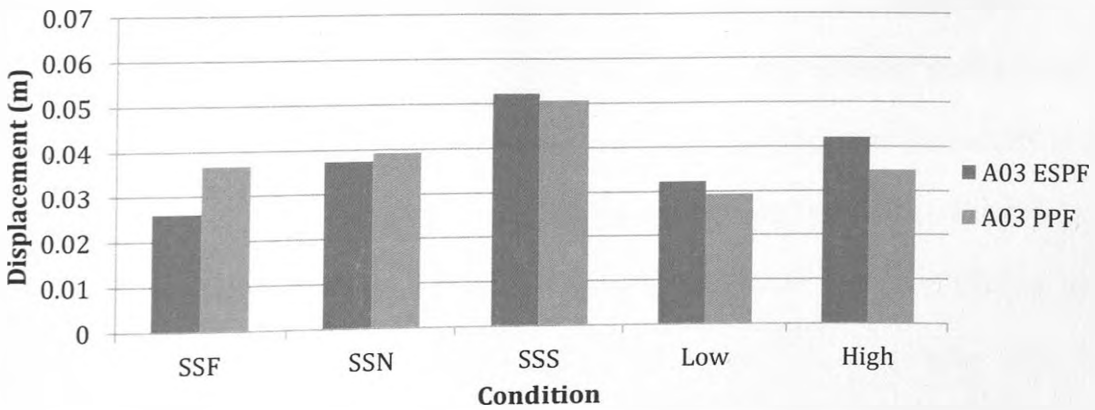


Figure 5. TTA subject A03 M-L CoM motion for level ground and obstacle-crossing conditions.

Figures 3-5 show that all the TTA subjects also independently experienced the increased M-L CoM motion as walking speeds decreased and obstacle heights increased. However, variations existed between the subjects. For TTA subjects A01 and A03, the PPF had resulted in a larger M-L CoM motion in both walking conditions SSF and SSN, while the ESPF resulted in the larger M-L CoM motion for condition SSS (Figures 3 and 5). It is also worth noting that TTA subject A01 experienced a relatively large 30% increase in M-L CoM motion when wearing the PPF compared to the ESPF during condition SSN. Similarly, TTA subject A03 also experienced a relatively large increase of 41% in M-L CoM motion during SSF when wearing the PPF. On the contrary, Figure 4 shows that TTA subject A02 experienced a relatively large decrease of 27% in M-L CoM motion when wearing the PPF. As for the obstacle-crossing conditions, TTA subject A01 experienced a greater M-L CoM motion when wearing the PPF in both Low and High, while TTA subject A03 experienced a greater M-L CoM motion when wearing the ESPF. While both TTA subjects A01 and A03 showed relatively little changes in M-L CoM motion as obstacle height increased, A02 experienced a relatively large increase in both types of prosthetic foot (Figure 4).

Discussion

Injuries associated with falling continue to be a major concern for lower-limb amputees despite the continual advancement in the field of prosthesis. Previous studies have analyzed the different TTA compensatory techniques utilized during obstacle crossing and how such adaptations affected subject dynamic balance control, yet neither looked at the effects of the PPF (Vrieling et al., 2007 and Hak et al., 2013). The majority of studies on the PPF have focused on its effect on lower-limb amputees' capacity for locomotion, specifically walking speeds and energetic efficiency (Au et al., 2007 and Herr et al., 2012). As a result, this study aimed to provide knowledge within the literature gap by investigate the effects of the PPF's reintroduction of active plantar flexion and propulsion on TTA dynamic balance control.

We hypothesized that M-L CoM motion would increase as walking speeds decreased and as obstacle height increased, and that TTA would consistently demonstrate a larger M-L CoM motion compared to that of able-bodied individuals. While trends of both the above statements were observed, they were not significantly correlated. Despite a lack of significance, our observed trends in M-L CoM motion for TTA subjects wearing ESPF and able-bodied subjects during different walking speeds and obstacle heights are similar to the findings of existing literatures (Orendurff et al., 2004 and Chou et al., 2001 and 2003). We were able to identify significant correlations for the condition effect on subject walking speeds. This is an important finding, as it confirms the effectiveness of our protocol in prescribing conditions that are significantly different from one another. However, we did not observe a significant difference in the group effect on M-L CoM displacement between TTA wearing the

PPF and ESPF. Individual TTA demonstrated high variability in their M-L CoM motion, which suggest that response to the reintroduction of active plantar flexion and propulsion is condition and subject specific.

The lack of significant findings in the group effect is likely attributed to the fact that we had only three TTA subjects participate in the study. We speculate that the PPF is more suitable for certain TTA. For example, TTA subject A02's large decrease in M-L CoM displacement during SSN indicates that the PPF could actually increases her day to day walking stability (Figure 4). This could be as a result of the fact that she was a less robust walker compared to the other two TTA subjects. This speculation is supported by the fact that her M-L CoM displacement increased drastically between conditions Low and High (Figure 4), while TTA subjects A01 and A03 maintained a relatively small increase (Figures 3 and 5). The large increase exemplifies her reduced ability for balance control in the presence of a balance challenge, which comes in the form of an obstacle in this case. This finding is also consistent with our qualitative observations of her general physical fitness during the study sessions. It turned out that she had benefited from the PPF so much that she was actively pursuing one for herself. She felt more mobile because of the reduced energetic cost of walking and her ability to walk faster. She even claimed that wearing the PPF allowed her, for the first time since amputation, to forget the fact that she was missing a limb.

Subject A02's fondness of the PPF, however, was not equally shared by the other two subjects. Both subjects A01 and A03 expressed that the PPF was not suitable for them, but for different reasons. A03 was relatively indifferent of the PPF's functionality, but expressed concerns about the environmental limitations, such as its

inability to get wet and burden of having to carry backup batteries. TTA subject A01, on the other hand, disliked the PPF. He felt that he was in less control of his movements and was "driven" by his foot. His increased M-L CoM motion across all of the conditions, except for SSS, supports his claim (Figure 3). This is likely as a result of the fact that he was an active individual that was capable of effective locomotion with his ESPF.

An important limitation of this study was the small number of subjects that participated. Given the fact that the control subjects were substantially younger and not age, sex, height, and weight matched to the TTA subjects, some of the study outcomes may have been due to the difference in age or general physical ability of the individuals. Not to mention, prosthetist tuning of the PPF may not be at the ideal setting given only two weeks of acclimation time. Lastly, the motion capture system used, restricted the study to a lab setting and does not ideally exemplify real life challenges.

Conclusion

In conclusion, there appeared to be a high degree of variability in TTA response to the PPF. The PPF seemed to be beneficial in restricting the M-L CoM motion for TTA that are less mobile and less capable of maintaining their dynamic balance control. However, the PPF could also have the opposite effect on robust TTA walkers that demonstrate control of their M-L CoM motion with the traditional ESPF. The findings and speculations of the study, if supported by further data, could assist prosthetists in determining the types of TTA that would benefit the most from a PPF.

Conflict of Interest Statement

None of the authors have any conflict of interest with the study.

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