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The effects of walking speed and prosthetic ankle adapters on upper extremity dynamics and stability-related parameters in bilateral transtibial amputee gait

Matthew J. Major¹, Rebecca L. Stine^{1,2}, and Steven A. Gard^{1,2}

¹Northwestern University Prosthetics-Orthotics Center, Northwestern University Feinberg School of Medicine, Chicago, IL

²Jesse Brown VA Medical Center, Chicago, IL

Abstract

Bilateral transtibial amputee (BTA) gait has been investigated less and is not as well understood compared to that of their unilateral counterparts. Relative to able-bodied individuals, BTAs walk with reduced self-selected speeds, increased step width, hip-hiking, and greater metabolic cost. The clinically observed upper body motions of these individuals have not been quantified, but appear substantially different from able-bodied ambulators and may impact upright balance. Therefore, the objective of this study was to characterize the upper extremity kinematics of BTAs during steady-state walking. We measured medial-lateral ground reaction forces, step width and extrapolated center-of-mass (XCoM) trajectory, and observed effects of walking speed and increased prosthetic ankle range-of-motion (ROM) on these parameters. Significantly, BTAs display greater lateral trunk flexion ROM and shoulder abduction than able-bodied individuals when walking at similar speeds, and the inclusion of prosthetic adapters for increasing passive ankle ROM slightly reduced step width. Overall, exaggerated lateral trunk flexion ROM was invariant with step width. Results suggest that lateral trunk motion is useful for shifting the body center-of-mass laterally onto the leading stance limb while simultaneously unloading the trailing limb. However, exaggerated lateral trunk flexion may introduce an unstable scenario if the XCoM is displaced beyond the lateral base-of-support. Further studies would be useful to identify if either prostheses that assist limb advancement and/or gait training may be effective in reducing this lateral sway while still maintaining efficient ambulation.

Keywords

Bilateral; transtibial amputation; gait; center-of-mass; trunk

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Correspondence to: Matthew J. Major.

Contact Information for Matthew J. Major (corresponding author): Northwestern University Prosthetics-Orthotics Center, Northwestern University 680 N Lake Shore Dr, Suite 1100, Chicago, IL, USA, 60611 Telephone: 1 312 503 5731; Matthew-Major@Northwestern.edu.

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1.0 Introduction

Compared to unilateral transtibial amputees, bilateral transtibial amputee (BTA) gait has received relatively less attention and this has limited our understanding of the unique rehabilitation needs of these individuals [1]. Recent investigations have demonstrated that BTAs walk with substantially reduced self-selected speeds and increased step width compared to able-bodied individuals [2]. Furthermore, despite close to normal knee and hip joint kinematics during the swing phase of gait, these individuals display considerable hip-hiking [2]. These compensatory gait mechanisms may contribute to their increased metabolic cost, which is reportedly 1.3-2.2 times higher than able-bodied individuals [3-5]. Importantly, exaggerated upper body motions, such as lateral trunk sway and abducted arms, have been observed clinically. In order to inform prosthetic prescription guidelines and other rehabilitation interventions for ensuring safe and efficient mobility of these individuals, it is important to understand the role these upper body motions play during gait and their relationship with upright balance.

Upright balance during gait is of significant concern as lower limb amputees are at an increased fall risk, and hence fall-related injuries, compared to able-bodied individuals. Indeed, 52.4% of community-dwelling unilateral lower limb amputees reported at least one fall within the past year [6] and have reduced balance confidence [7]. This increased fall risk is a function of reduced capability to produce rapid compensatory movements to maintain the body center-of-mass (BCoM) within the base-of-support (BoS) [8, 9], resulting from diminished sensory feedback [10-13], reduced muscle strength [14, 15], and inability to actively adjust the mechanical behavior of the prosthesis (i.e., joint movement and impedance) [9, 16-20]. BTAs are at a further disadvantage as they lack additional compensatory mechanisms otherwise provided by the sound limb.

Anecdotally, it appears that the motions of the arms and trunk during BTA gait may play a substantial role in ambulation and upright balance. However, their upper body dynamics have not yet been characterized, thereby limiting understanding of their function and effects on gait. Therefore, the primary objective of this retrospective study was to quantify the upper body kinematics of BTAs during steady-state walking. A secondary objective was to observe the effects of walking speed and prosthetic ankle adaptors that allow for increased ROM on these characteristics to provide insight into gait adaptations and prescription guidelines. Further insight into these dynamics may help inform intervention strategies to satisfy the unique ambulatory and rehabilitation needs of these individuals with the objective of maximizing upright balance.

2.0 Methods

2.1 Subjects

A retrospective analysis was performed on ten BTA subjects (50 ± 18 yrs, 1.73 ± 0.08 m, 82 ± 16 kg) from a previous investigation (ethical approval and informed consent previously obtained) [2, 21, 22]. Subject inclusion criteria required that they: walk freely without the use of an aid; be a minimum of two years post-amputation; have independent functional ambulation; and possess no other pathological conditions that might affect ambulation or balance. Amputation etiology included: vascular ($n=5$), traumatic ($n=3$), congenital ($n=1$), and meningitis ($n=1$). Data for 13 age- and speed-matched able-bodied controls (51 ± 6 yrs, 1.72 ± 0.09 m, 74 ± 15 kg) were used for comparison.

2.2 Experimental protocol

Subjects performed overground walking trials along a ten meter level walkway at three self-selected speeds in this order: 1) “normal,” 2) “fastest comfortable” (fast), and 3) “slowest

comfortable” (slow). Data were collected while subjects walked with two prosthetic configurations (same configuration for each leg) that are representative of different levels of prosthetic ankle joint motion: **PC1**) Seattle Lightfoot II prosthetic foot (Seattle Systems, Poulsbo, WA), and **PC2**) Seattle Lightfoot II foot, Multiflex Ankle (Endolite, Miamisburg, OH), and Torsion Adapter (Ottobock, Duderstadt, Germany). The combined adapters of PC2 provide additional motion in the sagittal, frontal, and transverse planes and were thus used to observe prosthetic effects on all planes of kinematic movement. Components were individually prescribed based on subject height, weight, and activity level, and were attached to the subjects’ own socket and suspension for testing. Prostheses were aligned by the same certified prosthetist, PC1 was always tested before PC2, and subjects were permitted two weeks for acclimation prior to data collection.

2.3 Data collection and analysis

Kinematic data were collected with a digital motion capture system (Motion Analysis Corporation (MAC), Santa Rosa, CA) at 120 Hz using a lower extremity Helen Hayes marker set [23] with additional markers on the right and left acromion process and lateral epicondyle of the humerus. Kinetic data were collected at 960 Hz with six force plates (AMTI, Watertown, MA) embedded within the walkway. Upper body kinematics (i.e., shoulder abduction and trunk lateral flexion, transverse rotation, and forward flexion), ankle kinematics (i.e., plantar/dorsiflexion and transverse rotation), step width, medial-lateral ground reaction force (MLGRF), and speed were determined across five walking trials for each walking and prosthetic condition using OrthoTrak software (MAC). The trunk reference frame was defined by the two acromion markers and pelvis reference frame center: medial-lateral axis—line connecting acromion markers; inferior-superior axis—line connecting midpoint between acromion markers and pelvis reference frame center; anterior-posterior axis—orthogonal to the plane formed by the other two axis. All trunk angles are estimated relative to the laboratory reference frame (i.e., zero angular displacement represented by coincidence of the trunk reference frame with the laboratory reference frame). Shoulder abduction was defined as the relative angle between the inferior-superior axis of the trunk and upper arm segment defined by the acromion and epicondyle marker. Kinematic and kinetic data were filtered with a bidirectional 4th order low-pass Butterworth filter at 6 and 10 Hz, respectively. Right-left gait symmetry was established for these subjects in a previous investigation [21] and data were averaged across both legs for each subject over a minimum of ten steps for all conditions. Kinetic data were normalized by subject body weight (BW).

BCoM position for each subject was estimated in OrthoTrak from the anthropometric segmental mass properties of able-bodied individuals. The XCoM—an estimated vertical projection of the BCoM plus its velocity component as normalized by the leg eigenfrequency [8, 9]—was calculated using custom software in Matlab (Mathworks, Natick, MA). The XCoM can be used to provide an estimation of the *lateral margin of stability*, which is the instantaneous distance between the position of the XCoM and approximated center-of-pressure (CoP) under the stance limb representative as the lateral BoS. A negative margin of stability (i.e., XCoM beyond the lateral edge of the BoS) indicates a dynamically unstable scenario when action must be taken by the ambulator to redirect the XCoM and avoid a fall [8, 9].

Because the BTA subjects walked slower than the able-bodied controls at the slow and normal speeds, only data from the BTA subjects’ fast walking trials can be used for speed-matched comparisons with able-bodied data.

2.4 Statistical analysis

Kinematic data were averaged over all ten BTA subjects, while kinetic data were averaged over nine subjects due to a hardware malfunction. A two-way repeated-measures ANOVA was used to statistically analyze between-speed differences of amputee parameters, effects of prosthetic configuration, and interactions between speed and prosthetic configuration. A Bonferroni adjustment was used for multiple comparisons during posthoc analyses. An independent t-test was used to compare amputee and able-bodied data. Relationship between two variables was assessed by the Pearson's Correlation. Data normality and sphericity were confirmed using the Shapiro-Wilk and Mauchly's test, respectively. If the sphericity assumption was violated, a Greenhouse-Geisser correction was used and the results interpreted accordingly. Statistical analyses were conducted using SPSS (IBM, Armonk, NY) and the critical alpha was set at 0.05.

3.0 Results

BTA subjects displayed an average increase in prosthetic ankle motion of $2^{\circ} \pm 3^{\circ}$, $2^{\circ} \pm 4^{\circ}$, and $3^{\circ} \pm 4^{\circ}$ in transverse ROM ($p=0.096$), and $5^{\circ} \pm 3^{\circ}$, $6^{\circ} \pm 3^{\circ}$, and $5^{\circ} \pm 3^{\circ}$ in sagittal (i.e., plantar/dorsiflexion) ROM ($p<0.001$) with the use of PC2 during slow, normal, and fast walking, respectively.

Peak shoulder abduction angles of the amputee subjects were greater than able-bodied controls ($p<0.001$ at similar walking speed; Figure 1), and was directly proportional with speed ($p=0.001$). Amputees displayed a sinusoidal profile of lateral trunk flexion (Figure 2A) with a ROM that was substantially greater than controls ($p=0.002$ at similar walking speed; Figure 2B). They also displayed a slight increase in forward trunk lean compared to controls (Figure 2C) at similar walking speeds ($p=0.035$; Figure 2D) and the amount of lean was directly proportional with speed ($p=0.007$). Furthermore, amputees displayed a similar temporal profile of trunk rotation as controls (Figure 2E) with a slight increase in ROM ($p=0.048$ for PC1; $p=0.058$ for PC2) and an inverse relationship with speed ($p=0.002$). Although use of PC2 increased lateral trunk flexion ROM and peak shoulder abduction and decreased trunk rotation ROM compared to PC1, these differences were not significant ($p=0.083$).

A relatively small reduction in step width was observed with increasing speed ($p=0.017$; Figure 3) and use of PC2 ($p=0.009$), while lateral trunk flexion ROM remained invariant ($p=0.410$; Figure 2B). Step widths of the amputees were greater than controls at all speeds and for each prosthetic configuration ($p=0.002$ at similar speed). The behavior displayed in Figure 4 was observed in all amputee subjects and none of the controls, in which the XCoM exceeded the lateral borders of the BoS (Figure 4C: average instantaneous lateral distance between peak XCoM position and CoP was 2.3cm) during fast walking and use of PC2. Lateral trunk flexion was in-phase with the lateral motion of the BCoM and XCoM, with neutral flexion occurring near initial contact and maximum flexion occurring just after contralateral toe-off as speed increased (Figure 4B and 4D).

A direct correlation was found between step width and MLGRF for both amputee and able-bodied data ($p=0.05$, $r=0.664$). Consequently, MLGRF were normalized by subject weight and also average step width for comparison. Both sets of MLGRF increased with speed ($p=0.003$), but were not significantly different between prosthetic configurations ($p=0.116$). While amputee MLGRF normalized by weight were greater than controls at all speeds and for each prosthetic configuration (significant at $p=0.012$ for similar speed), forces also normalized by step width were consistently less than controls (Figure 5) although these differences were not significant ($p=0.112$).

No interaction effects (walking speed $\times$ prosthetic configuration) for any variable were found to be significant ($p = 0.389$).

4.0 Discussion

BTAs displayed significantly greater peak shoulder abduction and lateral trunk flexion ROM than able-bodied controls at comparable speeds. Across all conditions, the average lateral trunk flexion ROM displayed by the BTAs ($11^\circ \pm 5^\circ$) was greater than that reported for unilateral transfemoral amputees ($9^\circ \pm 3^\circ$) [24]. Despite the substantial difference in lateral trunk motion, the amputees' forward lean and rotation of the trunk remained similar to controls. Therefore, exaggerated motion is primarily isolated to the coronal plane and is accompanied by significantly greater step width. BTAs appear to walk with a wide step width to maintain a larger medial-lateral BoS due to a perception of instability [2]. Furthermore, an increased step width may be used as a safety precaution for maintaining the BCoM and XCoM above the BoS in case of inaccurate foot placement [9]. Although minimal, step width appears to be reduced through an increase in speed (average decrease of 1.4cm for fast versus slow walking) and by providing increased prosthetic motion through the addition of ankle and torsion adapters (average decrease of 1.8cm for PC2 versus PC1). Given this abducted gait, such exaggerated shoulder ROM and lateral trunk flexion may be expected as BTAs must rapidly transfer their BCoM from one foot to the other across this wide BoS [5]. These subjects may further abduct their shoulders to facilitate this transfer and promote medial-lateral stability [25], with increased necessity at higher speeds. As expected, medial-lateral forces increased with speed, but the normalized forces were comparable to those of able-bodied individuals walking at similar speeds.

It is reasonable to assume that with a reduction in step width, BTAs would require less lateral trunk flexion. However, lateral trunk flexion ROM remained close to the average of 11° irrespective of speed and prosthetic condition. This may suggest that lateral BCoM shift is not the primary reason for producing this exaggerated lateral trunk flexion and that such motion provides an alternative, consistent benefit regardless of step width. Similar to unilateral transtibial and transfemoral amputees [26], BTAs display hip-hiking during the swing phase, which has been attributed to the need for producing sufficient prosthetic foot-ground clearance during swing phase [2, 5, 26]. Therefore, it may be possible that lateral trunk flexion provides a mechanism to assist with pelvic obliquity, as pelvic rise has been previously observed to be coupled with lateral trunk flexion [27].

Not surprisingly, lateral trunk flexion was nearly in-phase with the BCoM lateral position. Furthermore, when observing the temporal profile of lateral flexion (Figure 4), lateral trunk flexion commenced prior to ipsilateral initial contact and peak lateral trunk flexion occurred after contralateral toe-off. These results suggest that lateral trunk flexion moves the BCoM laterally towards the leading stance limb during double support phase and facilitates weight acceptance, while simultaneously unloading the trailing limb in preparation for swing. Due to the nature of passive transtibial prostheses, BTAs suffer from a lack of active ankle plantarflexion motion and limited power generation for advancing the limb forward during terminal stance [2, 21]. Thus, lateral trunk flexion utilization by BTAs could ultimately be a mechanism for aiding in the transfer of body weight from the trailing to the leading leg during double support and providing adequate foot clearance for swing. If indeed the case, this may explain why lateral trunk flexion remained invariant to both walking speed and prosthetic configuration. Consequently, prosthetic components that provide active power generation during terminal stance may be useful for reducing lateral sway in these individuals; further studies would be useful to explore this hypothesis.

Importantly, given that approximately 68% of the total body mass is concentrated in head, arms, and trunk [28], subtle movements in the trunk could have substantial effects on the BCoM position and its relation to the BoS [29]. Indeed, a characteristic example of these effects can be seen in Figure 4, in which maintaining large lateral trunk flexion ROM in combination with an increase in walking speed and reduced step width pushes the BCoM projection further toward the lateral edge of the BoS. The XCoM is displaced beyond this edge following initial contact, a characteristic not observed in able-bodied gait (Figure 4E), which places these individuals at an increased fall risk if subjected to an external lateral perturbation [8, 9]. The exaggerated trunk lean seems responsible for this unstable scenario and also creates a slight veering of the BCoM despite the subjects being requested to walk in a straight line. In able-bodied individuals, ankle musculature may be used to produce minor adjustments to the plantar CoP following initial contact if the XCoM strays too close to the CoP and hence, lateral BoS [9]. However, considering that BTAs are lacking ankle musculature, these individuals would then be reliant on adjustments produced at the hip joints to successfully recover, possibly requiring a risky crossover step depending on the perturbation magnitude. Furthermore, BTAs are at a greater deficit compared to their unilateral counterparts since they lack compensatory mechanisms that could be produced by the sound limb. Therefore, whatever mechanical advantage that this lateral trunk flexion may provide during walking could be offset by an increase in fall risk. Theoretically, any attempt to reduce this exaggerated trunk motion may be beneficial for helping maintain upright balance in BTAs.

Study limitations include the mixed subject population with regards to amputation etiology and age. A previous investigation on BTA gait identified differences in parameters between individuals who had amputation due to trauma and peripheral vascular disease [30]. Furthermore, the range of subject age was 23-67 and this may have had some, albeit minimal [5], impact on between-subject kinematics. Due to the nature of this study, selection of subjects was limited to those who satisfied specific inclusion criteria and that should be considered when interpreting between-group results. However, despite the variation in amputation etiology and age, the dynamic behaviors analyzed in this study were consistent at some degree across all subjects and will have negligible effect on the repeated-measures statistical results. Custom amputee models were not used as prosthetic mass data were not available, but inclusion of these data would likely not substantially improve accuracy of the BCoM calculation given the distal amputations (approximately 88% of total body mass is comprised by segments superior to the knee [28]). Additionally, instantaneous CoP position of the foot was not actually measured and may therefore not follow the idealized trajectory assumed for this study. Consequently, the XCoM results may differ slightly from those reported. Finally, the order of prosthetic configuration testing was not randomized as subjects were allowed to first accommodate to the test prosthetic feet without adaptors as this represented a baseline condition. Therefore, this study may suffer from task order effects and this should be considered when interpreting for clinical significance.

In conclusion, BTAs display significantly greater lateral trunk flexion ROM that seems invariant to changes in step width. The results suggest that this lateral trunk motion is useful for shifting the BCOM laterally and unloading the trailing limb. However, as BTAs reduce their step width, the exaggerated lateral sway may place them in an unstable scenario in which the XCoM is displaced beyond the lateral BoS. Importantly, as inclusion of prosthetic adaptors for increasing passive ankle ROM slightly reduced step width, these types of components may further challenge upright balance, particularly at faster speeds. Therefore, it may be advantageous for the patient's safety if clinicians were to focus on minimizing this lateral sway. Further studies are needed to identify if either particular prosthetic components that assist limb advancement and/or gait training may be effective in reducing this lateral sway while still maintaining efficient ambulation.

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Research Highlights

1. We observed upper body kinematics of bilateral transtibial amputee gait.
2. Subjects walked with 3 speeds and 2 prostheses of varying range-of-motion.
3. Amputees walk with greater lateral trunk flexion range-of-motion than controls.
4. Exaggerated lateral trunk sway may expose amputees to reduced walking stability.

Figure 1.

Average peak shoulder abduction (± 1 standard deviation) for able-bodied controls and amputee subjects during both prosthetic configurations PC1 and PC2. Significant differences between walking condition are denoted by asterisks (* $p=0.004$); Significant differences between amputee and able-bodied data are denoted by crosses († $p<0.001$, ‡ $p<0.001$).

Figure 2.

Lateral trunk flexion, trunk forward lean and trunk rotation profiles (ensemble average; ± 1 standard deviation represented by shaded area) during a complete gait cycle (i.e., initial contact to next initial contact of the right foot) for able-bodied controls and amputee subjects during use of PC1 when walking at three speeds (A, C, E). Average lateral trunk flexion ROM, trunk forward lean, and trunk rotation ROM (± 1 standard deviation) for able-bodied controls and amputee subjects during use of PC1 and PC2 (B, D, F). Significant differences between walking condition are denoted by asterisks (trunk forward lean: * $p=0.018$, ** $p=0.003$; trunk rotation: * $p=0.02$). Significant differences between amputee and able-

bodied data are denoted by crosses (lateral trunk flexion: †p=0.001, ‡p=0.002; trunk forward lean: †p<0.035, ‡p=0.005; trunk rotation: †p=0.048).

Figure 3.

Average step width (± 1 standard deviation) for able-bodied controls and amputee subjects during use of PC1 and PC2. Significant differences between walking condition are denoted by asterisks (* $p=0.049$). Significant differences between amputee and able-bodied data are denoted by crosses († $p<0.001$, ‡ $p=0.002$).

Figure 4.

Exemplary temporal profile of BCoM and XCoM lateral position (first column) and lateral trunk flexion (second column) during walking at the slow speed (0.6 m/s) and using PC1 (A and B) and walking at the fast speed (1.1 m/s) using PC2 (C and D), and for an age-matched able-bodied control (E and F) walking at 1.4 m/s. Positive lateral trunk flexion is representative of left shoulder rise, or leaning to the right. The black dashed lines connect position of the heel marker and toe marker at initial contact and toe-off of the same foot, respectively, to represent an idealized CoP trajectory that passes under ankle joint center (A, C, and E). Corresponding lateral trunk flexion at times of initial contact and toe-off have been denoted (B, D, and F). Step width for the first amputee condition (row 1), second

amputee condition (row 2), and control condition (row 3) was 28.1, 22.1, and 17.0 cm, respectively. The reduction in step width for this amputee subject (6 cm) represents 27% of the step width of the first condition. RIC = right initial contact; LIC = left initial contact; RTO = right toe-off; LTO = left toe-off.

Figure 5.

Average MLGRF (± 1 standard deviation) for able-bodied controls and amputee subjects during use of both prosthetic configurations PC1 and PC2 as (a) normalized by subject weight and (b) normalized by weight and step width. Significant differences between walking condition are denoted by asterisks ((a) * $p=0.029$, ** $p=0.016$; (b) * $p=0.019$, ** $p=0.001$)).