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Dynamic Stability of Superior vs. Inferior Body Segments in Individuals with Transtibial Amputation Walking in Destabilizing Environments☆

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Abstract

Interestingly, young and highly active people with lower limb amputation appear to maintain a similar trunk and upper body stability during walking as able bodied individuals. Understanding the mechanisms underlying how this stability is achieved after lower leg amputation is important to improve training regimens for improving walking function in these patients. This study quantified how superior (i.e., head, trunk, and pelvis) and inferior (i.e., thigh, shank, and feet) segments of the body respond to continuous visual or mechanical perturbations during walking. Nine persons with transtibial amputation (TTA) and 12 able bodied controls (AB) walked on a 2m × 3m treadmill in a Computer Assisted Rehabilitation Environment (CAREN). Subjects were perturbed by continuous pseudo random mediolateral movements of either the treadmill platform or the visual scene. TTA maintained a similar local and orbital stability in their superior body segments as AB throughout both perturbation types. However, for their inferior body segments, TTA subjects exhibited greater dynamic instability during perturbed walking. In TTA subjects, these increases in instability were even more pronounced in their prosthetic limb compared to their intact leg. These findings demonstrate that persons with unilateral lower leg amputation maintain upper body stability in spite of increased dynamic instability in their impaired lower leg. Thus, transtibial amputation does significantly impair sensorimotor function, leading to substantially altered dynamic movements of their lower limb segments. However, otherwise relatively healthy

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Authors' contributions
RB was involved in the analysis and interpretation of the data, as well as drafting the manuscript. JD and JW participated in the design of the study and critically revised the manuscript for its intellectual content. All authors have read and approved the final manuscript.

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patients with unilateral transtibial amputation appear to retain sufficient remaining sensorimotor function in their proximal and contralateral limbs to adequately compensate for their impairment.

Introduction

Individuals with lower limb amputation experience significant changes in walking capability and greatly increased risk of falling (Miller et al., 2001; Grumillier et al., 2008). Most falls occur during whole body movements like walking (Tinetti et al., 1995). During everyday activities, individuals need to respond to various gait disturbances (e.g., uneven/slippery terrain, crowded spaces, etc.) to avoid stumbles and falls. To maintain walking stability and improve gait function in patients with amputation, measures related to maintaining stability and/or balance while walking (Dingwell and Cusumano, 2000) and the individual strategies used to achieve gait stability need to be quantified.

In destabilizing environments, able bodied individuals alter how they walk. Hip and knee flexion increases to lower center of mass, and ankle dorsi flexion and hip and knee flexion increase during swing phase to increase toe clearance (Gates et al., 2012). When exposed to side to side perturbations of the visual field or walking surface, healthy young adults exhibited greater dynamic instability (McAndrew et al., 2011; Sinitski et al., 2012) and increased stepping and trunk movement variability (McAndrew et al., 2010). These became more pronounced with larger perturbation amplitudes (Sinitksi et al., 2012; Terry et al., 2012). In some studies, individuals with amputations similarly adopted a slower more conservative gait on challenging surfaces (Paysant et al., 2006; Kendell et al., 2010; Lamoth et al., 2010). In other studies, however, persons with transtibial amputation exhibited similar stepping parameters as able bodied subjects when walking on irregular surfaces (Curtze et al., 2011) and were nearly as efficient in recovering from evoked falls (Curtze et al., 2010).

Interestingly, when young highly active individuals with and without transtibial amputation were subjected to mediolateral perturbations, subjects in both groups took shorter and wider steps (Beurskens et al., 2014), thus increasing their bases of support (Beltran et al., 2014). Likewise, all subjects exhibited greatly increased stepping variability and trunk kinematic variability when perturbed (Beurskens et al., 2014). When subjected to side to side mechanical (walking platform) perturbations, subjects with amputation exhibited greater variability of step widths and trunk (C7) kinematics than controls (Beurskens et al., 2014). These subjects with amputation also exhibited larger mean lateral margins of stability (Beltran et al., 2014). Larger mean stability margins might suggest these subjects were more stable (Hof et al., 2005; Hof et al., 2007) when perturbed. However, these patients also exhibited greater step to step *variability* in their margins of stability, which could indicate they still exhibited an increased likelihood of experiencing steps with very small or possibly negative margins of stability (Beltran et al., 2014). However, these patients did not exhibit increased local or orbital instability of their trunk (C7) movements compared to healthy controls (Beurskens et al., 2014). Two recent studies similarly found that persons with transtibial amputation exhibited either only slightly increased local instability of pelvic movements during perturbed walking (Hak et al., 2013) or slightly *less* dynamic instability of center of mass (sacrum) movements during normal running (Look et al., 2013). Thus,

active persons with lower limb amputation appear to be able to achieve similar trunk stability as able bodied individuals. However, the strategies used to maintain this trunk stability remain unknown.

The trunk comprises over 50% of total body mass and significantly affects whole body dynamics (Winter, 2009). Thus, one primary aim of walking is to stabilize these trunk movements. Indeed, the movements of inferior segments (e.g. legs, feet), acting through the complex coupling dynamics between the upper and lower body segments, likely contribute to stabilizing the motions of superior segments (e.g. pelvis, trunk) (Prince et al., 1994; Cromwell et al., 2004), and thus also movements of the whole body center of mass. Young adults exhibited greater local dynamic instability in their shanks and feet compared to their pelvis and trunk during unperturbed walking (Kang and Dingwell, 2009). Patients with transtibial amputation exhibited increased local instability of lower extremity joint kinematics compared to able bodied persons (Wurdeman et al., 2013), but that study did not measure upper body motions. Although another study reported increased trunk instability in patients with transfemoral amputation, counterintuitively, those subjects also became *more* stable when they walked outdoors over irregular terrain (Lamoth et al., 2010). However, one study reported simultaneously *greater* local instability of leg movements accompanied by *lower* instability of center of mass (sacrum) movements in patients with transtibial amputation during running (Look et al., 2013). This finding most directly supports the hypothesis inferior segment movements do appear to help stabilize superior segment movements (Cromwell et al., 2004; Kang and Dingwell, 2009). It is not known whether individuals with lower limb amputation exhibit similar differences between trunk and lower limb dynamic stability during either unperturbed or, most importantly, perturbed walking.

Here, we determined the dynamic stability of superior (i.e. head, trunk, and pelvis) and inferior (i.e. thighs, shanks, and feet) body segments in individuals with and without transtibial amputation. We hypothesized (a) that dynamic instability would increase across all body segments during perturbed walking (McAndrew et al., 2011), (b) that in individuals with transtibial amputation, inferior segments (shanks, thighs, feet) would exhibit greater increases in local and orbital dynamic instability for perturbed compared to unperturbed walking, especially in their impaired limb (Look et al., 2013), and (c) that inferior segments would exhibit greater local and orbital dynamic instability than superior segments (Kang and Dingwell, 2009).

Methods

Nine individuals with traumatic unilateral transtibial amputation (TTA) and twelve able bodied controls (AB) participated (Table 1). All TTA were able to walk without help or assistive devices and were concurrently enrolled in the rehabilitation program at the Center for the Intrepid at Brooke Army Medical Center. Before participating, all participants signed informed consent forms approved by the Institutional Review Boards at both Brooke Army Medical Center and The University of Texas.

The experimental protocol is described in detail elsewhere (Beltran et al., 2014; Beurskens et al., 2014). Briefly, all subjects walked on a 2m x 3m treadmill in a Computer Assisted

Rehabilitation Environment (CAREN) virtual reality system (Motek, Amsterdam, Netherlands) (Fig. 1). Each participant walked at a constant speed *(v)*, non dimensionally scaled to their leg length (*l*) using (Vaughan and O'Malley, 2005):

$$\nu = \sqrt{0.16 \cdot g \cdot l}, \quad (1)$$

where leg length (*l*) was defined as distance in meters from the walking surface to the great trochanter. This yielded an average walking speed of approximately 1.22 ± 0.28 m/s across all subjects.

Following a 6 minute acclimation period, each participant completed five 3 minute walking trials each in each of the following conditions (Beurskens et al., 2014):

- ***NOP:*** No perturbations and normal visual optic flow with the visual scene movement matched to the treadmill belt speed.
- ***PLAT:*** Perturbations imposed as continuous mediolateral (ML) translations of the treadmill platform, while participants walked with normal visual optic flow.
- ***VIS:*** Perturbations imposed as continuous ML translations of the visual display, while participants walked on a stationary treadmill platform.

PLAT and VIS perturbations were both generated as a pseudo random sum of 4 sine waves (Eq. 2) with incommensurate frequencies (McAndrew et al., 2010; Beurskens et al., 2014):

$$A(t) = A_W \left[1.0 sin(0.16\text{x}2\pi t) + 0.8 sin(0.21x2\pi t) + 1.4 sin(0.24\text{x}2\pi t) + 0.5 sin(0.49\text{x}2\pi t)\right], \quad (2)$$

where *A(t)* was the perturbation amplitude in meters, A_w was a weighting factor in meters and *t* was time in seconds. Perturbation magnitudes were set to A_w=0.05 for PLAT and to A_w=0.50 for VIS to generate approximately comparable responses for each perturbation type (Terry et al., 2012). Throughout each trial, participants' head orientation was monitored visually to ensure they were looking at the horizon directly in front of them. The order of conditions was randomized for each individual. Rest breaks were provided between conditions or at participants' request.

Kinematic data of the head, trunk, pelvis, thighs, shanks and feet were collected at 60Hz using a 24 camera motion capture system (Vicon Motion Systems, Oxford, UK) and 55 reflective markers. Marker positions were combined with digitized joint centers and then post processed using Vicon Nexus and Visual3D software (C Motion Inc., Germantown, MD) to define a 13 segment model (Wilken et al., 2012). This model was used to calculate each segment's center of mass (COM). The mediolateral (ML) and anterior posterior (AP) trajectories of these segmental COM movements were then analyzed.

To determine heel strikes, we used a velocity based detection algorithm (Zeni et al., 2008). The first 150 strides for each trial were used for subsequent analyses. For platform perturbation trials, however, the recorded data included both subject and platform movements. Therefore, the platform displacements were first subtracted from each segments' COM displacements prior to calculating COM velocities.

To compute dynamic stability measures, delay embedded state spaces (Gates and Dingwell, 2009) were constructed for both the ML and the AP velocities of each segment's COM using each original time series data and their time delayed copies (Dingwell and Cusumano, 2000; Dingwell and Marin, 2006):

$$S(t) = [v(t), v(t+\tau), \ldots, v(t+(d_E - 1)\tau)], \quad (3)$$

where $S(t)$ is the d_E dimensional state vector, $v(t)$ is the original 1 dimensional data, is the time delay and d_E is the embedding dimension. Time delays were determined from the first minimum of the Average Mutual Information function (Fraser, 1986), yielding an average time lag of 20 frames. An embedding dimension of $d_E = 5$ was used for all trials (Dingwell and Cusumano, 2000).

Dynamic stability of these segmental COM state spaces was quantified in two ways, by calculating both local divergence exponents (λ^*_s) and maximum Floquet multipliers (MaxFM). To compute λ^*_s, the 150 continuous strides of data were each first re sampled to 15,000 total data points (average of 100 data points per stride) prior to state space construction (England and Granata, 2007; Bruijn et al., 2009). Short term local divergence exponents, λ^*_s, were then computed as the average exponential rates of divergence in each state space over a normalized time frame of 0 1 strides, following established methods (Dingwell and Cusumano, 2000; Kang and Dingwell, 2008). Positive exponents indicate that small (i.e., local) perturbations grow over time signifying a *locally* unstable system. Larger positive exponents indicate greater local instability (Dingwell and Cusumano, 2000; Dingwell and Marin, 2006).

To calculate MaxFM, state space data for each trial were first segmented into individual strides and the data for each stride were then normalized to 0 100% of the gait cycle (Dingwell and Kang, 2007). A Poincaré map was defined at each percent of the gait cycle to quantify how small perturbations from the average trajectory increased or decreased across successive strides (Donelan et al., 2004; Kang and Dingwell, 2008). The magnitude of MaxFM for each Poincaré section was calculated and averaged over the gait cycle. MaxFM < 1 indicates that the system is orbitally stable. If MaxFM increases, but remains < 1, the system is less orbitally stable (Kang and Dingwell, 2008).

For each segment (*head, trunk, pelvis, thighs, shanks, and feet*) and each dependent measure (MaxFM and λ^*_s), 2 factor (Group × Condition) repeated measures ANOVAs were used to determine differences between both groups (*AB* vs. *TTA*) across conditions, separately for platform (*NOP vs. PLAT*) and visual (*NOP vs. VIS*) perturbation trials. We then conducted a second set of analyses adding a "Leg" factor to our ANOVA to determine how participants' limbs (i.e., right vs. left for AB and healthy vs. prosthetic for TTA) contributed to our findings. All statistical analyses were performed using SPSS 19 (SPSS Inc., Chicago, IL).

Results

For AP segmental COM movements (Fig. 2A; Table 2A), local instability (λ^*_s) increased considerably from the head to the feet, with AP movements of the foot segment being locally most unstable (Fig. 2A). Differences between body segments were statistically

significant, as the means for the shank and feet segments were well above the 95% CI upper bounds for the head, trunk and pelvis segments. When comparing NOP to PLAT, there were no significant differences between subject groups (AB vs. TTA) for these movements (Table 2A). However, when comparing NOP to VIS, the trunk and pelvis segments of AB subjects were slightly more unstable than those of TTA.

For ML COM movements (Fig. 2B; Table 2B), λ^*_s decreased from the head to the thigh segments, but increased from the thigh to the feet segments. Mean values for head and trunk were above the 95% CI upper bounds for all inferior segments. Likewise, mean values for shank and feet were mostly above the 95% CI upper bounds for the thigh, particularly for TTA subjects. The only significant group difference (between TTA and AB) occurred at the shank segment, where TTA exhibited greater local instability than AB during both PLAT and VIS perturbations. Additionally, for ML head, trunk, pelvis and thigh segment movements, AB exhibited greater increases in λ^*_s than did TTA during visually perturbed walking (Table 2B).

For AP segmental COM movements (Fig. 3A; Table 3A), orbital stability (MaxFM) remained relatively constant across body segments. Although quite small, differences in MaxFM between NOP and PLAT (Fig. 3A) for all segments except the pelvis, and differences between NOP and VIS for the feet (Fig. 3A) were statistically significant (Table 3A). Most importantly, however, there were no statistically significant Group (AB vs. TTA) differences or Group × Condition interaction effects (Table 3A).

For ML segmental COM movements (Fig. 3B; Table 3B), MaxFM decreased somewhat from head to the feet segments, across Groups and Conditions. Differences between body segments were statistically significant, as the means for the shank and feet were below the 95% CI lower bounds for the head and trunk (Fig. 3B). During both PLAT and VIS perturbations, MaxFM increased significantly in all body segments (Fig. 3B; Table 3B). The only significant Group (TTA vs. AB) difference occurred for ML foot movements during PLAT perturbations. There, the feet segments of AB were slightly more unstable than those of TTA.

When comparing between legs (Fig. 4; Table 4), for λ^*_s, the main effects of Condition (Table 4A) remained significant, just as before (Table 2A). However, several significant differences between right and left (for AB) and between prosthetic and intact (for TTA) legs were also apparent (Table 4A). Most notably, TTA subjects exhibited greater local instability on their prosthetic side compared to their unimpaired leg and compared to AB. The largest of these differences occurred in the shank and feet segments (Fig. 4A). This increase in local instability also depended on both perturbation (NOP vs. PLAT and NOP vs. VIS) and Group (TTA vs. AB), as indicated by significant Leg × Condition and Group × Leg × Condition interactions (Table 4A).

For MaxFM (Fig. 4B), the only statistically significant differences between limbs occurred for the shank and feet segments when comparing the NOP to VIS perturbation conditions (Table 4A). However, these differences between limbs were generally quite small (Fig. 4B),

and depended on both Condition and Group, as indicated by the significant Leg × Condition and Group × Leg × Condition interactions (Table 4B).

Discussion

Individuals with lower leg amputations experience significant motor and sensory deficits that result in increased risk of falling (Miller et al., 2001; Grumillier et al., 2008). At the same time, their trunk kinematics and stability appear similar to that of able bodied adults (Look et al., 2013; Beltran et al., 2014; Beurskens et al., 2014). Understanding how humans remain stable during locomotion is important to reveal insights into how destabilizing environments affect persons with and without lower limb amputation. The present study determined how persons with transtibial amputation and able bodied adults respond to different perturbations known to substantially destabilize human walking (McAndrew et al., 2011) and how these perturbations affect superior and inferior segments of the body (Kang and Dingwell, 2009).

Our first hypothesis, that all subjects would be strongly destabilized by the imposed perturbations, was very strongly supported. All subjects exhibited large increases in both λ^*_s (Fig. 2; Table 2) and MaxFM (Fig. 3; Table 3) during both PLAT and VIS perturbations for nearly all body segments. Our second hypothesis, that TTA subjects would exhibit greater increases in dynamic instability than AB controls, particularly for movements of their impaired limbs, was also mostly supported. TTA subjects exhibited greater λ^*_s for anterior posterior (AP) trunk and pelvis motions (Fig. 2A; Table 2A) and mediolateral (ML) shank motions (Fig. 2B; Table 2B). However, TTA also exhibited consistently (Table 3B), but only very slightly (Fig. 3B) decreased MaxFM for ML feet motions compared to AB controls. Most importantly, the greatest differences were seen in the AP λ^*_s for the prosthetic vs. intact legs of TTA subjects (Fig 4A).

Our third hypothesis, that inferior segments would exhibit greater dynamic instability than superior segments was also mostly supported. For segmental movements in the AP direction, superior segments (i.e., head, trunk and pelvis) were locally (λ^*_s) most stable compared to inferior segments (i.e. shank and feet) (Fig. 2A). The head and pelvis segments were most stable, while other segments exhibited increased local instability with the greatest instability occurring at the feet (Fig 2A). Local dynamic instability generally increased from the head to the feet, supporting previous work in both healthy subjects (Kang and Dingwell, 2009) and patients with amputation (Look et al., 2013). Conversely, MaxFM of mediolateral movements decreased slightly from the head to the feet (Fig. 3B), also consistent with previous findings (Kang and Dingwell, 2009). Overall, these findings suggest that, compared to inferior segments, the motions of superior segments were mostly less sensitive to small perturbations.

One possible explanation for these findings is that the nervous system prioritizes maintaining stability of the upper body over that of inferior segments (Prince et al., 1994; Cromwell et al., 2004; Winter, 2009). On the other hand, it could also be that there is just a greater range of AP movements in the lower legs (i.e., feet) compared to superior segments (i.e., trunk). Larger amplitude movements would likely induce larger amplitude divergence

and thus potentially larger divergence rates also. While such a trend was observed for λ^*_s (Fig. 2A), this did not occur for MaxFM (Fig. 3A). However, this idea was supported by the finding that in the ML direction, both local (Fig. 2B) and orbital (Fig. 3B) instability actually decreased slightly from the head to the feet. These differences in dynamic stability between AP and ML movements were not observed in previous studies that either pooled ML and AP movements into the same state spaces (Kang and Dingwell, 2009; Hak et al., 2013) or that measured only sagittal plane joint kinematics (Look et al., 2013; Wurdeman et al., 2013) or only trunk movements (Lamoth et al., 2010). Orbital stability in both movement directions (AP vs. ML) remained more similar across all body segments (Fig. 3) than did local stability (Fig. 2). Thus, the greater range of movements in AP direction at the inferior body segments affected the local dynamic stability of ML movements and the orbital stability of both AP and ML movements much less.

The perturbations applied here clearly affected subjects' dynamic stability in the way they were intended to, and similar to our previous work (McAndrew et al., 2011; Beurskens et al., 2014). When subjected to ML perturbations, individuals both with and without transtibial amputation exhibited dramatic and significant increases in both local (λ^*_s; Fig. 2B, Table 2) and orbital (MaxFM; Fig. 3B, Table 3) dynamic instability of the ML COM movements of all body segments. This indicates that visual and walking surface perturbations significantly affect whole body stability, as previously shown (McAndrew et al., 2011; Sinitski et al., 2012; Terry et al., 2012). Importantly, the present findings extend these prior results to other body segments. Similar patterns were observed for orbital stability (Fig. 4B; Table 4). However, differences in MaxFM measures were much smaller, supporting previous work showing that λ^*_s is more sensitive than MaxFM to capture changes in these dynamic stability properties (Dingwell and Kang, 2007; McAndrew et al., 2011).

During visual and platform perturbations, individuals with transtibial amputation exhibited significantly greater local instabilities for ML shank movements compared to controls (Fig. 2B; Table 2B). Although able bodied controls exhibited greater local instability for AP movements during visual perturbations for their trunk, pelvis and thigh segments (Fig. 2A; Table 2A), these changes were rather small and did not exceed the thresholds (in % of mean) for a minimal detectable change (i.e., ~7.4% for λ^*_s during visual perturbations; (Rabago et al., 2013)). Differences between groups tested here were all < 7%. Thus, the most significant group difference that remains is for local instability of ML movements of the shank segment. For orbital dynamic stability, only the mediolateral movements of the trunk and foot segments exhibited significant impairment related differences (Fig. 3B; Table 3B). In both cases, the AB subjects were slightly more stable than the TTA subjects (Fig. 3B). These minimal between group differences in head, trunk, and pelvis COM dynamic stability are consistent with several papers reporting either no differences, or only very small differences, in movements of the C7 (Beurskens et al., 2014), the trunk (Lamoth et al., 2010), or the pelvis (Hak et al., 2013; Look et al., 2013). Thus, many different points on these superior upper body segments appear to exhibit similarly stable movements.

In general, able bodied individuals did not show differences in orbital or local dynamic stability between their right and left legs in any of the tested conditions or for any segment

(Fig. 4). Conversely, persons with amputations exhibited significantly greater local dynamic instability in the shank and foot segments of their prosthetic leg compared to their intact leg during both visual and platform perturbations and compared to unperturbed walking (Fig. 4). These findings are consistent with other recent reports of similar between limb differences observed during both walking (Wurdeman et al., 2013) and running (Look et al., 2013).

Taken together, these findings confirm previous work suggesting that patients with amputation adapt the movements of their impaired lower extremity segments (Cromwell et al., 2004; Kang and Dingwell, 2009; Winter, 2009; Look et al., 2013) to maintain sufficient dynamic stability of their upper body movements (Prince et al., 1994; McAndrew et al., 2011; Beurskens et al., 2014). Individuals with amputation develop an asymmetrical gait pattern spending more time standing on their intact limb and less on their impaired (Engsberg et al., 1993; Nolan et al., 2003). The present findings demonstrate that these individuals also exhibit more pronounced local instability of the movements of their prosthetic leg (Fig. 4). The present study was not designed to identify the specific individual neurophysiological and/or biomechanical adaptations that these patients make to achieve this outcome. However, the results of this study do provide a strong foundation to suggest that future work should focus on trying to determine how it is these patients are altering their lower extremity movements and stepping movements both within each stride, and in particular from each stride to the next (e.g., as in (Dingwell et al., 2010)), to achieve the overall upper body stability that they are. These findings further suggest that rehabilitation interventions designed to manipulate lower extremity movement patterns in specific ways intended to achieve a desired upper body movement pattern are likely to have promising outcomes.

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References

Bardy BG, Warren WH, Kay BA. Motion parallax is used to control postural sway during walking. Experimental Brain Research. 1996; 111(2):271–282. [PubMed: 8891657]

Beltran EJ, Dingwell JB, Wilken JM. Margins of stability in young adults with traumatic transtibial amputation walking in destabilizing environments. Journal of Biomechanics. 2014; 47(5):1138–1143. [PubMed: 24444777]

Beurskens R, Wilken JM, Dingwell JB. Dynamic stability of individuals with transtibial amputation walking in destabilizing environments. Journal of Biomechanics. 2014; 47(7):1675–1681. [PubMed: 24679710]

Bruijn SM, van Dieën JH, Meijer OG, Beek PJ. Is slow walking more stable? Journal of Biomechanics. 2009; 42(10):1506–1512. [PubMed: 19446294]

Cromwell R, Schurter J, Shelton S, Vora S. Head stabilization strategies in the sagittal plane during locomotor tasks. Physiotherapy Research International. 2004; 9(1):33–42. [PubMed: 15132026]

Curtze C, Hof AL, Otten B, Postema K. Balance recovery after an evoked forward fall in unilateral transtibial amputees. Gait & Posture. 2010; 32(3):336–341. [PubMed: 20609587]

Curtze C, Hof AL, Postema K, Otten B. Over rough and smooth: Amputee gait on an irregular surface. Gait & Posture. 2011; 33(2):292–6. [PubMed: 21176879]

Dingwell JB, Cusumano JP. Nonlinear time series analysis of normal and pathological human walking. Chaos: An Interdisciplinary Journal of Nonlinear Science. 2000; 10(4):848–863.

Dingwell JB, John J, Cusumano JP. Do humans optimally exploit redundancy to control step variability in walking? PLoS Computational Biology. 2010; 6(7):e1000856. [PubMed: 20657664]

Dingwell JB, Kang HG. Differences between local and orbital dynamic stability during human walking. Journal of Biomechanical Engineering. 2007; 129(4):586–593. [PubMed: 17655480]

Dingwell JB, Marin LC. Kinematic variability and local dynamic stability of upper body motions when walking at different speeds. Journal of Biomechanics. 2006; 39(3):444–452. [PubMed: 16389084]

Donelan JM, Shipman DW, Kram R, Kuo AD. Mechanical and metabolic requirements for active lateral stabilization in human walking. Journal of Biomechanics. 2004; 37(6):827–835. [PubMed: 15111070]

England SA, Granata KP. The influence of gait speed on local dynamic stability of walking. Gait & Posture. 2007; 25(2):172–178. [PubMed: 16621565]

Engsberg JR, Lee AG, Tedford KG, Harder JA. Normative ground reaction force data for able bodied and below knee amputee children during walking. Journal of Pediatric Orthopedics. 1993; 13(2): 169–73. [PubMed: 8459005]

Gates DH, Dingwell JB. Comparison of different state space definitions for local dynamic stability analyses. Journal of Biomechanics. 2009; 42(9):1345–1349. [PubMed: 19380140]

Gates DH, Wilken JM, Scott SJ, Sinitsky EH, Dingwell JB. Kinematic strategies for walking across a destabilizing rock surface. Gait & Posture. 2012; 35(1):36–42. [PubMed: 21890361]

Grumillier C, Martinet N, Paysant J, André JM, Beyaert C. Compensatory mechanism involving the hip joint of the intact limb during gait in unilateral trans tibial amputees. Journal of Biomechanics. 2008; 41(14):2926–2931. [PubMed: 18771768]

Hak L, van Dieën JH, van der Wurff P, Prins MR, Mert A, Beek PJ, Houdijk H. Walking in an unstable environment: Strategies used by transtibial amputees to prevent falling during gait. Archives of Physical Medicine and Rehabilitation. 2013; 94(11):2186–2193. [PubMed: 23916618]

Hof AL, Gazendam MGJ, Sinke WE. The condition for dynamic stability. Journal of Biomechanics. 2005; 38(1):1–8. [PubMed: 15519333]

Hof AL, van Bockel RM, Schoppen T, Postema K. Control of lateral balance in walking: Experimental findings in normal subjects and above knee amputees. Gait & Posture. 2007; 25(2):250–258. [PubMed: 16740390]

Kang HG, Dingwell JB. The effects of walking speed, strength and range of motion on gait stability in healthy older adults. Journal of Biomechanics. 2008; 41(14):2899–2905. [PubMed: 18790480]

Kang HG, Dingwell JB. Dynamic stability of superior vs. Inferior segments during walking in young and older adults. Gait & Posture. 2009; 30(2):260–263. [PubMed: 19502060]

Kendell C, Lemaire ED, Dudek NL, J. K. Indicators of dynamic stability in transtibial prosthesis users. Gait & Posture. 2010; 31:375–9. [PubMed: 20138523]

Lamoth CJC, Ainsworth E, Polomski W, Houdijk H. Variability and stability analysis of walking of transfemoral amputees. Medical Engineering & Physics. 2010; 32(9):1009–1014. [PubMed: 20685147]

Look N, Arellano CJ, Grabowski AM, McDermott WJ, Kram R, Bradley E. Dynamic stability of running: The effects of speed and leg amputations on the maximal lyapunov exponent. Chaos: An Interdisciplinary Journal of Nonlinear Science. 2013; 23(4)

McAndrew PM, Dingwell JB, Wilken JM. Walking variability during continuous pseudo random oscillations of the support surface and visual field. Journal of Biomechanics. 2010; 43(8):1470–1475. [PubMed: 20346453]

McAndrew PM, Wilken JM, Dingwell JB. Dynamic stability of human walking in visually and mechanically destabilizing environments. Journal of Biomechanics. 2011; 44(4):644–649. [PubMed: 21094944]

Miller WC, Speechley M, Deathe B. The prevalence and risk factors of falling and fear of falling among lower extremity amputees. Archives of Physical Medicine and Rehabilitation. 2001; 82(8): 1031–7. [PubMed: 11494181]

Nolan L, Wit A, Dudziñski K, Lees A, Lake M, Wychowañski M. Adjustments in gait symmetry with walking speed in trans femoral and trans tibial amputees. Gait & Posture. 2003; 17(2):142–151. [PubMed: 12633775]

Paysant J, Beyaert C, Datie AM, Martinet N, Andre JM. Influence of terrain on metabolic and temporal gait characteristics of unilateral transtibial amputees. Journal of Rehabilitation Research & Development. 2006; 43(2):153–60. [PubMed: 16847782]

Prince F, Winter DA, Stergiou P, Walt SE. Anticipatory control of upper body balance during human locomotion. Gait & Posture. 1994; 2:19–25.

Rabago CA, Dingwell JB, Wilken JM. Effect of perturbation modality selection on gait kinematic and stability metrics: A reliability study. Journal of NeuroEngineering and Rehabilitation. 2013 in revision.

Sinitski EH, Terry K, Wilken JM, Dingwell JB. Effects of perturbation magnitude on dynamic stability when walking in destabilizing environments. Journal of Biomechanics. 2012; 45(12):2084–2091. [PubMed: 22749389]

Terry K, Sinitski EH, Dingwell JB, Wilken JM. Amplitude effects of medio lateral mechanical and visual perturbations on gait. Journal of Biomechanics. 2012; 45(11):1979–1986. [PubMed: 22648147]

Tinetti ME, Doucette J, Claus E, Marottoli R. Risk factors for serious injury during falls by older persons in the community. Journal of the American Geriatrics Society. 1995; 43(11):1214–1221. [PubMed: 7594154]

Vaughan CL, O'Malley MJ. Froude and the contribution of naval architecture to our understanding of bipedal locomotion. Gait & Posture. 2005; 21(3):350–362. [PubMed: 15760752]

Wilken JM, Rodriguez KM, Brawner M, Darter BJ. Reliability and minimal detectible change values for gait kinematics and kinetics in healthy adults. Gait & Posture. 2012; 35(2):301–307. [PubMed: 22041096]

Winter, DA. Biomechanics and motor control of human movement. John Wiley & Sons, Inc; New York, NY: 2009.

Wurdeman SR, Myers SA, Stergiou N. Transtibial amputee joint motion has increased attractor divergence during walking compared to non amputee gait. Annals of Biomedical Engineering. 2013; 41(4):806–813. [PubMed: 23180032]

Zeni JA, Richards JG, Higginson JS. Two simple methods for determining gait events during treadmill and overground walking using kinematic data. Gait & Posture. 2008; 27(4):710–714. [PubMed: 17723303]

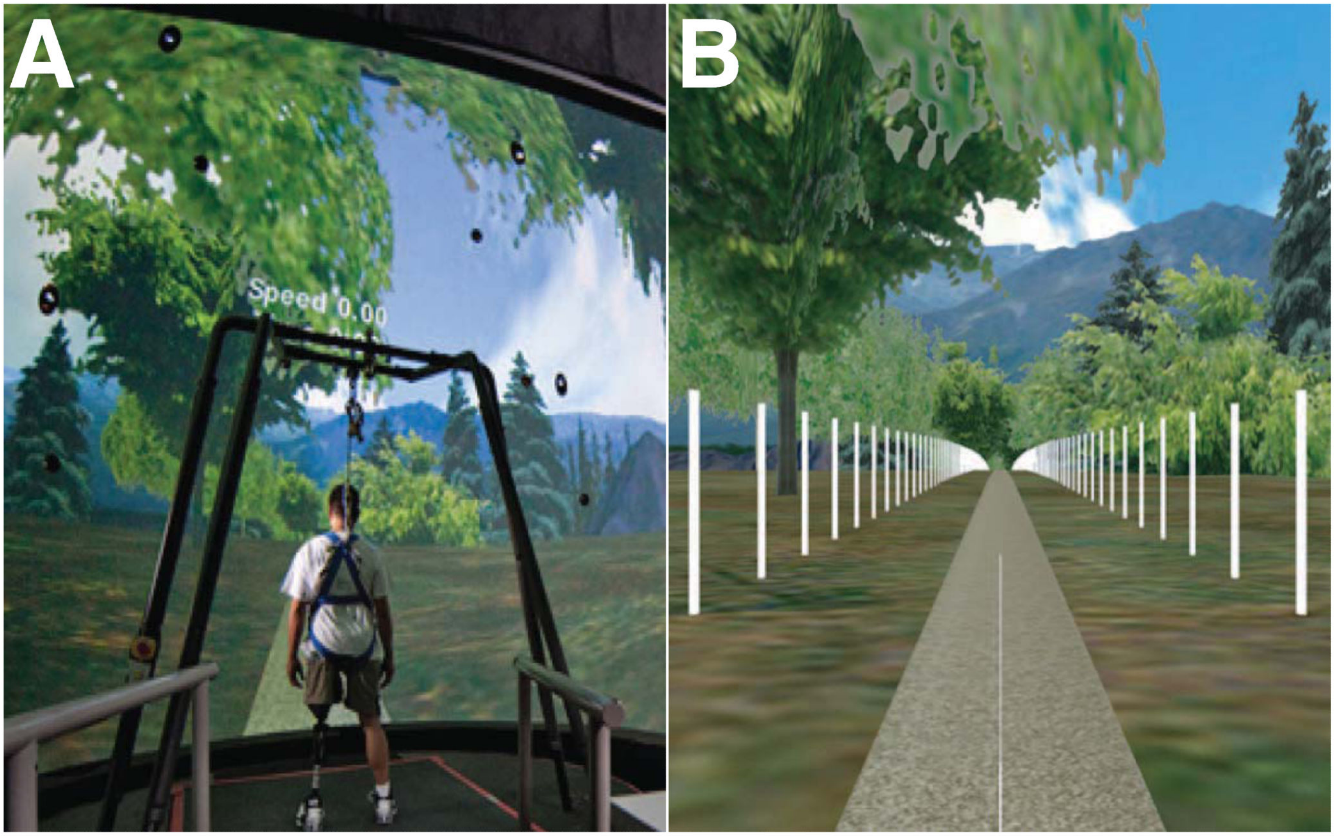

Figure 1. Experimental Setup

A: Example photo of a typical person with amputation standing inside the CAREN virtual reality system (Motek, Amsterdam, Netherlands). **B**: Portion of the visual scene used during CAREN trials, depicting a path through a forest with mountains in the background. Vertical white posts included on each side of the path were added to enhance motion parallax (Bardy et al., 1996; McAndrew et al., 2010).

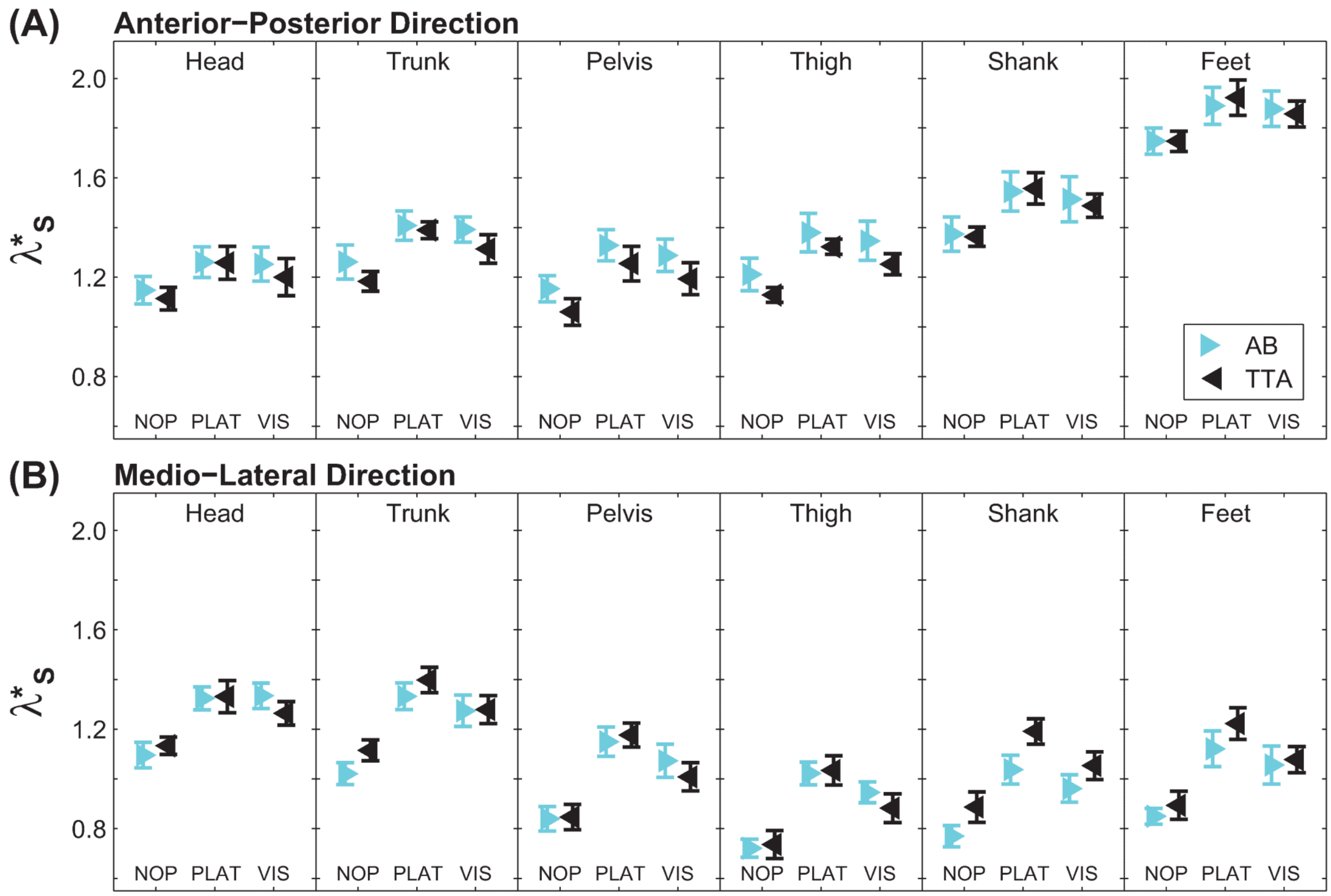

Figure 2. Local Dynamic Stability (λ^*_s) of Center of Mass Movements of Different Body Segments

A) Mean λ^*_s for segmental center of mass (COM) movements in the anterior posterior direction. **B**) Mean λ^*_s for segmental COM movements in the mediolateral direction. Both plots show data for able bodied (AB) subjects and individuals with amputation (TTA) walking under each condition (NOP, PLAT, and VIS). Error bars indicate the appropriate between subject ± 95% Confidence Intervals for each mean. Statistical outcomes for λ^*_s are presented in Table 2.

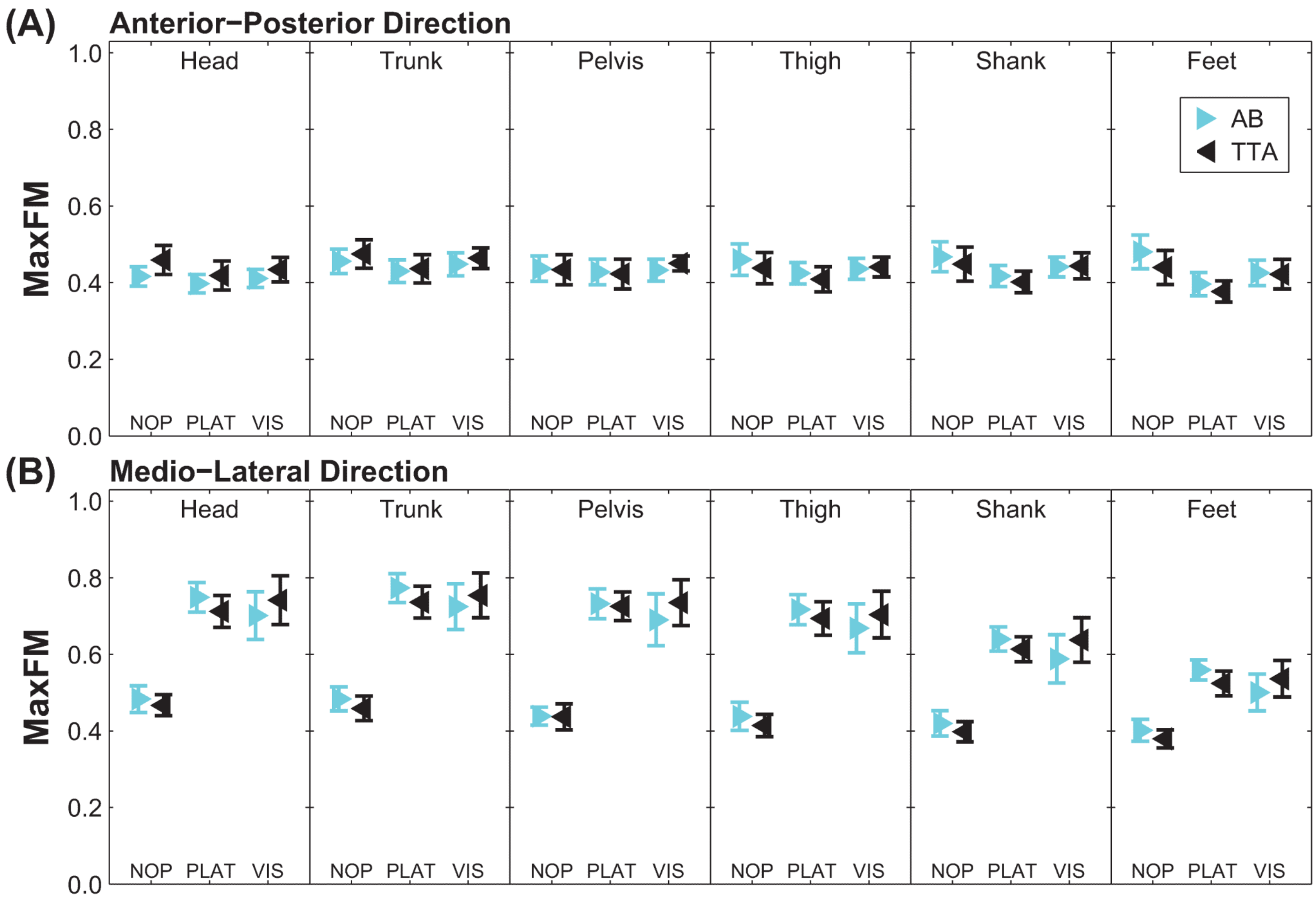

Figure 3. Orbital Dynamic Stability (MaxFM) of Center of Mass Movements of Different Body Segments

A) Mean MaxFM for segmental center of mass (COM) movements in the anterior posterior direction. **B**) Mean MaxFM for segmental COM movements in the mediolateral direction. Both plots show data for able bodied (AB) subjects and individuals with amputation (TTA) walking under each condition (NOP, PLAT, and VIS). Error bars indicate the appropriate between subject ± 95% Confidence Intervals for each mean. Statistical outcomes for MaxFM are presented in Table 3.

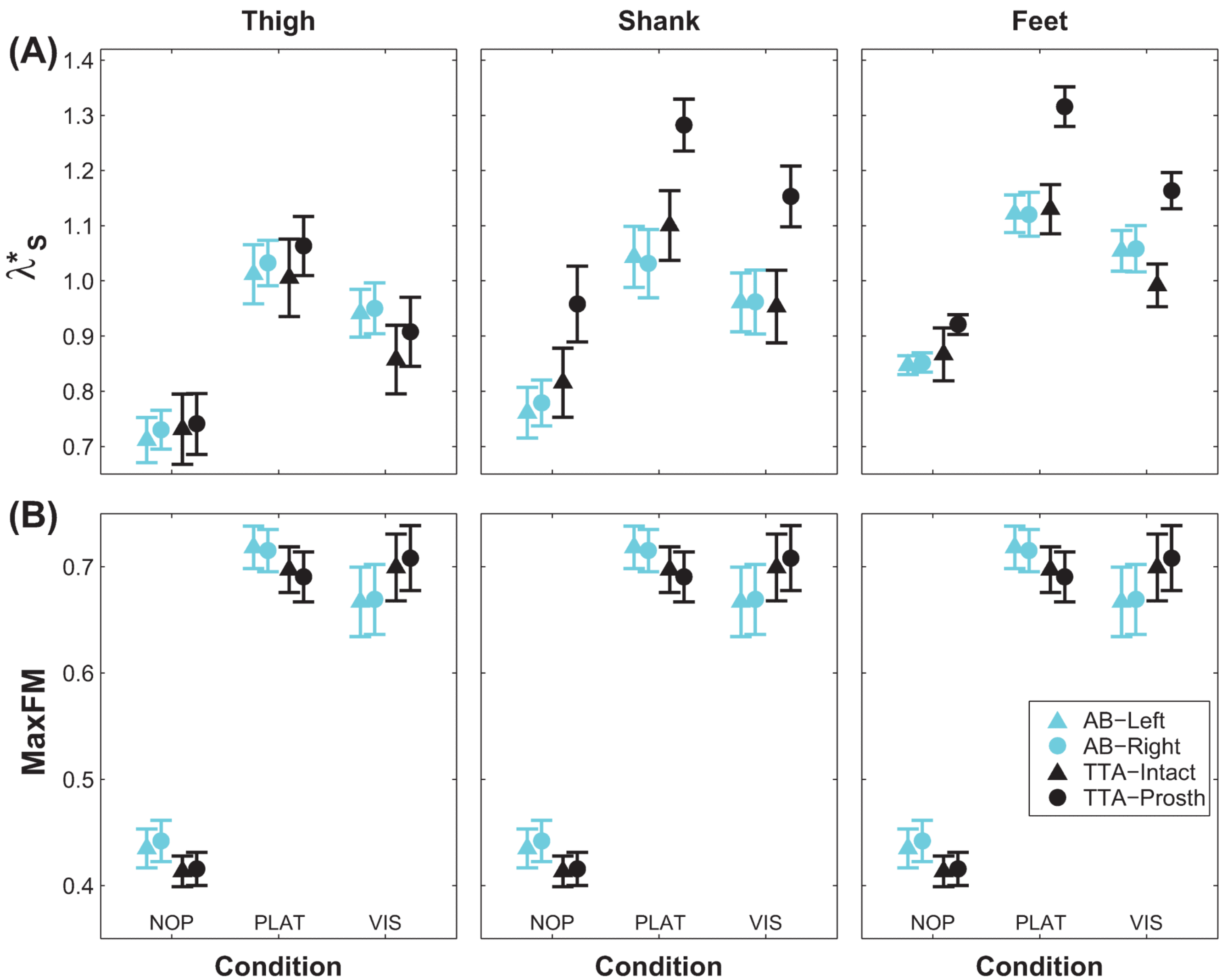

Figure 4. Local (λ^*_s) and Orbital (MaxFM) Dynamic Stability of Center of Mass Movements of Different Lower Limb Segments for Each Leg

A) Mean λ^*_s for mediolateral segmental center of mass (COM) movements. **B**) Mean MaxFM for mediolateral segmental center of mass (COM) movements. Both plots show data for left (light triangles) and right (light circles) legs of able bodied (AB) subjects and for the intact (dark triangles) and prosthetic (dark circles) legs of individuals with amputation (TTA) walking under each condition (NOP, PLAT, and VIS). Error bars indicate the appropriate between subject ± 95% Confidence Intervals for each mean. Statistical outcomes for λ^*_s and MaxFM separated by legs are presented in Table 4.

Table 1

Subject characteristics

	TTA (n = 9)	AB (n = 13)
Sex (M/F)[*]	9/0	10/3
Age (years)[*]	30.7 ± 6.75	24.8 ± 6.92
Height (cm)[*]	176.1 ± 0.11	174.8 ± 0.08
Body Mass (kg)[*]	90.2 ± 16.06	79.3 ± 11.56
BMI (kg/m^2)[*]	28.86 ± 2.26	26.0 ± 3.96
Leg Length (m)[*]	0.94 ± 0.07	0.95 ± 0.05
Cause of Amputation	8 Traumatic 1 Osteosarcoma	---
Time Since Amputation (mo)	19.8 ± 15.8	---
Residual Limb Length (%) [†]	55.1% ± 7.47%	---
Avg. Prosthetic Use (Hrs/Day)	13.94 ± 2.46	---
Pain Level (#) [†]	1.75 ± 1.28	---

[*] Note: t-tests for group differences: all $p < 0.05$

[†] Residual Limb Length is defined as a percent of length of the lower leg (from the knee to the ankle)

[‡] Pain ratings were taken on 10-pt visual analog scale (VAS). All subjects reported pain ≥ 3 at the time of testing.

Table 2

ANOVAs outcomes for λ^*_S

(A) Anterior-Posterior Direction (cf. Fig. 2A)

NOP vs. PLAT	***HEAD*** F (p-value)	***TRUNK*** F (p-value)	***PELVIS*** F (p-value)	***THIGH*** F (p-value)	***SHANK*** F (p-value)	***FEET*** F (p-value)
Group	0.19 (0.67)	1.63 (0.22)	4.19 (0.05)	2.92 (0.10)	0.00 (0.98)	0.15 (0.70)
Condition	**74.15 (0.00)**	**82.01 (0.00)**	**93.08 (0.00)**	**69.80 (0.00)**	**75.60 (0.00)**	**51.94 (0.00)**
Group × Condition	1.11 (0.31)	2.34 (0.14)	0.26 (0.62)	0.34 (0.57)	0.30 (0.59)	0.61 (0.44)
NOP vs. VIS						
Group	0.95 (0.34)	**4.66 (0.04)**	**5.28 (0.03)**	4.32 (0.05)	0.15 (0.71)	0.08 (0.78)
Condition	**33.08 (0.00)**	**31.22 (0.00)**	**53.93 (0.00)**	**36.88 (0.00)**	**39.03 (0.00)**	**34.29 (0.00)**
Group × Condition	0.31 (0.59)	0.00 (0.99)	0.00 (0.97)	0.08 (0.78)	0.15 (0.70)	0.23 (0.64)

(B) Medio-Lateral Direction (cf. Fig. 2B):

NOP vs. PLAT						
Group	0.56 (0.46)	7.54 (0.01)	0.21 (0.65)	0.19 (0.67)	**14.19 (0.00)**	3.72 (0.06)
Condition	**85.36 (0.00)**	**175.63 (0.00)**	**307.65 (0.00)**	**272.25 (0.00)**	**271.58 (0.00)**	**205.54 (0.00)**
Group × Condition	0.46 (0.51)	0.39 (0.54)	0.31 (0.58)	0.01 (0.93)	1.15 (0.29)	1.89 (0.19)
NOP vs. VIS						
Group	0.23 (0.64)	1.97 (0.18)	0.52 (0.49)	0.65 (0.43)	**9.23 (0.00)**	0.79 (0.38)
Condition	**157.47 (0.00)**	**119.89 (0.00)**	**137.61 (0.00)**	**111.76 (0.00)**	**79.18 (0.00)**	**71.11 (0.00)**
Group × Condition	**13.56 (0.00)**	**5.40 (0.03)**	**4.43 (0.04)**	**4.92 (0.03)**	0.38 (0.55)	0.24 (0.63)

Notes: Degrees of Freedom were F(1,19) for each ANOVA. Statistically significant effects ($p < 0.05$) are highlighted in bold text.

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Table 3

ANOVAs outcomes for MaxFM

(A) Anterior-Posterior Direction (cf. Fig. 3A)

NOP vs. PLAT	***HEAD***	***TRUNK***	***PELVIS***	***THIGH***	***SHANK***	***FEET***
	F (p-value)	***F (p-value)***	***F (p-value)***	***F (p-value)***	***F (p-value)***	***F (p-value)***
Group	2.51 (0.13)	0.35 (0.56)	0.04 (0.85)	0.69 (0.42)	0.58 (0.46)	1.49 (0.24)
Condition	**8.85 (0.01)**	**6.31 (0.02)**	0.32 (0.58)	**4.69 (0.04)**	**14.19 (0.00)**	**25.59 (0.00)**
Group × Condition	1.17 (0.29)	0.24 (0.63)	0.01 (0.93)	0.05 (0.83)	0.02 (0.89)	0.54 (0.47)
NOP vs. VIS						
Group	3.13 (0.09)	0.73 (0.41)	0.16 (0.69)	0.15 (0.71)	0.13 (0.73)	0.70 (0.41)
Condition	1.98 (0.18)	0.57 (0.46)	0.25 (0.63)	0.79 (0.39)	1.63 (0.22)	**5.92 (0.02)**
Group × Condition	0.85 (0.37)	0.01 (0.91)	0.58 (0.45)	1.20 (0.29)	0.83 (0.37)	1.54 (0.23)

(B) Medio-Lateral Direction (cf. Fig. 3B):

NOP vs. PLAT						
Group	3.02 (0.09)	4.25 (0.05)	0.06 (0.81)	2.01 (0.17)	2.79 (0.11)	6.50 (0.01)
Condition	**123.69 (0.00)**	**161. 94 (0.00)**	**253.77 (0.00)**	**143.51 (0.00)**	**130.26 (0.00)**	**71.55 (0.00)**
Group × Condition	0.19 (0.67)	0.07 (0.79)	0.02 (0.89)	0.00 (0.98)	0.01 (0.91)	0.13 (0.72)
NOP vs. VIS						
Group	0.19 (0.66)	0.01 (0.93)	0.68 (0.42)	0.05 (0.82)	0.29 (0.59)	0.13 (0.73)
Condition	**83.47 (0.00)**	**103.85 (0.00)**	**101.91 (0.00)**	**83.56 (0.00)**	**58.06 (0.00)**	**32.93 (0.00)**
Group × Condition	1.09 (0.31)	1.06 (0.32)	0.74 (0.40)	1.10 (0.31)	1.76 (0.20)	1.73 (0.21)

Notes: Degrees of Freedom were $F(1,19)$ for each ANOVA. Statistically significant effects ($p < 0.05$) are highlighted in bold text.

Table 4

ANOVAs outcomes for lower leg COMs only, separating right and left leg

(A) λ^*_s	NOP vs. PLAT			NOPvs. VIS		
	THIGH	*SHANK*	*FEET*	*THIGH*	*SHANK*	*FEET*
	F (p-value)	***F (p-value)***	***F (p-value)***	***F (p-value)***	***F (p-value)***	***F (p-value)***
Group	0.19 (0.67)	**14.19 (0.00)**	3.72 (0.07)	0.64 (0.43)	**9.23 (0.01)**	0.79 (0.38)
Leg	**6.34 (0.02)**	**64.33 (0.00)**	**9.73 (0.01)**	3.44 (0.08)	**64.49 (0.00)**	**9.04 (0.01)**
Condition	**272.25 (0.00)**	**271.58 (0.00)**	**205.54 (0.00)**	**111.76 (0.00)**	**79.18 (0.00)**	**71.11 (0.00)**
Group × Leg	0.46 (0.51)	60.05 (0.00)	9.13 (0.01)	0.48 (0.49)	51.97 (0.00)	7.79 (0.01)
Group × Condition	0.01 (0.93)	1.15 (0.29)	1.89 (0.19)	4.92 (0.04)	0.38 (0.55)	0.24 (0.63)
Leg × Condition	**8.09 (0.01)**	0.17 (0.69)	**15.58 (0.00)**	2.67 (0.12)	**5.86 (0.03)**	**14.81 (0.00)**
Group × Leg × Condition	**7.08 (0.01)**	**8.18 (0.01)**	**18.53 (0.00)**	**7.03 (0.02)**	**19.61 (0.00)**	**15.43 (0.00)**

(B) MaxFM	NOP vs. PLAT			NOP vs. VIS		
	THIGH	*SHANK*	*FEET*	*THIGH*	*SHANK*	*FEET*
	F (p-value)	***F (p-value)***	***F (p-value)***	***F (p-value)***	***F (p-value)***	***F (p-value)***
Group	2.01 (0.17)	2.79 (0.11)	**16.29 (0.00)**	0.05 (0.82)	0.29 (0.59)	0.13 (0.73)
Leg	0.00 (0.96)	1.58 (0.22)	0.02 (0.89)	4.13 (0.06)	**19.40 (0.00)**	**6.45 (0.02)**
Condition	**143.51 (0.00)**	**130.26 (0.00)**	**72.18 (0.00)**	**83.56 (0.00)**	**58.06 (0.00)**	**32.93 (0.00)**
Group × Leg	0.89 (0.36)	0.08 (0.78)	0.08 (0.78)	0.03 (0.87)	**12.55 (0.00)**	**6.33 (0.02)**
Group × Condition	0.00 (0.98)	0.01 (0.91)	0.71 (0.41)	1.10 (0.31)	1.76 (0.20)	1.73 (0.21)
Leg × Condition	**5.79 (0.03)**	0.38 (0.54)	3.97 (0.06)	0.06 (0.81)	**6.20 (0.02)**	1.46 (0.24)
Group × Leg × Condition	0.02 (0.89)	2.98 (0.10)	1.07 (0.31)	2.36 (0.14)	**33.83 (0.00)**	**4.43 (0.04)**

Notes: Degrees of Freedom were $F(1,19)$ for each ANOVA. Statistically significant effects ($p < 0.05$) are highlighted in bold text.