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# Effects of upper limb loss and prosthesis use on proactive mechanisms of locomotor stability

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ABSTRACT

Persons with upper limb loss (ULL) experience a high prevalence of falls, with the majority of falls occurring when walking. This issue may be related to altered arm dynamics, which play an important role in proactive mechanisms of locomotor stability. This study investigated effects of ULL and prosthesis use on proactive stability mechanisms, particularly if matching the mass and inertia of the impaired limb to the sound limb would enhance locomotor stability. Gait data were collected on adults with unilateral ULL during level walking while: (1) not wearing a prosthesis, (2) wearing their customary prosthesis, (3) wearing a mock prosthesis that matched the sound limb mass and inertia. Main and interaction effects of limb side and condition on trunk rotations, arm swing, step width, free vertical moment, and margin-of-stability were analyzed. Across conditions, arm swing, free vertical moment, and margin-of-stability were 2.27, 1.13, and 1.20 times greater, respectively, on the sound limb side than the impaired limb side. Persons with ULL display asymmetry in proactive mechanisms of locomotor stability with potentially greater medial-lateral stability on the sound limb side irrespective of prosthesis use, but heavier prostheses reduced the walking base of support. This bias may enhance fall risk on the impaired side if the prosthetic limb is used inappropriately to regain balance following a disturbance. Research is warranted to explore the consequences of this asymmetry on perturbation response.

## 1. Introduction

A recent investigation suggested that nearly half of individuals with upper limb loss (ULL) experience at least one fall per year (Major, 2019). Importantly, almost a third of persons with ULL will experience two or more falls, thereby suggesting a systemic effect of ULL on fall likelihood (Major, 2019). The single fall prevalence is comparable to that of individuals with major lower limb loss (52%) (Miller et al., 2001), but greater than the elderly (33%) (Sattin, 1992, Hartholt et al., 2011) and individuals post-stroke (40%) (Belgen et al., 2006, Kerse et al., 2008). As nearly a third of persons with ULL experience an injury due to their most recent fall, this high falls prevalence has important implications for healthcare costs and potential for reduced quality of life similar to that reported for older individuals (Sattin, 1992, Hartholt et al., 2011, Burns et al., 2016). Further, two-thirds of reported falls occurred during walking (Major, 2019). The high fall prevalence in this patient group and the fact that the majority of falls are experienced during walking is perhaps not unreasonable given the role of arm dynamics in locomotor stability, defined here as the ability to control posture when exposed to a disturbance while walking (Patla, 2003, Bruijn et al., 2013). Consequently, locomotor stability is critical to avoiding falls and is facilitated by either proactive mechanisms, defined as those anticipatory strategies for minimizing the impact of a postural disturbance on balance, or reactive mechanisms, defined as those responsive strategies to mitigate perturbation effects when recovering from a postural disturbance (Major et al., 2018). During steady-state gait, arm swing counterbalances the legs to minimize whole-body angular momentum (Elftman, 1939, Bruijn et al., 2008, Collins et al., 2009), which is directly reflected in low free vertical moments (FVMs), or transverse-plane ground reaction moments (Li et al., 2001, Collins et al., 2009). This transverse-plane control is suggested to be a proactive mechanism for aiding stability by effectively minimizing the effect of disturbance-added momentum around the inferior-superior body axis (Neptune and McGowan, 2011, Silverman and Neptune, 2011).

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Moreover, unilateral arm swing restriction in non-impaired individuals encourages increased contralateral arm swing motion to restore whole-body coordination (Ford et al., 2007) and serve as another form of proactive stability mechanisms (Ford et al., 2007, Hu et al., 2012, Nakakubo et al., 2014, Punt et al., 2015). Therefore, this evidence suggesting the role of natural arm dynamics in facilitating locomotor stability has particular implications to individuals with ULL when those dynamics may be absent or altered.

Despite studies characterizing the impact of arm swing in non-impaired persons (Eke-Okoro et al., 1997, Ford et al., 2007, Yizhar et al., 2009), there have been limited investigations on persons with ULL. Characterization of proactive stability mechanisms may help provide insight into the causes behind the high fall prevalence in this group (Bruijn et al., 2013). Generally, persons with unilateral ULL demonstrate asymmetric posture during standing and walking due to a lateral shift in the whole-body center-of-mass (BCoM) (Bertels et al., 2012, Imaizumi et al., 2016). Additionally, while the sound arm displays exaggerated arm swing when walking without an upper limb prosthesis (ULP), wearing a ULP reduces forward trunk transverse rotation on the impaired limb side (Bertels et al., 2012). Moreover, as mass and inertia of an ULP may not match the physiological limb (Bertels et al., 2012), prosthesis use may have implications for managing FVM peaks (Li et al., 2001, Collins et al., 2009). Quantification of such peaks may provide insight into the transverse-plane proactive strategies for controlling momentum to minimize the impact of perturbations around the inferior-superior axis (Li et al., 2001, Bleuse et al., 2006, Collins et al., 2009, Buckley et al., 2010). Similarly, adopted medial-lateral asymmetries may be revealed in the interaction between BCoM dynamics and the base-of-support (BoS) during walking, which would provide insight into frontal-plane proactive stability mechanisms. This interaction can be quantified through the medial-lateral margin-of-stability (MoS), defined as the physical distance between the velocity-weighted BCoM position and the lateral edge of the BoS (Hof et al., 2007, Hof, 2008). A BCoM shift towards one side of the body, whether physically imposed or due to perceptions of body asymmetry, may influence this distance that serves as a spatial buffer in case of a perturbation that laterally displaces the BCoM (Hof et al., 2007, Hof, 2008). Consequently, an improved understanding of locomotor stability in persons with ULL can be gained by characterizing arm swing motion, FVM regulation, and medial-lateral MoS during steady-state gait.

Given the suggested relationships between arm dynamics and locomotor stability and our incomplete understanding of the increased fall prevalence in persons with ULL, this preliminary study investigated the effects of unilateral ULL and prosthesis use on proactive mechanisms of locomotor stability. Specifically, we were interested in identifying if unilateral ULL generates asymmetries in proactive stability mechanisms and if these mechanisms would be affected by modifications to the impaired limb mass and inertial properties. We hypothesized that: (1) persons with unilateral ULL would demonstrate proactive mechanism asymmetry due to the nature of their impairment, and (2) stability would be affected by use of an ULP but matching the impaired limb mass and inertia to the sound limb would improve stability through passive-mechanic restoration of impaired limb arm swing (Collins et al., 2009). Importantly, as previous research has suggested that use of a ULP increases likelihood of frequent falls (2 or more in a given year) by six times (Major, 2019), results from this study will begin to assess if wearing a prosthesis influences locomotor stability. Given that some prosthesis users do not wear their prosthesis (or the same prosthesis) continuously throughout the day (Biddiss and Chau, 2007), the acute systemic effects of using an ULP on proactive stability mechanisms as observed in this study is of clinical relevance. Particularly, if matching the mass and inertia of the impaired limb to the sound limb aids stability, this finding may suggest simple clinical interventions to minimize fall risk.

## 2. Methods

This study was approved by the Northwestern University Institutional Review Board and subjects provided written informed consent.

### 2.1. Subjects

Information on ULL level, etiology, and time since occurrence were collected, as well as gender, age, height, weight, residuum length, leg length, and customary prosthesis type. To provide context within previous epidemiological results on fall prevalence (Major, 2019), subjects were asked how many falls they experienced in the past year, with falls defined as "coming to rest accidentally on the ground or other lower level, other than as a consequence of lost consciousness, a violent blow, stroke, or epileptic seizure" (Askham et al., 1990). Ten individuals with unilateral ULL participated (3 female, 7 transradial/3 transhumeral level ULL, 6 congenital/4 acquired, 50 (SD19) years, 75.3 (SD 18.6) kg, 1.75 (SD 0.08) m). The mean time since ULL was 38 (SD 20) years and there was an equal number of fallers and non-fallers which aligns closely with reported fall prevalence in individuals with ULL (46%) (Major, 2019). Subjects reported no other conditions or medication use that would affect walking or balance. Six subjects were regular prosthesis users (cosmetic (n = 2), body-powered (n = 3), myoelectric (n = 1)), while four subjects did not habitually use a prosthesis. Across participants, 70% rated their activity level as "very active" while 30% rated themselves as "moderately active'.

### 2.2. Experimental protocol

Subjects walked along a level walkway at their normal self-selected speed under three randomly presented prosthesis conditions: (1) without wearing a prosthesis, (2) wearing their customary prosthesis and (3) wearing a mock prosthesis (Fig. 1). The mock prosthesis allowed customization of its length, mass and inertial properties for matching the impaired limb to the sound limb. Length was adjusted by swapping plastic tubes of various lengths, mass was adjusted by adding/removing disc weights, and center-of-mass position was adjusted by translating the group of disc weights along the plastic tube

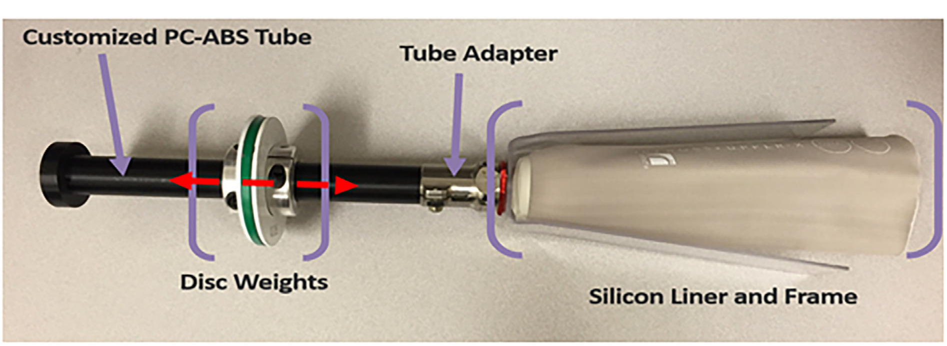


**Fig. 1.** Image of the mock prosthesis. The prosthesis was suspended by the silicon liner and a polyethylene U-frame surrounding the liner was wrapped with Coban (3M, Mapplewood, MN, USA) to minimize pendular motion of the components distal to the liner. The dashed red line denotes linear translation of disc weights which are secured in place with aluminum cuffs. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

and securing in place with cuffs. Using anatomical regression equations (Winter 2009), a MATLAB (Mathworks, Natick, MA) algorithm determined the length of the prosthesis, and magnitude and position of added mass (disc weights) assuming the published anatomical regression parameters applied to the sound limb. To solve for mock prosthesis length, added mass, and added mass position, the required algorithm inputs were:

(1) Sound limb length with fingers extended;
(2) Estimated sound limb mass using the regression equations (percentage of body mass but with consideration of the impaired limb reduced mass);
(3) Estimated sound limb center-of-mass position from the next proximal joint using the regression equations;
(4) Estimated residuum mass using regression equations (percentage of the sound limb segment based on ratio of missing limb length); and
(5) Masses of the plastic tube, silicon liner, frame, tube adapter and cuffs (Fig. 1).

The mock prosthesis was then built to match the sound limb parameters while accounting for the mass of the residuum limb, masses of prosthetic components, and residuum center-of-mass position (assumed to be halfway along its longitudinal axis). The mock prosthesis did not contain an elbow joint since this is common for standard transhumeral prostheses to lack a free-swinging elbow (Smith et al., 2004). When a subject did not habitually use a prosthesis, the mock prosthesis without added mass was used as the customary prosthesis. Across subjects, the customary prosthesis was on average 56% (SD 23) lighter than the mock. Subjects were provided with ten minutes of walking to accommodate to the mock prosthesis. Kinematic and kinetic data were collected with a digital motion capture system (Motion Analysis Corp. (MAC), Santa Rosa, CA) at 120 Hz and walkway-embedded force plates (AMTI, Watertown, MA) at 960 Hz, respectively, as subjects walked until five clean force plate strikes for each foot were recorded.

Reflective markers were attached to subjects according to a modified Helen Hayes Marker Set (Kadaba et al., 1990) with additional markers on the right/left fifth metatarsal, wrist joint equidistant between the ulnar and radial styloid processes, acromion processes, and humeral lateral epicondyles. For transhumeral cases, impaired limb epicondyle markers were attached to the lateral bottom third of the residuum and wrist markers were attached to the lateral residuum end when not wearing a prosthesis, and corresponding locations of the sound limb when wearing a prosthesis. For transradial cases, impaired limb wrist markers were attached to the lateral residuum end when not wearing a prosthesis, and corresponding locations of the sound limb when wearing a prosthesis.

### 2.3. Data analysis

Kinematic and kinetic data were filtered using a low-pass Butterworth filter with 6 and 10 Hz cutoff frequency, respectively. The BCoM position was estimated in OrthoTrak software (MAC) using a twelve-segment kinematic model. The model mass, limb segment lengths, and limb segment center-of-mass position were adjusted to account for each prosthesis condition. However, the mass distribution of the impaired limb was not adjusted and a subsequent analysis with such a model modification on five trials each for representative transradial and transhumeral level subjects with the most proximal amputations revealed no substantial changes to the calculated instantaneous BCoM position (average root-mean-square error was 0.34 cm for transradial and 0.54 cm for transhumeral or 4.7% and 7.4% of the average MoS across all subjects/conditions). The lateral BCoM position ($BCoM_{Lat}$) was used to calculate the MoS, defined as the minimum distance between the lateral positions of the extrapolated BCoM (XCoM), the BCoM position extended by the BCoM velocity, and the stance limb fifth metatarsal ($BoS_{R/L}$) (Hof et al., 2007):

$$MoS = BoS_{R/L} - \left(BCoM_{Lat} + VBCoM_{Lat}/\sqrt{\frac{g}{l}}\right) \tag{1}$$

where $VBCoM_{Lat}$ is lateral velocity of the $BCoM_{Lat}$, $g$ is gravitational acceleration (9.81 m/s$^2$), $l$ is effective height of the BCoM from ground (trochanter height (m) times 1.34), and $R/L$ refers to the right or left stance limb. The XCoM always remained within the BoS, so all MoS values were positive. The MoS were averaged across trials for each limb side separately. Lateral MoS reflects frontal-plane control of the BCoM in relation to the walking BoS (Hof et al., 2007, Hof, 2008), may be affected by trunk motion (Major et al., 2013), and has been shown to increase in lower-limb prosthesis users to proactively enhance locomotor stability (Hof et al., 2007, Major et al., 2018).

Through OrthoTrak, trunk kinematics were estimated as the angle between the anatomical axes (i.e., segment coordinate system) and corresponding global axes (zero displacement = axis coincidence). The medial-lateral trunk axis was defined by connecting the acromion markers, inferior-superior axis was defined by connecting the pelvis reference-frame center (geometric center of the plane created by the pelvis markers) and midpoint between acromion markers, and anterior-posterior axis was defined as the axis orthogonal to the other two axes. The angle between the axis formed by acromion marker and lateral epicondyle and the inferior-superior trunk axis defined arm swing (flexion/extension). For each trial, mean trunk forward flexion, lateral flexion range-of-motion (RoM), transverse rotation RoM, and arm swing RoM were calculated. The medial-lateral distance between heel markers at initial contact events defined step width, and walking speed was estimated as the average velocity of the sacrum in the direction of walking progression. To gain insight into the ability of subjects to bilaterally control transverse rotational dynamics, the FVM was calculated using MATLAB software according to the convention of Li et al. (2001) as the sum of instantaneous vertical moment and moments of the medial-lateral and anterior-posterior (shear) forces, with body-weight-normalized peak (i.e., maximum absolute) values during stance averaged across trials for each side separately. Peak FVM was selected as a means of direct measurement of proactive control strategies separated by limb side as it is proportionally related to whole body angular momentum (Collins et al., (2009) and results may inform the value of characterizing independent body segment angular momentum in future analyses. All metrics were averaged across trials.

### 2.4. Statistical analysis

Data normality was confirmed using the Shapiro-Wilk test. For FVM, MoS, and arm swing RoM, a two-way ANOVA analyzed main and interaction effects of limb side (sound, impaired) and prosthesis condition (no, customary, mock prosthesis). For the trunk kinematics, step width, and walking speed, a one-way ANOVA analyzed main effects of prosthesis condition. Results were interpreted according to a Greenhouse-Geisser correction when Mauchly's Test of Sphericity was violated. A Bonferroni correction was used for multiple comparison post-hoc tests. The critical $\alpha$ was set at 0.05 for all analyses.

## 3. Results

Walking speed was not affected by condition ($P = 0.973$, *partial* $\eta^2 = 0.003$) and was on average 1.20 (SD 0.01) m/s. The main effect of side was significant for MoS ($F(1, 9) = 13.446$, $P = 0.005$, *partial* $\eta^2 = 0.599$) and arm swing RoM ($F(1, 9) = 33.627$, $P < 0.001$, *partial* $\eta^2 = 0.789$), and marginally significant for FVM ($F(1, 9) = 5.026$, $P = 0.052$, *partial* $\eta^2 = 0.358$), with greater values on the sound limb side for all variables (Fig. 2). Peak FVM asymmetry was due to differences in the magnitude of late-stance moments (Fig. 3). The interaction effect of side*condition was significant ($F(2, 18) = 4.729$, $P = 0.022$, *partial* $\eta^2 = 0.344$) only for arm swing RoM, with differences between limbs greatest for the mock prosthesis condition (Fig. 2). Although the

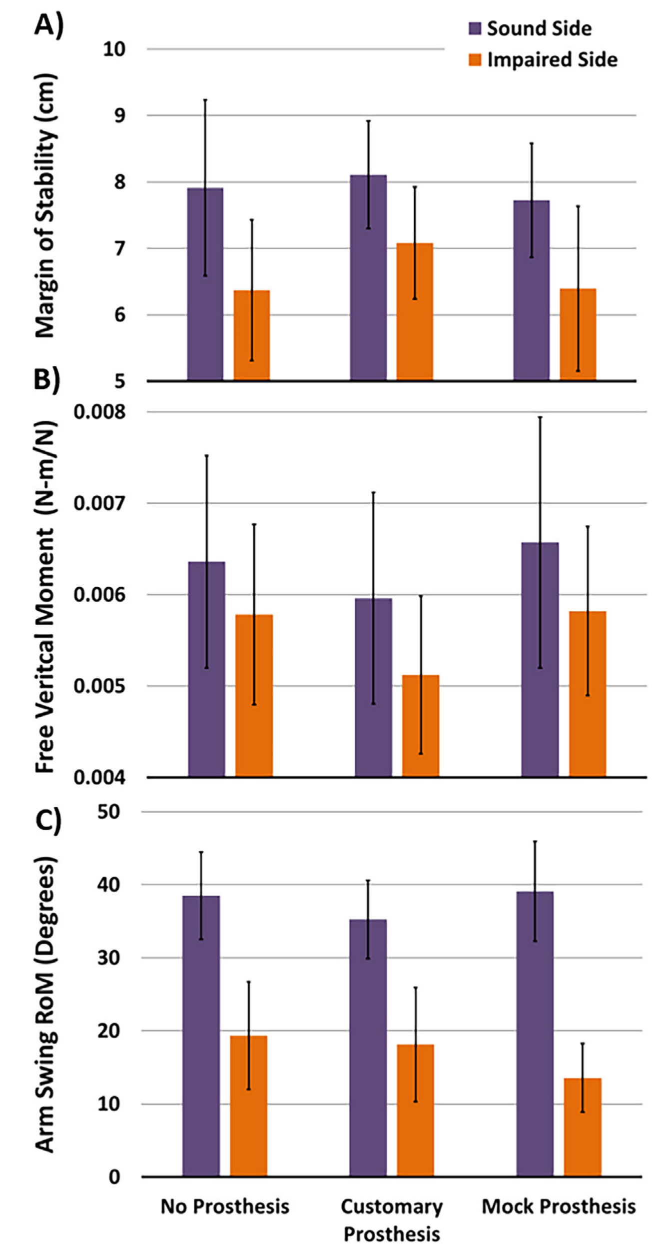


**Fig. 2.** Average (A) margin of stability, (B) free vertical moment, and (C) arm swing range-of-motion (RoM) across subjects as separated by limb side and prosthesis condition. Error bars denote 95% confidence intervals.

mock prosthesis lacked an elbow joint, average arm swing RoM for transhumeral participants (12°) was not substantially different than transradial participants (14°). Although asymmetry was consistently present and statistically significant, the main effect of condition was not significant for MoS ($P = 0.478$, *partial* $\eta^2 = 0.067$), FVM ($P = 0.091$, *partial* $\eta^2 = 0.233$), and arm swing RoM ($P = 0.205$, *partial* $\eta^2 = 0.161$), and no post-hoc comparisons were significant ($P \geq 0.104$).

However, the main effect of condition was significant for step width ($F(2, 18) = 4.176$, $P = 0.032$, *partial* $\eta^2 = 0.317$) and trunk transverse RoM ($F(2, 18) = 4.688$, $P = 0.023$, *partial* $\eta^2 = 0.342$), but not significant for mean trunk forward flexion ($P = 0.554$, *partial* $\eta^2 = 0.063$) and lateral flexion RoM ($P = 0.892$, *partial* $\eta^2 = 0.013$) (Figs. 4–6). No transverse RoM post-hoc comparisons were significant ($P \geq 0.117$), but step width for the mock prosthesis condition was significantly ($P = 0.024$) smaller than the no prosthesis condition (Fig. 5).

## 4. Discussion

This study investigated the effects of ULL and prosthesis use on proactive mechanisms of locomotor stability. The first hypothesis that persons with unilateral ULL would demonstrate asymmetry in proactive mechanisms was supported. Arm swing RoM and MoS were significantly greater on the sound limb side, with a similar trend in FVM (Fig. 2). The impaired limb arm swing RoM was approximately half of the sound limb, and similar to observations in individuals with unilateral shoulder disarticulation (Bertels et al., 2012), not impacted by prosthesis use. The sound limb displayed greater RoM than that documented for non-impaired individuals (Perry and Burnfield, 2010) and the sound limb of persons with shoulder disarticulation (32°) (Bertels et al., 2012) walking at slightly faster speeds (1.4–1.5 m/s). Exaggerated arm swing in the absence of contralateral arm swing during walking may act as a compensatory mechanism for coordination to counterbalance leg-generated momentum, thereby controlling whole-body angular momentum (Ford et al., 2007, Bruijn et al., 2008, Lulic et al., 2008, Collins et al., 2009) and minimizing the impact of momentum-generating perturbations such as trips (Pijnappels et al., 2010). Furthermore, exaggerated arm swing has been shown to enhance locomotor stability through decreasing the motor control system's sensitivity to perturbations as estimated through divergence exponents (Hu et al., 2012, Punt et al., 2015, Wu et al., 2016), and increasing the smoothness and rhythmicity of medial-lateral trunk kinematics which has been linked to reduced fall risk (Nakakubo et al., 2014).

The observed MoS asymmetry has implications to locomotor stability in the context of responding to laterally-directed perturbations. The MoS represents the physical buffer between the XCoM and the BoS, whereas an extension of the XCoM beyond the BoS would necessitate a response to redirect the BCoM and maintain balance (Hof et al., 2007, Hof, 2008). The smaller impaired limb side MoS may suggest increased

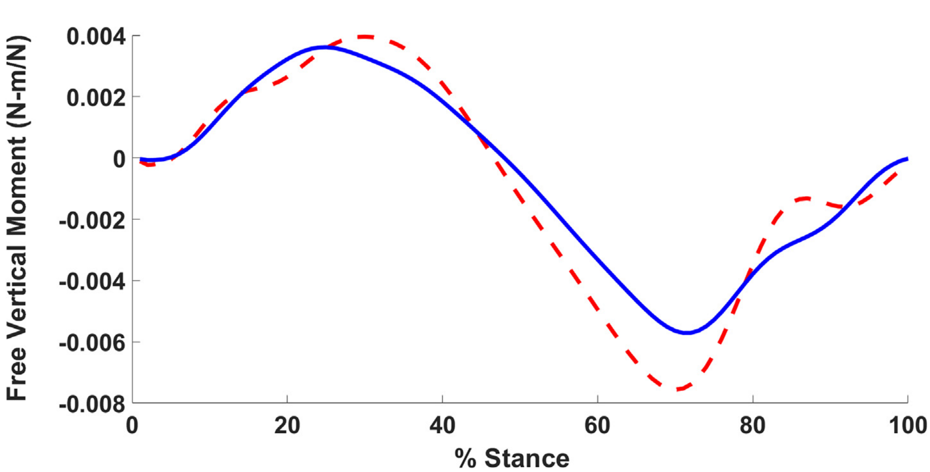


**Fig. 3.** Ensemble average free vertical moment with respect to stance time for the impaired limb side leg (blue solid line) and sound limb side leg (red dashed line) of one representative subject when walking without a prosthesis. Positive values denote (top-down) counter-clockwise reaction moments while negative values denote clockwise reaction moments. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

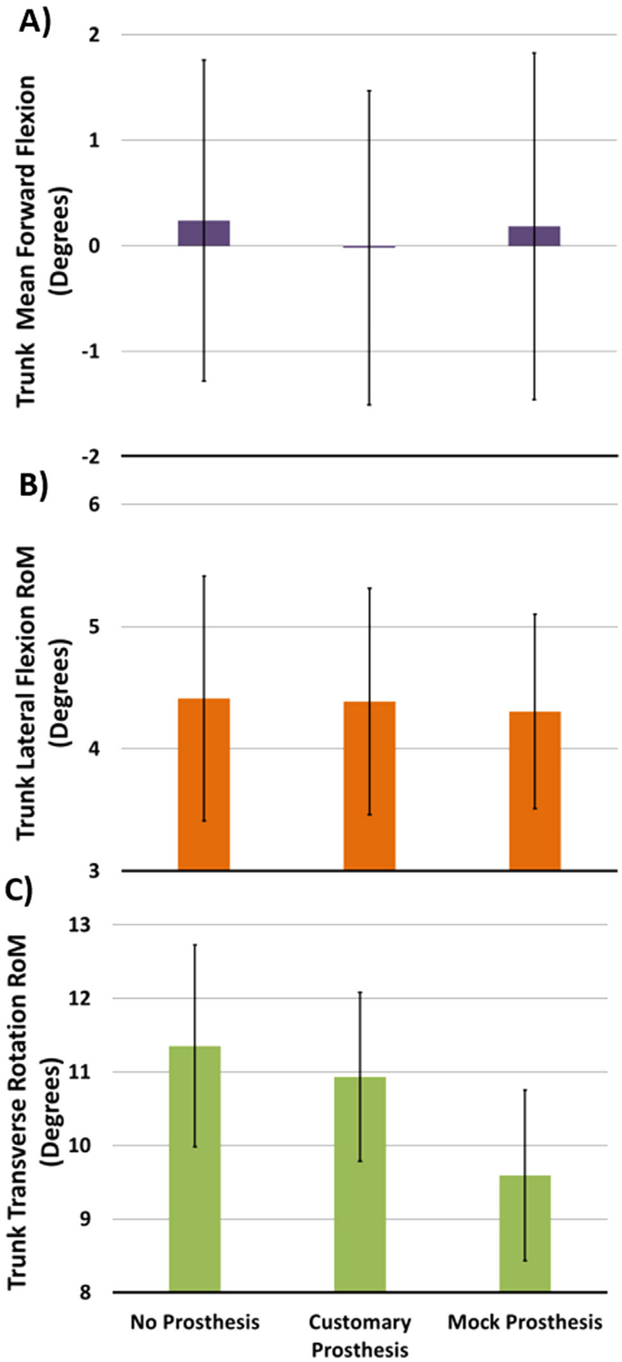


**Fig. 4.** Average trunk mean forward flexion, lateral flexion range-of-motion (RoM), and transverse rotation RoM separated by prosthesis condition. Error bars denote 95% confidence intervals.

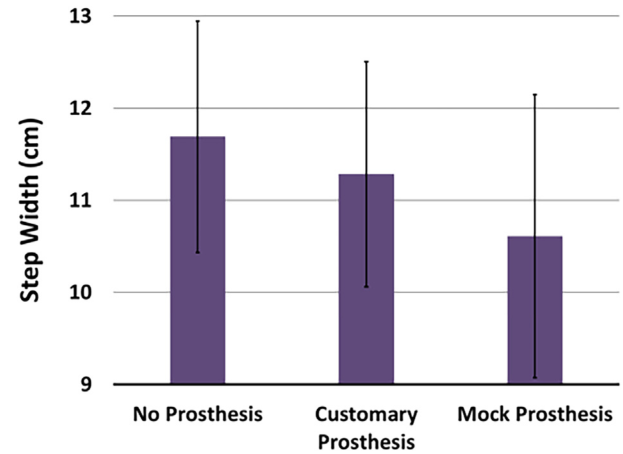


**Fig. 5.** Average step width across subjects separated by prosthesis condition. Error bars denote 95% confidence intervals.

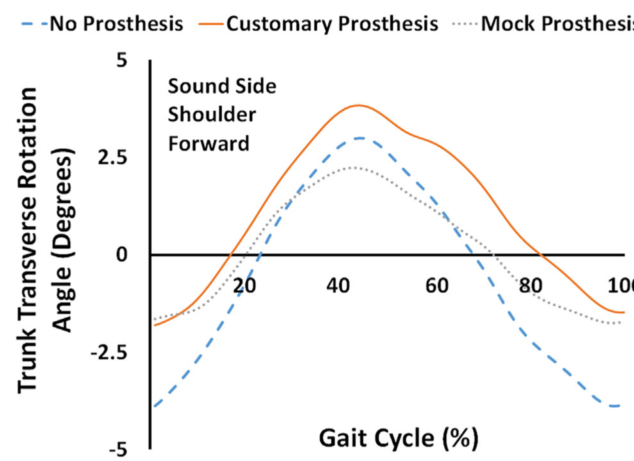


**Fig. 6.** Ensemble average trunk transverse rotation angle with respect to gait cycle time for the three prosthesis conditions of one representative subject.

likelihood of the XCoM position breaching the BoS if a perturbation was directed towards that side. To recover from such a perturbation, individuals with ULL may be at a disadvantage if their directed motor responses do not effectively account for the inertial characteristics of the impaired limb (residuum plus prosthesis) to aid successful redirection of the BCoM and maintain balance (Marigold and Patla, 2002, Marigold et al., 2003, Roos et al., 2008, Pijnappels et al., 2010). A further disadvantage to arresting falls may result from directly using the prosthetic limb to control body descent (e.g., grasping, eccentric contractions) when active joint control is absent. In fact, the recent epidemiological study on falls in persons with ULL reported that 10% of the most recent falls were related to 'failure of arresting a the fall with the arms' (Major, 2019). Ballistic arm extension is a common response when losing balance to avoid bodily harm (Hsiao and Robinovitch, 1998, Robinovitch et al., 2003, Feldman and Robinovitch, 2007), and using a prosthetic arm not intended for this purpose could uniquely place ULP users at an increased risk of falls. Given that some ULP users do not wear their prosthesis at all times but only as dependent on task demands (Biddiss and Chau, 2007), they must continuously accommodate for these changes in upper limb anatomical properties. This result may partially explain the nearly six times greater likelihood of experiencing a fall when using a ULP compared to individuals with ULL who report not using a ULP (Major, 2019).

Evidence suggests that the FVM is carefully regulated through segment dynamics and muscle activation to proactively enhance locomotor stability (Li et al., 2001, Bleuse et al., 2006, Collins et al., 2009). Consequently, presence of FVM asymmetry may have implications for perturbation response strategies that would be dependent on the timing and direction of the perturbation with respect to the gait cycle. In this study, an asymmetry in peak FVM was observed, with differences localized to larger late-stance moments on the sound side limb (Fig. 3). The FVM peak values were on average slightly greater than those reported for non-impaired individuals, and approached values when able-bodied persons walk with their arm swing restricted (Li et al., 2001, Collins et al., 2009). An expectation that the FVM peaks would be greater on the impaired limb side is reasonable given the absence of arm momentum to counterbalance leg momentum, but this assumption may neglect the excessive forward sound arm swing observed in this cohort with minimal prosthetic arm swing (Fig. 2). However, further research is warranted to determine if such FVM asymmetry is real given its marginal significance in this study and if so what its relationship is with whole-body angular momentum.

It remains unclear if the second hypothesis–use of a prosthesis would affect locomotor stability and that creating upper limb mass and inertia symmetry would benefit stability–was supported due to inconsistent results in the observed selection of proactive mechanisms. Wearing a prosthesis had no noticeable effect on MoS, FVM, or arm swing RoM (Fig. 2). The mock prosthesis did not benefit from passive dynamics to generate increased arm swing, and this result may be partially due to the specific reductions in trunk transverse rotation (Fig. 3). Agreeing with findings in persons with unilateral shoulder disarticulation (Bertels et al., 2012), the smaller transverse RoM was due to reduced forward rotation of the impaired limb side (Fig. 6). The mock prosthesis was heavier than customary prostheses and this may have restricted transverse-plane motion that would facilitate arm swing and more symmetric FVM peaks (Collins et al., 2009). To note, use of a prosthesis did not affect trunk lateral RoM and mean forward flexion which were near non-impaired values (Major et al., 2013). The mock prosthesis did however result in a significant decrease in step width, which would suggest use of a less conservative walking strategy and perceived greater balance confidence (Maki, 1997, Major et al., 2013, Major et al., 2016). Previous research has suggested that increased balance confidence in persons with ULL is protective against fall likelihood (Major, 2019). An alternative interpretation is that step width reduction could be a reaction to the imposed limitations on trunk and arm motion which have been shown to influence temporal-spatial parameters (Eke-Okoro et al., 1997, Yizhar et al., 2009), and this would further reduce the medial-lateral BoS. However, results from this study suggest that MoS was not significantly affected by prosthesis condition, indicating that proactive mechanisms are employed to maintain a consistent MoS on both limb sides despite the observed asymmetry. Consequently, although properties of a traditional prosthesis were not adjusted in this case, the benefits of modifying prosthesis properties as a clinical intervention to enhance stability warrants further investigation. Conversely, a clinical takeaway from these results is that the while prosthesis use may not encourage proactive mechanisms for improved locomotor stability, it does not seem to impair stability in the context of this study. However, the effects of prostheses use on overall lower extremity kinematics and kinetics should be considered to account for its holistic impact on user mobility.

There are several study limitations to consider when interpreting these results. First, there is no criterion standard for quantifying locomotor stability and these results are based on a selection of steady-state biomechanical measures to characterize proactive stability mechanisms. Other techniques including non-linear analysis, such as estimates of system dynamic stability via divergence exponents (Hu et al., 2012, Punt et al., 2015), and perturbation experiments, such as controlled trips or slips (Marigold and Patla, 2002, Marigold et al., 2003, Roos et al., 2008, Pijnappels et al., 2010), may provide additional answers to this question. Systematic studies on the motor responses of persons with ULL to controlled perturbations can characterize reactive stability mechanisms and may help reveal strategies that are useful for successful disturbance recovery. Second, the cohort consisted of an unbalanced number of subjects with ULL levels. The sample size of this study did not allow for analysis on the main effect of level with adequate statistical power, but epidemiological research suggests that ULL level is not a significant contributor to frequent fall likelihood in this group (Major, 2019). Furthermore, there were four subjects that did not regularly use a prosthesis and as limited sample size precluded analysis of habitual use main effects, these results of acute effects should be interpreted accordingly. However, this scenario is not uncommon and therefore the results of this preliminary study hold clinical relevance despite the variability of the customary prosthesis condition. Finally, the mock prosthesis did not resemble a typical ULP and although this model standardized the protocol, it may have masked the true effects of prosthesis use. One study suggests that ULP embodiment may influence standing balance (Imaizumi et al., 2016) and the effects of this relationship on locomotor stability are unknown. Overall, despite the limitations imposed by the study sample and methodological accommodation, the results provide motivation and help guide study designs for future explorations on this important topic.

## 5. Conclusion

In conclusion, persons with ULL display noticeable asymmetry in proactive mechanisms of locomotor stability, which may have implications for motor strategies when responding to a walking disturbance. Furthermore, as acute changes in prosthesis use did not considerably affect proactive stability mechanisms, this observation suggests that these individuals consistently implement different balance control strategies between the sound and impaired limb side. When considering that an ULP may be maladapted for balance control, the smaller MoS on the impaired limb side may suggest greater fall risk on that side of the body. Future work should focus on reactive mechanisms of locomotor stability by characterizing the motor strategies of persons with ULL when responding to a perturbation.

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## Declaration of Competing Interest

The authors declare no conflict of interest with this work.

## Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jelekin.2019.07.012.

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