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Trunk-Pelvis motions and spinal loads during upslope and downslope walking among persons with transfemoral amputation

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ABSTRACT

Larger trunk and pelvic motions in persons with (vs. without) lower limb amputation during activities of daily living (ADLs) adversely affect the mechanical demands on the lower back. Building on evidence that such altered motions result in larger spinal loads during level-ground walking, here we characterize trunk-pelvic motions, trunk muscle forces, and resultant spinal loads among sixteen males with unilateral, transfemoral amputation (TFA) walking at a self-selected speed both up ("upslope"; 1.06 ± 0.14 m/s) and down ("downslope"; 0.98 ± 0.20 m/s) a 10-degree ramp. Tri-planar trunk and pelvic motions were obtained (and ranges-of-motion [ROM] computed) as inputs for a non-linear finite element model of the spine to estimate global and local muscle (i.e., trunk movers and stabilizers, respectively) forces, and resultant spinal loads. Sagittal- ($p = 0.001$), frontal- ($p = 0.004$), and transverse-plane ($p < 0.001$) trunk ROM, and peak mediolateral shear ($p = 0.011$) and local muscle forces ($p = 0.010$) were larger (respectively 45, 35, 98, 70, and 11%) in upslope vs. downslope walking. Peak anteroposterior shear ($p = 0.33$), compression ($p = 0.28$), and global muscle ($p = 0.35$) forces were similar between inclinations. Compared to previous reports of persons with TFA walking on level ground, 5–60% larger anteroposterior and mediolateral shear observed here (despite ~0.25 m/s slower walking speeds) suggest greater mechanical demands on the low back in sloped walking, particularly upslope. Continued characterization of trunk motions and spinal loads during ADLs support the notion that repeated exposures to these larger-than-normal (i.e., vs. level-ground walking in TFA and uninjured cohorts) spinal loads contribute to an increased risk for low back injury following lower limb amputation.

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1. Introduction

Low back pain (LBP) is commonly reported among persons with lower limb amputation (LLA) with prevalence rates nearly twice that of the general population (52–71% vs. 6–33%, respectively) (Ehde et al., 2001; Smith et al., 1999), and has a negative impact on long-term quality of life (Kulkarni et al., 2005; Taghipour et al., 2009). Persons with transfemoral amputation (TFA) are at particular risk for LBP with up to 81% reporting LBP (Kulkarni et al., 2005). Differences in trunk and pelvic motions/postures, relative to uninjured individuals, during ADLs are often perceived as primary contributors to the onset and recurrence of LBP (Devan et al., 2015). One proposed mechanism for LBP secondary to LLA suggests these altered motions adversely affect the mechanical demands on the low back, thereby increasing spinal loads (Davis and Marras, 2000; Shojaei et al., 2016) and thus increasing risk for LBP development due to cumulative fatigue of the lower back tissues (Coenen et al., 2014; Kumar, 2001). For example, previous reports have observed larger mechanical demands and resultant spinal loads while walking in persons with vs. without LLA (Hendershot and Wolf, 2014; Shojaei et al., 2016). Among those with vs. without LLA, increases in walking speed also resulted in larger increases in muscle responses and spinal loads

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(Hendershot et al., 2018). However, little is known regarding the mechanical demands on the low back for persons with LLA while walking over non-level ground (i.e., slopes).

Prior work characterizing trunk and pelvic motions during sloped walking has largely focused on uninjured populations. Although sagittal-plane trunk and pelvic movement patterns and ranges of motion (ROM) remain unchanged while walking on a sloped (vs. level) surface, anterior trunk posture (i.e., mean flexion) tends to be linearly related with surface inclination (Hong et al., 2014; Leroux et al., 2002; Leu et al., 2012). That is, as a surface becomes more inclined (i.e., larger positive slope), mean trunk flexion will increase (Leroux et al., 2002; Leu et al., 2012). Conversely, as a surface becomes more declined (i.e., larger negative slope), mean trunk flexion will decrease (Hong et al., 2014; Leroux et al., 2002). Pelvic posture in the sagittal plane follows a similar pattern with increasing incline (Hong et al., 2014; McIntosh et al., 2006), but remains similar in downslope vs. level walking (McIntosh et al., 2006). In the frontal plane, trunk and pelvic movement tended to increase in upslope vs. level walking (Leu et al., 2012) and decrease in downslope vs. level walking (Hong et al., 2014). Despite these reports, no study to date has characterized trunk and pelvic motion during sloped walking among individuals with LLA.

Instead, previous efforts to characterize sloped walking in individuals with LLA have focused on characterizing lower-limb strategies in response to walking on sloped surfaces (Okita et al., 2018; Villa et al., 2015; Vrieling et al., 2008). Persons with transtibial amputation make adjustments similar to those of uninjured individuals; knee flexion is increased during swing in upslope walking and stance in downslope walking (Vrieling et al., 2008). Respectively, these strategies help provide toe clearance over a higher surface and facilitate the lowering of the body onto a lower surface (Vrieling et al., 2008). Persons with TFA, however, cannot readily control knee flexion during gait and thus rely on compensatory hip motions to navigate sloped surfaces (Villa et al., 2015; Vrieling et al., 2008). While these reports provide insight into altered motions within the lower-limb among persons with LLA on sloped surfaces, they do not investigate upper-body (i.e., trunk and pelvic) motions. Therefore, the extent to which walking on sloped surfaces affects the mechanical loads experienced at the low back remains unclear. Thus, the purpose of this study was to characterize trunk-pelvic motions along with the corresponding trunk muscle forces and resultant spinal loads while walking on inclined and declined surfaces among persons with TFA. As prior work in uninjured populations has observed larger trunk and pelvic motions in upslope vs. downslope walking, it is hypothesized that this relationship will remain in individuals with TFA. Consequently, trunk muscle forces, and resultant spinal loads are also hypothesized to be larger in upslope vs. downslope walking.

2. Methods

2.1. Experimental procedures

This study retrospectively evaluated biomechanical data from 16 males with traumatic, unilateral TFA – mean (standard deviation) age: 32.3 (5.9) years, stature: 179.0 (6.4) cm, body mass: 86.3 (10.0) kg – while walking at a self-selected pace both up ("upslope") and down ("downslope") a 10 m ramp set at 10°. Twelve participants used microprocessor-controlled knees while four used single-axis hydraulic knees. All participants reported no functional impairments to the contralateral (i.e., intact) limb, and were independently ambulatory without the use of an assistive device (e.g., cane or walker) for 4.8 (1.6) months prior to data collection. This retrospective study was approved by Institutional Review Boards at both Walter Reed National Military Medical Center and University of Kentucky.

Three-dimensional trunk and pelvis kinematics were collected (120 Hz) via a motion capture system consisting of at least 10 cameras (Motion Analysis Corp., Santa Rosa, CA, USA, or Vicon Inc., Oxford, UK) and reflective markers placed at the C7 and T10 spinal processes, sternal notch, xiphoid, and bilaterally across the acromion and anterior/posterior superior iliac spines. The trunk anatomical reference frame was constructed based on prior recommendations (Armand et al., 2014). Foot kinematics were similarly tracked via markers placed bilaterally on the calcaneus, second and fifth metatarsals, and the hallux. Raw kinematic data (i.e. marker trajectories) were low-pass filtered (Butterworth, cut-off frequency 6 Hz).

2.2. Dependent measures and analyses

Temporal-spatial measures, and kinematic data of the trunk and pelvis, were calculated and analyzed in Visual3D (C-Motion, Germantown, MD). Stride width was calculated as the absolute difference in mediolateral positions of the two calcaneus markers at successive heel strikes. Stride length was defined as the absolute difference in anteroposterior position of the right calcaneus at right heel strike (RHS) and subsequent RHS. Stride time was calculated as the time between consecutive RHS and speed was defined as stride length divided by stride time.

Global trunk and pelvis angles and pelvis center of mass position were time-normalized over each stride (RHS to RHS) and averaged across all trials for each participant. Tri-planar global trunk and pelvic ROM were calculated as the difference between the maximum and minimum angles in all three planes. Trunk and pelvic postures were defined as the average sagittal plane angle (e.g., trunk flexion/extension) throughout the gait cycle.

To estimate trunk muscle forces, and resultant spinal loads, three-dimensional angular kinematics of the trunk and pelvis, three-dimensional translational kinematics of the pelvis COM, and participant body mass were used as inputs to a non-linear finite element model of the spine with an optimization-based iterative procedure (Bazrgari et al., 2007). A detailed description of modeling procedures and validations can be found in previous works (Bazrgari et al., 2007; Bazrgari et al., 2008a; Bazrgari et al., 2008b). Briefly, the kinematics-driven model consists of six rigid elements, representing the thorax (T1-T12) and each lumbar vertebra (L1-L5), and six flexible, non-linear beam elements characterizing the non-linear stiffness of each lumbar motion segment between T12 and S1. Mass and inertial properties were distributed according to previously reported ratios (de Leva, 1996; Pearsall et al., 1996; Zatsiorsky and Seluyanov, 1983). In total, 56 muscles are represented in the model (Shirazi-Adl et al., 2005) – 46 connecting individual lumbar vertebrae to the pelvis (i.e., local) and 10 connecting the thoracic spine/rib cage to the pelvis (i.e., global).

Muscle responses were estimated using a heuristic optimization procedure (Shojaei et al., 2015; Shojaei et al., 2018) wherein the finite element model, subjected to above described kinematics and kinetics boundary conditions, is implemented to find a set of lumbar segmental kinematics (i.e., model inputs) and muscle forces (i.e., model outputs) that are associated with the minimum sum of squared muscle stress across all muscles in the model (i.e., the cost function). A custom MATLAB (Mathworks, Inc., Natick MA, USA, version 8.6) controlled the optimization procedure while a finite element software package (ABAQUS; version 6.13, Dassault Systemes Simulia, Providence, RI, USA) was used to estimate muscle forces and associated tri-planar spinal loads (i.e., compressive forces and both mediolateral [ML] and anteroposterior [AP] shear forces) at all levels of the lumbar spine. Rather than reporting individual muscle forces, summations were taken across the 10 global muscles as well as the 46 local muscles, hereby referred to as simply "global" and "local" muscle forces respectively. Similarly, spinal loads were compiled from the L5/S1 level, rather than at all levels of the lumbar spine, as that is where the maximum spinal loads occur. Peak values of local and global muscle forces and spinal loads across the gait cycle were extracted and normalized to participant body mass. All dependent variables were compared between Upslope and Downslope walking using paired t-tests, with significance concluded at $p < 0.05$.

3. Results

Self-selected walking speeds were similar ($p = 0.24$) between inclinations (Table 1); however, when walking downslope vs. upslope, participants walked with shorter stride times ($p < 0.001$) and stride lengths ($p = 0.001$), and larger stride widths ($p = 0.007$).

Tri-planar trunk ROM were larger ($p < 0.004$) in upslope vs. downslope walking. However, pelvis ROM was larger only in the frontal plane ($p = 0.002$) in upslope vs. downslope walking, and similar between inclinations in the sagittal ($p = 0.90$) and transverse ($p = 0.33$) planes (Table 2). In upslope vs. downslope walking, participants walked with larger ($p < 0.001$) anterior lean of both the trunk (15.0 [6.8]° vs. 3.0 [4.7]°) and pelvis (25.7 [7.8]° vs. 14.2 [4.3]°; Fig. 1).

Peak ML shear forces ($p = 0.011$) were larger in upslope vs. downslope walking, while AP shear ($p = 0.33$) and compression ($p = 0.28$) forces were similar between inclinations (Fig. 2A). Peak local muscle forces were also larger ($p = 0.010$) in upslope vs. downslope walking, while global muscle forces were similar ($p = 0.35$) between inclinations (Fig. 2B).

4. Discussion

In this study, trunk-pelvis motions, trunk muscle forces, and resultant spinal loads were investigated among persons with TFA during walking on inclined surfaces. In upslope vs. downslope walking, peak local muscle forces and peak ML shear forces were larger, in accordance with larger trunk ROM which partially support our hypotheses.

Here, persons with LLA walking upslope resulted in larger sagittal and lateral (frontal) pelvic tilt as well as larger lateral, axial, and sagittal trunk motions relative to those previously reported in level walking (Goujon-Pillet et al., 2008; Russell Esposito and Wilken, 2014). Such larger motions among persons with LLA is contrary to prior reports in uninjured populations that suggest sagittal-plane trunk and pelvic movement patterns and ROM remain unchanged while walking upslope vs. a level surface (Hong et al., 2014; Leroux et al., 2002; Leu et al., 2012). The increased differences in multiple planes seen in persons with TFA may indicate the increased need for neuromuscular control and demand of trunk musculature. Additionally, these alterations are likely indicative of the compensatory strategies adopted by persons with TFA while walking upslope. For example, the larger lateral pelvis ROM is indicative of a hip-hiking strategy in which persons with TFA raise the pelvis on the prosthetic side (i.e., lateral tilt towards the intact side; Fig. 1) to help provide toe clearance over a raised surface. Meanwhile, the large trunk ROM, particularly axial twist (Table 2), may be a mechanism to generate the power necessary to move the swing limb forward and upwards. However, the trunk and pelvis postures in this study (Fig. 1) showed larger trunk and pelvis flexion in upslope vs. downslope walking, which is consistent with prior literature in uninjured controls (Leroux et al., 2002; Leu et al., 2012).

During downslope walking, trunk and pelvis motions observed here were similar to previous reports of persons with LLA walking on level ground (Goujon-Pillet et al., 2008; Russell Esposito and Wilken, 2014). Such similarities may be a result of the decreased stability when walking downhill, indicated by shorter, wider, and more frequent steps when compared to upslope walking (Table 1). Due to this instability, it's likely that participants reduced trunk and pelvic motion, and thus overall center-of-mass motion, in order to reduce the likelihood of falls. However, the movement patterns of the pelvis, particularly in the frontal plane, suggest persons with TFA adopt compensatory motions when walking downslope. Throughout the gait cycle, the pelvis is laterally tilted towards the intact side, which effectively lengthens the limb and helps facilitate lowering the body COM onto a lower surface. In uninjured populations, this is accomplished via increased contralateral knee flexion (Lay et al., 2006). However, persons with TFA cannot actively control contralateral (i.e., prosthetic) knee flexion and thus compensate at the pelvis. Moreover, the concomitant findings of both the trunk and pelvis postures exhibiting similar patterns suggest persons with TFA utilize a "guarding" strategy (Arendt-Nielson et al., 1995; Russell Esposito and Wilken, 2014) not present in uninjured controls (Hong et al., 2014; Leu et al., 2012).

Unsurprisingly, such changes in kinematics affected the mechanical environment of the low back. Compared to prior reports in level walking (at a self-selected pace), larger AP/ML shear forces were observed in persons with TFA when walking upslope (7.4 (3.8) vs. 5.7 (2.0) N/kg and 15.2 (8.1) vs. 9.5 (4.1) N/kg respectively) (Hendershot et al., 2018), though compression loads, and both global and local muscle forces, were similar (25.2 (4.9) vs. 25.5 (6.0) N/kg, 12.3 (3.6) vs. 13.2 (4.3) N/kg, and 9.5 (1.1) vs. 10.0 (2.7) N/kg respectively) (Hendershot et al., 2018; Shojaei et al., 2016). Notably, slower self-selected walking speeds were observed in persons with TFA when walking upslope (Table 1) compared to prior reports in level-ground walking (1.06 ± 0.14 vs. 1.24 ± 0.14 m/s, respectively; (Hendershot et al., 2018). Despite these slower speeds, the results indicate mechanical demand at the low back is increased in upslope vs. level walking, suggesting that increases in slope present a greater demand on the lower back relative to increases in walking speed. When comparing upslope walking to level-ground walking at 1.0 m/s (in persons with TFA), the previously described differences in peak AP/ML shear are more pronounced and compression loads are also larger; however, muscle forces remained similar (Hendershot et al., 2018; Shojaei et al., 2016).

When walking downslope, AP and ML shear were similar to previously reported values in level walking at a self-selected pace (6.0 (4.1) vs. 5.7 (2.0) N/kg and 8.9 (4.3) vs. 9.5 (4.1) N/kg respectively) (Hendershot et al., 2018) while compression, and global and local muscle forces were smaller (22.5 (4.1) vs. 25.5 (6.0) N/kg, 11.0 (4.3) vs. 13.2 (4.3) N/kg, and 8.6 (0.9) vs. 10.0 (2.7) N/kg respectively) (Hendershot et al., 2018). As in upslope walking, downslope walking speeds were considerably slower than previously reported

Table 1
Mean (standard deviation) speed, stride time, stride length, and stride width for upslope and downslope walking.

	Speed (m/s)	Stride Time (s)	Stride Length (m)	Stride Width (cm)
Upslope	1.06 (0.14)	1.28 (0.08)	1.35 (0.17)	6.10 (1.55)
Downslope	0.98 (0.20)	1.17 (0.10)*	1.14 (0.22)*	14.39 (3.49)*

* Different from Upslope ($p < 0.05$)

Table 2
Mean (standard deviation) global ranges of motion for the trunk and pelvis in upslope and downslope walking.

	Trunk (°)			Pelvis (°)		
	Sagittal	Frontal	Transverse	Sagittal	Frontal	Transverse
Upslope	10.0 (2.8)	11.9 (3.2)	20.4 (4.6)	9.6 (4.3)	16.5 (5.4)	11.6 (3.5)
Downslope	6.9 (2.0)*	8.8 (2.2)*	10.3 (3.6)*	9.1 (5.1)	11.0 (3.1)*	12.8 (4.3)

* Different from Upslope ($p < 0.05$).

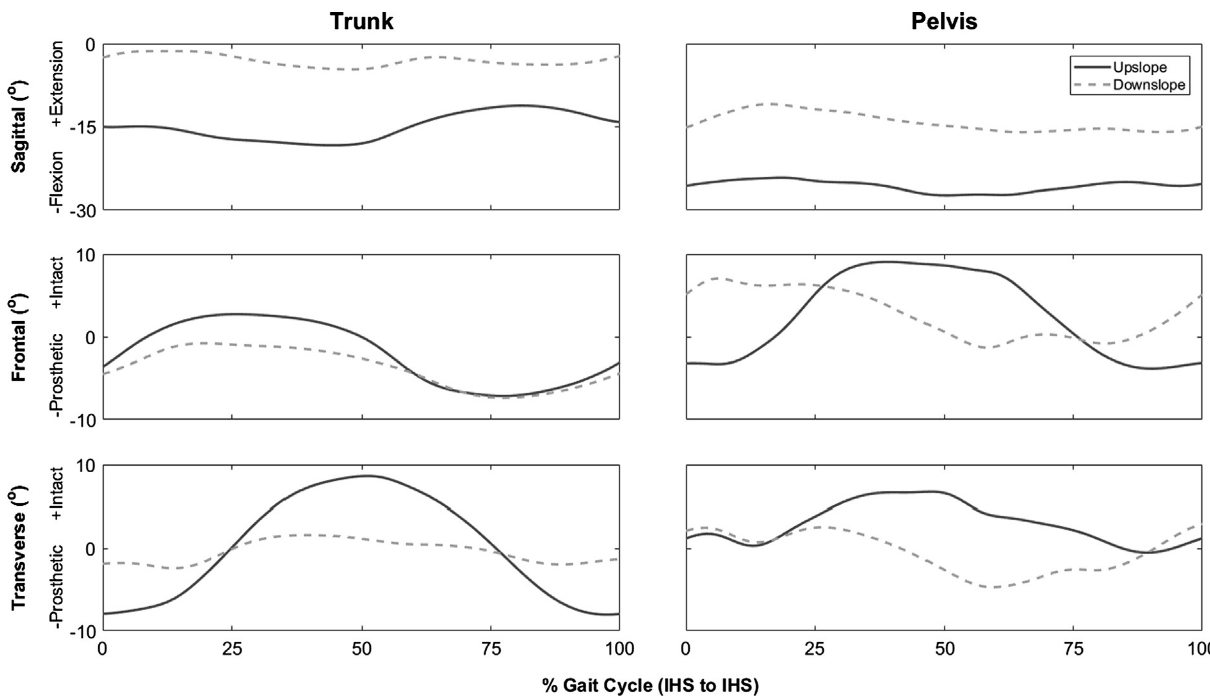

Fig. 1. Mean trunk (left) and pelvis (right) angles in the sagittal (top), frontal (middle), and transverse (bottom) planes during the intact-side gait cycle (intact heel strike [IHS] to IHS) while walking upslope (solid) and downslope (dashed). Negative values indicate anterior lean, lateral lean towards the prosthetic side (i.e., intact side higher), and prosthetic-side leading (i.e., prosthetic-side shoulder more anterior) axial rotations in the sagittal, frontal, and transverse planes respectively.

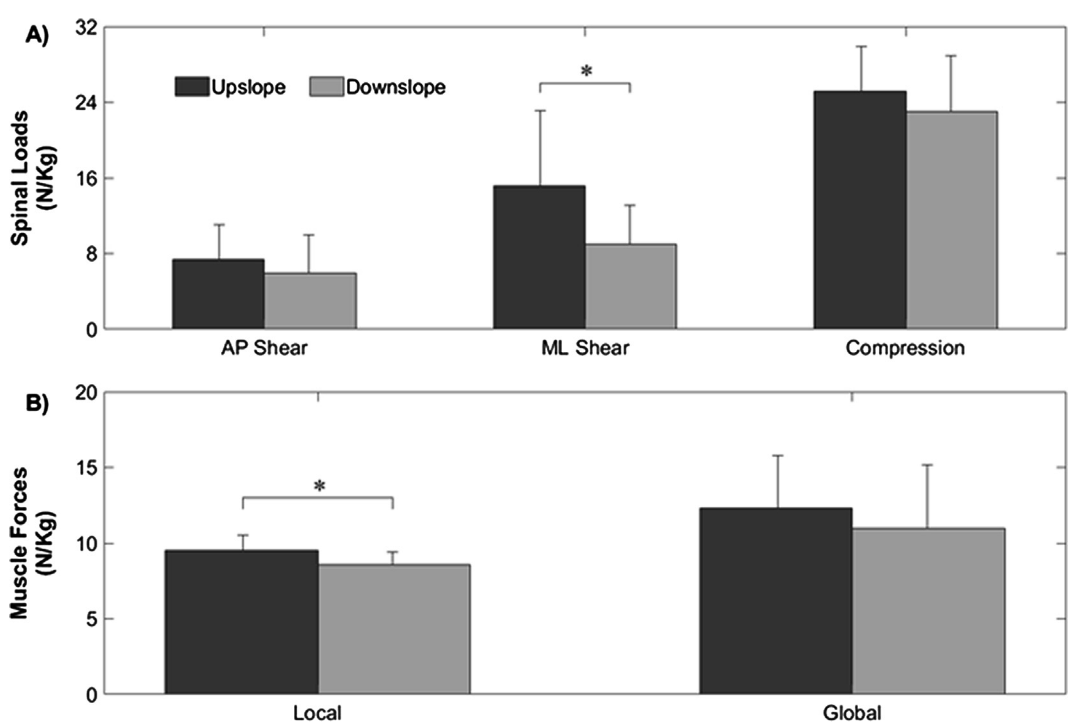

Fig. 2. Ensemble mean (standard deviation) peak spinal loads (A) and muscle forces (B) during upslope (dark) and downslope (light) walking. Asterisks (*) indicate a difference between upslope and downslope conditions ($p < 0.05$). AP = anteroposterior; ML = mediolateral.

level-walking data (0.98 m/s and 1.24 m/s respectively; (Hendershot et al., 2018)). When comparing the present results to level walking at a similar speed (1.0 m/s), peak AP and ML shear were larger during downslope walking, but compression, and global and local muscle forces, were similar compared to prior reports (Hendershot et al., 2018). Unlike upslope walking, larger spinal loads (vs. level walking) were only present when comparing walking at similar speeds, suggesting upslope walking is more demanding on the low back than downslope walking. This is further evidenced by the larger ML shear and local forces noted in upslope vs. downslope walking ($p < 0.01$, Fig. 2). Though it is possible that sloped vs. level walking imparts additional demands necessitating co-activations of trunk muscles that are not accounted within this particular modeling approach, therefore underestimating spinal loads.

There are several limitations that should be considered when interpreting these results. The data obtained for individuals with TFA in this study were young, active military personnel with traumatic injuries. The injuries were not characterizable to other amputation etiologies (e.g., vascular, neurological). Additionally, self-selected walking speeds were quite variable, with coefficients of variation of 13.2% and 20.4% for upslope and downslope walking, respectively, which may have influenced trunk muscle forces and spinal loads. However, post-hoc correlation analyses revealed only weak ($0.01 < R < 0.19$) correlations between walking speed and the spinal loads and muscle forces in the present study. In regards to the model, it assumes individuals with LLA respond the same as those without LLA, which is not necessarily true, particularly with respect to trunk muscle co-activations Future work should incorporate electromyography and/or a stability requirement (El Quaaid et al., 2009) to account for these co-activations that may not be accurately represented with the current model. Additionally, the retrospective nature of the study presents a number of limitations. The presence of LBP is a potential confounding factor which cannot be accounted for as we cannot with certainty confirm whether any of the subjects had LBP at the time of testing. Prior work has observed alterations in trunk and pelvis kinematics in persons with LLA with and without LBP (Fatone et al., 2016; Morgenroth et al., 2010), though it is unclear if these changes are independently related to LBP (Fatone et al., 2016). The lack of a control cohort (i.e., uninjured individuals) also limits the scope of the present study. While inferences are made from prior work, these comparisons warrant consideration due to differences in experimental design (e.g., subject population, walking speed, and inclination magnitudes). Future studies should address this with a more robust, cross-sectional study investigating sloped and level walking in both persons with TFA and uninjured populations standardizing the experimental setup. Such investigations should also include kinetic (i.e., force platform) data which were not included in the present study.

In summary, data reported here provide additional insights into the physical demands associated with ADLs among persons with TFA. Specific adaptations in the trunk and pelvis, likely due to the loss of joints and muscles of the affected limb, ultimately induce additional mechanical consequences within the low back (i.e., larger spinal loads). Thus, repeated exposures to these larger-than-normal loads can contribute to an increased risk of low back injury. In continuation of recent work (Hendershot et al., 2018; Shojaei et al., 2019; Shojaei et al., 2016), additional work is still needed to investigate other ADLs (e.g., stair climbing).

Declaration of Competing Interest

We declare that all authors have no financial or personal relationships with other persons or organizations that might inappropriately influence our work presented therein.

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