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Persons with unilateral transfemoral amputation experience larger spinal loads during level-ground walking compared to able-bodied individuals

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Abstract

Background—Persons with lower limb amputation walk with increased and asymmetric trunk motion; a characteristic that is likely to impose distinct demands on trunk muscles to maintain equilibrium and stability of the spine. However, trunk muscle responses to such changes in net mechanical demands, and the resultant effects on spinal loads, have yet to be determined in this population.

Methods—Building on a prior study, trunk and pelvic kinematics collected during level-ground walking from 40 males (20 with unilateral transfemoral amputation and 20 matched controls) were used as inputs to a kinematics-driven, nonlinear finite element model of the lower back to estimate forces in 10 global (attached to thorax) and 46 local (attached to lumbar vertebrae) trunk muscles, as well as compression, lateral, and antero-posterior shear forces at all spinal levels.

Findings—Trunk muscle force and spinal load maxima corresponded with heel strike and toe off events, and among persons with amputation, were respectively 10–40% and 17–95% larger during intact vs. prosthetic stance, as well as 6–80% and 26–60% larger during intact stance relative to controls.

Interpretation—During gait, larger spinal loads with transfemoral amputation appear to be the result of a complex pattern of trunk muscle recruitment, particularly involving co-activation of antagonistic muscles during intact limb stance; a period when these individuals are confident and likely to use the trunk to assist with forward progression. Given the repetitive nature of walking,

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repeated exposure to such elevated loading likely increases the risk for low back pain in this population.

Keywords

Amputation; Gait; Muscle forces; Spinal loads; Low back pain

1. INTRODUCTION

The prevalence of low back pain (LBP) is considerably higher in persons with lower limb amputation (LLA) compared with able-bodied individuals (Friberg, 1984; Sherman, 1989; Sherman et al., 1997; Smith et al., 1999). As a secondary health-related concern, LBP is suggested to be the most important condition that adversely affects the physical performance and quality of life in persons with LLA (Ehde et al., 2001; Taghipour et al., 2009). Given the projected increase in the number of people with LLA (Ziegler-Graham et al., 2008), it is important to investigate the underlying mechanism(s) responsible for the elevated prevalence of LBP in this cohort (Reiber et al., 2010; Devan et al., 2014).

Considering spine biomechanics, spinal loads are the resultant of interactions between internal tissue forces (primarily from muscles) and net mechanical demands of a given activity on the lower back (Cholewicki and McGill, 1996; Calisse et al., 1999; Arjmand and Shirazi-Adl 2005; Adams et al., 2007; McGill et al., 2014). During gait, increased and asymmetric trunk motion following LLA has been reported to impose higher net mechanical demands on the lower back (Cappozzo and Gazzani, 1982; Hendershot and Wolf, 2014). Such an increase in net mechanical demand of a common daily activity, like walking, would require larger responses from internal trunk tissues to assure equilibrium¹ and stability² of the spine, hence leading to larger spinal loads that would presumably increase the risk for LBP due to the repetitive nature of such activities (Adams et al., 2007).

There is limited information in the literature related to internal trunk tissue responses and resultant spinal loads during walking (Cappozzo et al., 1982; Cappozzo, 1983; Cappozzo, 1983; Khoo et al., 1995; Cheng et al., 1998; Callaghan et al., 1999; Yoder et al., 2015). All but two of these few earlier studies included relatively small sample sizes of able-bodied male participants and have reported spinal loads at either the L4–L5 or L5–S1 discs. The estimated pattern of spinal loads in these studies included symmetric local maxima occurring around heel strike and toe off within the gait cycle, with values ranging between 1.2 to 3.0 times body weight. The other two studies regarding internal tissue responses and resultant spinal loads during walking also included persons with LLA (Cappozzo and Gazzani, 1982; Yoder et al., 2015). Using kinematics data obtained from two subjects (one with transfemoral amputation and one with knee ankylosis), Cappozzo and Gazzani (1982) used a rigid link-segment model of the whole body to obtain mechanical demands of walking on the lower back. A simple muscle model was then used to calculate internal tissue responses and the resultant spinal loads. Contrary to the patterns of spinal loads observed in able-bodied individuals, the occurrence of local maxima among persons with LLA did not

¹The balance of internal tissue forces and net mechanical demand of activity

²The ability to return to the equilibrium condition after perturbation

have a symmetric pattern. Rather, the maximum compression forces were larger at the instance of prosthetic vs. intact toe off (2–3.0 vs. 1.0 times body weight). Similar differences in patterns of trunk muscular responses during walking, and the resultant effect on spinal loads (but at much higher magnitudes), between persons with and without transtibial LLA have been recently reported by Yoder et al. (2015). Although these earlier studies highlight the impact of altered and asymmetric gait on loads experienced in the lower back, they were limited to small samples and/or a relatively simple biomechanical model of the lower back.

Using a larger sample size, along with a biomechanical model of the lower back with more bio-fidelity, the objective of this study was to investigate the differences in internal tissue responses, specifically muscle forces, and resultant spinal loads during level-ground walking between individuals with (n=20) and without (n=20) unilateral LLA. Considering that alterations in trunk motion following amputation impose higher (and asymmetric) net mechanical demands on the lower back (Cappozzo and Gazzani, 1982; Hendershot and Wolf, 2014), it was hypothesized that compared to able-bodied individuals, persons with LLA will require larger muscle forces in the lower back to overcome the net mechanical demands of walking while maintaining spinal stability and equilibrium. Such increases in trunk muscle forces would, in turn, result in larger spinal loads.

2. METHODS

2.1 Experimental study

Kinematic data collected in an earlier study were used in these analyses (Hendershot and Wolf, 2014). Briefly, full-body kinematics from 20 males with transfemoral amputation and 20 male able-bodied controls (Table-1) were collected using a 23-camera motion capture system during level-ground walking across a 15 m level walkway at a self-selected speed (mean ≈ 1.35 m/s in both groups). Here, kinematic data of interest included three-dimensional pelvic and thorax motions that were collected by tracking markers positioned in the mid-sagittal plane over the S1, T10, and C7 spinous processes, sternal notch, and xiphoid; and bilaterally over the acromion, ASIS, and PSIS. All amputations were a consequence of traumatic injuries with a mean (standard deviation) duration of 3.1 (1.4) years since amputation. Main inclusion criteria were: (1) unilateral transfemoral amputation with no contralateral functional impairments, (2) daily use of a prosthetic device (1 year post-amputation), (3) no use of an upper-extremity assistive device (e.g., cane, crutches, walker), and (4) having no other musculoskeletal or neurologic problem, except amputation, that may affect gait results. Details of inclusion and exclusion criteria and other experimental methodology can be found in Hendershot and Wolf (2014). This retrospective study was approved by Institutional Review Boards of both University of Kentucky and Walter Reed National Military Medical Center.

2.2 Modeling study

The biomechanical model used to estimate trunk muscle responses and resultant spinal loads included a non-linear finite element (FE) model of the spine that estimated the required muscle forces to complete the activity using an optimization-based iterative procedure (Arjmand and Shirazi-Adl, 2005; Arjmand and Shirazi-Adl, 2006; Bazrgari et al., 2007;

Bazrgari et al., 2008a; Bazrgari et al., 2009b; Arjmand et al., 2010). In this model, muscle forces are estimated such that equilibrium equations are satisfied across the entire lumbar spine. The finite element model included a sagittally symmetric thorax-pelvis model of the spine composed of six non-linear flexible beam elements and six rigid elements (Figure 1) (Arjmand and Shirazi-Adl, 2005; Bazrgari et al., 2008b). The six rigid elements represented the thorax, and each of the lumbar vertebrae from L1 to L5, while the six flexible beam elements characterized the nonlinear stiffness of each lumbar motion segment (i.e. intervertebral discs and ligaments) between the T12 and S1 vertebrae. Stiffness of lumbar motions segments was defined using nonlinear axial compression-strain relationships along with moment-rotation relationships in sagittal/coronal/transverse planes that were obtained from earlier numerical and experimental studies of lumbar spine motion segments (Yamamoto et al., 1989; Oxland et al., 1992; Shirazi-Adl et al., 2002). Upper-body mass and mass moments of inertia were distributed along the spine according to reported ratios (Zatsiorsky and Seluyanov, 1983; De Leva, 1996; Pearsall et al., 1996). Inter-segmental damping with properties defined based on earlier experimental studies were also considered using connector elements (Markolf, 1970; Kasra et al., 1992). The muscle architecture in the biomechanical model included 56 muscles (Fig. 1); 46 muscles connecting lumbar vertebrae to the pelvis (i.e., local muscles) and 10 muscles connecting thoracic spine/rib cage to the pelvis (i.e., global muscles; Arjmand and Shirazi-Adl, 2005; Arjmand and Shirazi-Adl, 2006; Bazrgari et al., 2008a; Bazrgari et al., 2008b).

To determine the required muscle forces for satisfaction of equilibrium across the entire lumbar spine, segmental kinematics in the lumbar region were required. Since only kinematics of the thorax and the pelvis were available from the experimental measurements, a heuristic optimization procedure (Figure 2) was used in the biomechanical model to determine a set of segmental kinematics in the lumbar region (i.e., from L1 to L5) such that the corresponding set of estimated muscles forces minimized a cost function (Shojaei et al., 2015). The cost function used for this heuristic optimization procedure was the sum of squared muscle stress across all lower back muscles. Specifically for each time step during the analysis, a set of possible segmental kinematics in the lumbar region that was within the reported range of motion of lumbar motion segments was initially prescribed on the FE model, and the equations of motion were solved using an implicit integration algorithm inside an FE software (ABAQUS, Version 6.13, Dassault Systemes Simulia, Providence, RI). The outputs of equations of motion were three-dimensional moments at each spinal level, from T12 to L5, that were to be balanced by muscles attached to these same spinal levels. Because the number of attached muscles to these levels (i.e., 10 muscles in each level from T12 to the L4 and 6 muscles at L5) was more than the number of equilibrium equations (i.e., three at each vertebra), a local optimization problem had to be solved for each level to obtain a set of muscle forces that minimize the aforementioned cost function only at that specific level (Arjmand and Shirazi-Adl, 2006). To account for the non-linear mechanical response of passive elements (i.e., intervertebral discs and ligaments) to compression, the estimated portion of compression forces generated by muscles were applied to the spine model and the procedure was repeated until two sequent sets of estimated muscle forces were approximately equal (less than 1% different). The local optimization procedures were performed using the Lagrange Multiplier Method. The above procedure was repeated inside

the heuristic optimization for as many possible sets of segmental kinematics, determined using a genetic algorithm, until a set of segmental kinematics was obtained that meets the optimization criterion. In summary, the global optimization is conducted to determine a set of lumbar segmental kinematics that is associated with minimal sum of squared stress across the entire lower back muscles. Meanwhile, for each possible set of lumbar segmental rotations within the global optimization procedure, a local optimization is conducted for the redundant equilibrium problem at each spinal level to determine the set of muscle forces that optimize the cost function of the local optimization at that specific level (Shojaei et al., 2015). The muscle forces associated with the optimal local kinematics (i.e., the output of the global optimization procedure) were then used to estimate spinal loads at all lumbar levels. These spinal loads included compression forces, along with anterior-posterior and medio-lateral components of the shear forces, relative to the mid-plane of the intervertebral disc and at each lumbar level. The heuristic optimization procedure was developed in Matlab (The MathWorks Inc., Natick, MA, USA, version 7.13).

2.3 Statistical analyses

Rather than comparing the estimated forces in all 56 muscles between the two groups, the summation of estimated forces in the respective global and local muscles (hereafter referred to as equivalent global and local muscle forces) were used for statistical analyses. Similarly, rather than comparing spinal loads at each level, levels with the highest spinal loads (i.e., L4–L5 or L5–S1 for compression forces and L5–S1 for shear forces) were considered for subsequent statistical analyses. For each outcome measure, local maxima were extracted from the stance phase of each limb, resulting in the following values: 1) two peaks in equivalent global muscle forces (Fig. 3; Peak-1 at heel strike of the ipsilateral limb and Peak-2 at toe off of the contralateral limb), 2) two peaks in equivalent local muscle forces (Fig. 3; Peak-1 at heel strike of the ipsilateral limb and Peak-2 at toe off of the contralateral limb), 3) two peaks in estimated compression forces (Fig. 4; Peak-1 at heel strike of the ipsilateral limb and Peak-2 at toe off of the contralateral limb), and 4) one peak in each of the lateral (Fig. 5; at toe off of the contralateral limb), anterior (Fig. 5; at toe off of the contralateral limb), and posterior shear forces (Fig. 5; at heel strike of the ipsilateral limb). It is of note that the gait cycle was defined from right heel strike to subsequent right heel strike for controls, and from heel strike of the intact limb to subsequent intact heel strike for persons with LLA. Prior to statistical analyses, all maxima were normalized with respect to total body mass. Since there were no significant differences ($p > 0.21$ from paired t-tests) in any of our outcome measures between the right and left limbs among controls (as expected due to symmetry), one-way analyses of variance (ANOVAs), instead of two-way ANOVAs (Hendershot and Wolf, 2014), were used to compare outcome measures between three groups: 1) intact limb stance of persons with LLA, 2) prosthetic limb stance of persons with LLA, and 3) mean of left and right limb stance of controls. Significant ANOVAs were followed by post hoc analyses (Tukey). In all statistical analyses, significance was concluded when $p < 0.05$.

3. RESULTS

Mean equivalent global and local muscle forces during a complete gait cycle for controls and persons with LLA are depicted in Figure 3. Equivalent global muscle forces were 2.6 N/kg larger ($p = 0.048$) at ipsilateral heel strike during intact vs. prosthetic limb stance among persons with LLA (Table 2); and 2.7 N/kg larger ($p = .014$) at ipsilateral heel strike during intact limb stance in persons with LLA vs. controls. There were no significant differences ($p > 0.38$) in equivalent local muscle forces at the instant of ipsilateral heel strike. At toe off of the contralateral limb, equivalent global muscle forces were 3.6 N/kg larger ($p = 0.009$) in intact vs. prosthetic limb stance among persons with LLA; at this instant, equivalent global muscle forces were also 5.6 N/kg larger ($p < 0.001$) in intact limb stance among persons with LLA vs. controls, but similar ($p = 0.14$) between stance phases of the prosthetic and control limbs. While there were no significant differences ($p = 0.29$) in equivalent local muscle forces between the stance phase of intact and prosthetic limbs of persons with LLA at the instant of toe off of the contralateral limb, they were, respectively, 2.5 N/kg ($p < 0.001$) and 1.5 N/kg ($p = 0.041$) larger than the corresponding values in controls.

Mean compression forces at the L4–L5 and L5–S1 intervertebral disc during a complete gait cycle for controls and persons with LLA are depicted in Figure 4. At the instant of ipsilateral heel strike, peak compression forces were 3.4 N/kg larger ($p = 0.039$) during stance of the intact vs. prosthetic limb among persons with LLA; at this instant, peak compression forces were also 4.8 N/kg larger ($p < 0.001$) during stance of the intact limb of persons with LLA vs. the corresponding value in controls (Table 2). Peak compression forces at toe off of the contralateral limb were similar ($p = 0.09$) between stance of the intact and prosthetic limbs among persons with LLA, but were 8.6 (4.7) N/kg larger ($p < 0.001$ ($p = 0.002$)) during intact (prosthetic) limb stance of persons with LLA vs. the corresponding value in controls.

Mean values for shear forces at the L5–S1 intervertebral disc during a complete gait cycle for controls and persons with LLA are depicted in Figure 5. In the lateral direction, peak shear forces were 4.3 N/kg larger ($p < 0.001$) in the stance phase of the intact vs. prosthetic limb among persons with LLA (Table 2). These peak shear forces were also 3.3 N/kg larger ($p = 0.015$) during stance of the intact limb of persons with LLA vs. the corresponding value in controls; there were no significant differences in lateral direction peak shear force ($p = 0.33$) between stance phases of prosthetic and control limbs. In the posterior direction, peak shear forces among controls were respectively 1.3 and 1.8 N/kg larger (both $p < 0.001$) than the corresponding values during intact and prosthetic stance among persons with LLA. Peak posterior shear forces were similar ($p = 0.08$) between intact and prosthetic stance among persons with LLA. In the anterior direction, peak shear forces were 1.4 N/kg larger ($p < 0.001$) in the stance phase of the intact vs. prosthetic limb among persons with LLA; these peak shear forces were also 1.8 N/kg larger ($p < 0.001$) during stance of the intact limb vs. the corresponding value in controls.

4. DISCUSSION

In this study, trunk muscle responses to walking demands and the resultant spinal loads were estimated in individuals with and without unilateral LLA. It was hypothesized that individuals with LLA would require larger muscle forces to overcome the net mechanical demands of walking while maintaining spinal equilibrium and stability, which would in turn result in larger spinal loads compared to individuals without amputation. The results obtained through computational simulations and subsequent statistical analyses confirmed our hypothesis. Higher trunk muscle forces and larger spinal loads on the lower back of individuals with unilateral LLA during walking may be, in part, responsible for the reported higher prevalence of LBP among persons with vs. without LLA.

The local maxima for muscle forces and resultant spinal loads occurred at the instants of heel strike and toe off within the gait cycle. These time points also happen to correspond with instances of large axial twist of the trunk (i.e., heel strike) and asymmetric trunk posture (i.e., toe off, where there were relatively large motions in all three planes (Hendershot and Wolf, 2014)). In addition to individual muscle responses, co-activations of antagonistic muscles were needed under such trunk motions to assure spine equilibrium in three-dimensional space. The effects of such an increased and asymmetric motion on muscle forces is more evident when comparing the kinematics and associated muscle forces in the stance phase of intact and prosthetic limbs among individuals with LLA. The complex orientation of the trunk at the instance of heel strike and toe off were more pronounced during stance phase of the intact limb of persons with LLA that resulted in much larger muscle forces during stance of the intact vs. prosthetic limb. Such an effect may also be a result of proximal compensations to assist with toe clearance (Michaud et al., 2000), or simply because these individuals feel more confident during intact (vs. prosthetic) stance to advance their center of mass (i.e., “fear” of fully loading the prosthetic limb).

The sum of forces in global muscles (i.e., equivalent global muscle force) during the gait cycle had the same order of magnitude of the sum of forces in the local muscles (i.e., equivalent local muscle force; Fig. 3). It should be mentioned, however, global muscles were the primary responders to activity demands during the first iteration of muscle force calculations in our model (i.e., the local loop in Fig. 2). As the effects of such global muscle forces were applied into the model, during the subsequent iterations, local muscles became activated to prevent buckling of the spine under the penalties of global muscle forces. If the summation of forces in global and local muscles is assumed to represent the required energy to respectively equilibrate and stabilize the spine, our results suggest that relatively equal amounts of energy were consumed by global and local muscles to provide equilibrium and stability to the spine during walking. However, with such an assumption, it seems that overcoming the equilibrium demands of walking impact the spinal loads of individuals with LLA more than overcoming its segmental stability demands when compared with able-bodied individuals. This observation is reflected in cumulative differences in mean equivalent global muscle forces (i.e., assumed to represent differences in equilibrium demands) between persons with and without LLA that were 955 N larger than the cumulative differences in mean equivalent local muscle forces (i.e., assumed to represent differences in stability demands) between the same two groups during one complete gait

cycle. We should, however, also emphasize that such interpretation is limited to assumptions made in our optimization-based method for estimation of muscle responses to activity demand(s) and would require verification; for example, qualitatively via measurement of muscle activity to determine relative muscle on/off timing. A stabilizing response from local muscles as suggested here should occur sooner than equilibrating response from global muscles.

The estimated spinal loads for controls were in agreement (in terms of patterns and magnitudes) with those obtained in earlier studies (Cappozzo, 1983; Khoo et al., 1995; Cheng et al., 1998; Callaghan et al., 1999; Yoder et al., 2015). Depending on walking speed, the reported values of maximum compression force at the lower spinal level ranged between 1.0 to 2.95 times body weight for walking speeds ranging from 0.9 to 2.2 m/s (Table 3). The mean maximum compression force from these studies, along with average walking speed, were respectively ~ 1.94 times body weight at 1.4 m/s, which are comparable with our estimations of a maximum spinal load of ~ 1.85 times body weight for an average walking speed of ~ 1.35 m/s. Maxima in estimated compression forces in this study occurred around heel strike and toe off instances within the gait cycle, which are also consistent with reported timing of maximum compression forces in earlier studies: around toe off (Callaghan et al., 1999), within a short time interval around toe off (Cappozzo, 1983), right after heel strike and before complete toe off (Cheng et al., 1998), and around 20% and 80% of the gait cycle (Khoo et al., 1995).

The results obtained from individuals with unilateral LLA in this study were also consistent in pattern and magnitude with those reported by Cappozzo and Gazzani (Cappozzo and Gazzani, 1982). This earlier study reported spinal loads for two subjects (i.e., one with transfemoral amputation and one with knee ankylosis) during level-ground walking. The reported maxima of estimated compression forces for the person with LLA ranged from 2 to 3 times body weight for walking speeds between 1.0 m/s and 1.5 m/s (Table 3), which is consistent with the range of maxima of estimated compression forces in this study (~ 2 to 2.6 body weight). In both studies, the maximum compression forces occurred during intact limb stance at the instance of prosthetic toe off. In a more recent study (Yoder et al 2015), much smaller maxima (i.e., $\sim$ body weight) have been reported for maximum spinal loads among persons with transtibial LLA; though smaller maxima could be due, in part, to the relatively slower walking speed and/or more distal amputation.

The failure threshold for vertebral endplates under static loading has been reported to be ~ 5.2 (1.8) kN (Adams et al., 2007). Comparing with such a threshold, estimated spinal loads of 2.6 and 1.85 times body weight for individuals with and without LLA, respectively, may not appear harmful; however, since walking is a highly repetitive task, the cumulative effect and fatigue failure should be considered when assessing injury risk. In particular, the failure threshold can sharply decrease if the number of loading cycles increases (Adams et al., 2007); for example, compressive injury threshold of endplate fracture has been reported to decrease 30% (50%) under 10 (5000) loading cycle when compared with static loading (Brinckmann et al., 1988). Given that 5000 walking steps is easily achievable during a typical day, peak spinal loads experienced by persons with LLA (~ 2 kN for a person with

body mass of 80 kg) become of the same order of injury threshold (i.e. ~ 2.6 kN) and thus, may impose a significant risk for fatigue failure of spinal tissues.

The sample of persons with LLA in this study included young and physically fit members of the military with transfemoral amputations resulting from traumatic injuries. Thus, the results cannot be generalized to groups with other levels or etiologies of amputation. This cross sectional study also does not provide any information about lower back biomechanics in these individuals before the amputations, and history of LBP was not controlled in the participants, though those with current LBP were excluded from the study. Although we accounted for individual differences in trunk inertial properties in the non-linear FE model of spine, we used the same passive tissue properties for all subjects since we had no access to the subject-specific behavior of such tissues (i.e., ligaments, intervertebral discs, passive behavior of muscles and bony structures) for these participants. Furthermore, because motions of the arms were not measured during the original data collection, arms were simulated as point masses that were rigidly connected to the trunk at shoulder height. Therefore, our results do not include the mechanical impact of relative motion of arms on estimated trunk muscle forces and spinal loads. Same heights were considered in the spine model for all subjects, since stature was not significantly different between groups. The precision of model estimations of trunk muscle forces and spinal loads also depends on the accuracy and reliability of measured kinematics. We have performed several intra-lab reliability analyses and found minimal differences in each plane. Across 10 participants (each with 3 distinct sessions), standard errors of measurement for trunk-pelvic kinematics were 1.2, 0.9, and 1.4° for the sagittal, frontal, and transverse planes, respectively. Fidelity of our model-based estimations has been demonstrated in both static (Arjmand and Shirazi-Adl 2005; Arjmand and Shirazi-Adl 2006; Arjmand et al., 2010) and dynamic (Bazrgari et al., 2007; Bazrgari et al., 2008a) situations. More specifically, model estimations have good-excellent correlations with measured trunk muscle activity ($R>0.7$) (Bazrgari et al., 2009a), with measured ground reaction forces ($R>0.8$) (Bazrgari et al., 2008b), and with measured trunk resistance to sudden loading ($R>0.8$ unpublished data from (Bazrgari et al., 2011)). Finally, while a one-way ANOVA was used to compare the outcome measures between the three groups, the measures related to the intact and prosthetic limb stances of person with LLA were, in fact, repeated measures, statistically.

5. CONCLUSION

Asymmetric and larger trunk motion of individuals with LLA during walking were found to be associated with 22% and 63% increases in the respective local and global trunk muscle forces compared to controls. As a result, such neuromuscular adaptations during level ground walking were further found to be associated with substantial increases in spinal loads (i.e., compression: 39%, lateral shear: 60%, and anterior shear: 42%). Considering the effects of cyclic loading due to the large number of daily steps (Buis et al., 2014), repeated exposures to the peak spinal loads experienced by persons with LLA (i.e., 2.6 body weight) potentially pose a high risk for fatigue failure of spinal tissues. It is therefore imperative to investigate whether those with LLA consistently experience higher levels of spinal loads during other important activities of daily living (e.g., ascending and descending ramps or stairs) as a result of an alteration in internal tissue responses to the demands of these

activities. Such knowledge can inform future development of effective clinical programs (e.g., neuromuscular training aimed at minimizing trunk motion/acceleration while also reducing trunk muscle co-activation) aimed at reducing the risk for developing LBP via management of spinal loads during daily activities.

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References

- Adams, MA.; Burton, AK.; Dolan, P.; Bogduk, N. The biomechanics of back pain. Churchill Livingstone; 2007.
- Arjmand N, Gagnon D, Plamondon A, Shirazi-Adl A, Larivière C. A comparative study of two trunk biomechanical models under symmetric and asymmetric loadings. *Journal of biomechanics*. 2010; 43(3):485–491. [PubMed: 19880122]
- Arjmand N, Shirazi-Adl A. Biomechanics of changes in lumbar posture in static lifting. *Spine*. 2005; 30(23):2637–2648. [PubMed: 16319750]
- Arjmand N, Shirazi-Adl A. Sensitivity of kinematics-based model predictions to optimization criteria in static lifting tasks. *Medical engineering & physics*. 2006; 28(6):504–514. [PubMed: 16288897]
- Bazrgari B, Nussbaum MA, Madigan ML, Shirazi-Adl A. Soft tissue wobbling affects trunk dynamic response in sudden perturbations. *Journal of biomechanics*. 2011; 44(3):547–551. [PubMed: 20888563]
- Bazrgari B, Shirazi-Adl A, Arjmand N. Analysis of squat and stoop dynamic liftings: muscle forces and internal spinal loads. *European Spine Journal*. 2007; 16(5):687–699. [PubMed: 17103232]
- Bazrgari B, Shirazi-Adl A, Kusra M. Computation of trunk muscle forces, spinal loads and stability in whole-body vibration. *Journal of Sound and Vibration*. 2008a; 318(4):1334–1347.
- Bazrgari B, Shirazi-Adl A, Larivière C. Trunk response analysis under sudden forward perturbations using a kinematics-driven model. *Journal of Biomechanics*. 2009a; 42(9):1193–1200. [PubMed: 19375707]
- Bazrgari B, Shirazi-Adl A, Parnianpour M. Transient analysis of trunk response in sudden release loading using kinematics-driven finite element model. *Clinical Biomechanics*. 2009b; 24(4):341–347. [PubMed: 19285367]
- Bazrgari B, Shirazi-Adl A, Trottier M, Mathieu P. Computation of trunk equilibrium and stability in free flexion–extension movements at different velocities. *Journal of biomechanics*. 2008b; 41(2): 412–421. [PubMed: 17897654]
- Brinckmann P, Biggemann M, Hilweg D. Fatigue fracture of human lumbar vertebrae. *Clinical biomechanics*. 1988; 3:i-S23.
- Calisse J, Rohlmann A, Bergmann G. Estimation of trunk muscle forces using the finite element method and in vivo loads measured by telemeterized internal spinal fixation devices. *Journal of biomechanics*. 1999; 32(7):727–731. [PubMed: 10400361]
- Callaghan JP, Patla AE, McGill SM. Low back three-dimensional joint forces, kinematics, and kinetics during walking. *Clinical Biomechanics*. 1999; 14(3):203–216. [PubMed: 10619108]
- Cappozzo A. Compressive loads in the lumbar vertebral column during normal level walking. *Journal of orthopaedic research*. 1983; 1(3):292–301. [PubMed: 6481513]
- Cappozzo A. The forces and couples in the human trunk during level walking. *Journal of biomechanics*. 1983; 16(4):265–277. [PubMed: 6863342]

- Cappozzo A, Figura F, Gazzani F, Leo T, Marchetti M. Angular displacements in the upper body of AK amputees during level walking. *Prosthetics and orthotics international*. 1982; 6(3):131–138. [PubMed: 7155809]
- Cappozzo, A.; Gazzani, F. *Biomechanics: Principles and Applications*. Springer; 1982. Spinal loading during abnormal walking; p. 141-148.
- Cheng K, Chen HH, Chen GS, Lee SJ. Influences of walking speed change on the lumbosacral joint force distribution. *Bio-medical materials and engineering*. 1998; 8(3):155–165. [PubMed: 10065882]
- Cholewicki J, McGill SM. Mechanical stability of the *in vivo* lumbar spine: implications for injury and chronic low back pain. *Clinical Biomechanics*. 1996; 11(1):1–15. [PubMed: 11415593]
- De Leva P. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of biomechanics*. 1996; 29(9):1223–1230. [PubMed: 8872282]
- Devan H, Carman AB, Hendrick PA, Ribeiro DC, Hale LA. Perceptions of low back pain in people with lower limb amputation: a focus group study. *Disability & Rehabilitation*. 2014; (0):1–11.
- Ehde DM, Smith DG, Czerniecki JM, Campbell KM, Malchow DM, Robinson LR. Back pain as a secondary disability in persons with lower limb amputations. *Archives of physical medicine and rehabilitation*. 2001; 82(6):731–734. [PubMed: 11387575]
- Friberg O. Biomechanical significance of the correct length of lower limb prostheses: a clinical and radiological study. *Prosthetics and orthotics international*. 1984; 8(3):124–129. [PubMed: 6240634]
- Hendershot BD, Wolf EJ. Three-dimensional joint reaction forces and moments at the low back during over-ground walking in persons with unilateral lower-extremity amputation. *Clinical Biomechanics*. 2014; 29(3):235–242. [PubMed: 24393361]
- Kasra M, Shirazi-Adl A, Drouin G. Dynamics of human lumbar intervertebral joints: experimental and finite-element investigations. *Spine*. 1992; 17(1):93–102. [PubMed: 1536019]
- Khoo B, Goh J, Bose K. A biomechanical model to determine lumbosacral loads during single stance phase in normal gait. *Medical engineering & physics*. 1995; 17(1):27–35. [PubMed: 7704340]
- Markolf, KL. Stiffness and damping characteristics of the thoracic-lumbar spine. *Proceedings of Workshop on Bioengineering Approaches to the Problems of the Spine*; NIH; 1970.
- McGill SM, Cannon J, Andersen JT. Muscle activity and spine load during pulling exercises: Influence of stable and labile contact surfaces and technique coaching. *Journal of Electromyography and Kinesiology*. 2014
- Michaud SB, Gard SA, Childress DS. A preliminary investigation of pelvic obliquity patterns during gait in persons with transtibial and transfemoral amputation. *Journal of rehabilitation research and development*. 2000; 37(1):1–10. [PubMed: 10847567]
- Oxland TR, Lin RM, Panjabi MM. Three-Dimensional mechanical properties of the thoracolumbar junction. *Journal of orthopaedic research*. 1992; 10(4):573–580. [PubMed: 1613631]
- Pearsall DJ, Reid JG, Livingston LA. Segmental inertial parameters of the human trunk as determined from computed tomography. *Annals of biomedical engineering*. 1996; 24(2):198–210. [PubMed: 8678352]
- Reiber GE, McFarland LV, Hubbard S, Maynard C, Blough DK, Gambel JM, Smith DG. Servicemembers and veterans with major traumatic limb loss from Vietnam war and OIF/OEF conflicts: Survey methods, participants, and summary findings. *Journal of rehabilitation research and development*. 2010; 47(4):275–298. [PubMed: 20803399]
- Sherman RA. Stump and phantom limb pain. *Neurologic clinics*. 1989
- Sherman, RA.; Devor, M.; Heermann-Do, K. *Phantom pain*. Springer; 1997.
- Shirazi-Adl A, Sadouk S, Parnianpour M, Pop D, El-Rich M. Muscle force evaluation and the role of posture in human lumbar spine under compression. *European Spine Journal*. 2002; 11(6):519–526. [PubMed: 12522708]
- Shojaei I, Arjmand N, Bazrgari B. An optimization-based method for prediction of lumbar spine segmental kinematics from the measurements of thorax and pelvic kinematics. *International journal for numerical methods in biomedical engineering*. 2015; 10.1002/cnm.2729

- Smith DG, Ehde DM, Legro MW, Reiber GE, Del Aguila M, Boone DA. Phantom limb, residual limb, and back pain after lower extremity amputations. *Clinical orthopaedics and related research*. 1999; 361:29–38. [PubMed: 10212593]
- Taghipour H, Moharamzad Y, Mafi AR, Amini A, Naghizadeh MM, Soroush MR, Namavari A. Quality of life among veterans with war-related unilateral lower extremity amputation: a long-term survey in a prosthesis center in Iran. *Journal of orthopaedic trauma*. 2009; 23(7):525–530. [PubMed: 19633463]
- Yamamoto I, Panjabi MM, Crisco T, Oxland T. Three-dimensional movements of the whole lumbar spine and lumbosacral joint. *Spine*. 1989; 14(11):1256–1260. [PubMed: 2603060]
- Yoder AJ, Petrella AJ, Silverman AK. Trunk-pelvis motion, joint loads, and muscle forces during walking with a transtibial amputation. *Gait & Posture*. 2015
- Zatsiorsky V, Seluyanov V. The mass and inertia characteristics of the main segments of the human body. *Biomechanics viii-b*. 1983; 56(2):1152–1159.
- Ziegler-Graham K, MacKenzie EJ, Ephraim PL, Travison TG, Brookmeyer R. Estimating the prevalence of limb loss in the United States: 2005 to 2050. *Archives of physical medicine and rehabilitation*. 2008; 89(3):422–429. [PubMed: 18295618]

HIGHLIGHTS

- Persons with lower limb amputation have been reported to walk with large and asymmetric trunk motion
- The recruited patterns of muscle forces to generate such changes in trunk motion following amputation were estimated
- Level over ground walking was found to be associated with ~ 45% larger spinal loads in person with versus without amputation
- Repeated exposure to larger spinal loads during walking may elevate low back pain risk

Figure 1.

Sagittal view of the biomechanical model including FE model of the spine and 56 trunk muscles (dimensions in mm). ICPL: iliocostalis lumborum pars lumborum, ICPT: iliocostalis lumborum pars thoracis, IP: iliopsoas, LGPL: longissimus thoracis pars lumborum, LGPT: longissimus thoracis pars thoracis, MF: multifidus, QL: quadratus lumborum, IO: internal oblique, EO: external oblique and RA: rectus abdominus.

Figure 2.

The process used to estimate muscle forces and spinal loads. Each set of possible segmental kinematics is generated using a genetic algorithm subjected to measured kinematics of thorax and pelvis as well as the reported values of lumbar segments' range of motion. The convergence in the local and global loops are achieved when the changes in respectively sum of estimated muscle forces in two consecutive local iterations and the value of the cost function of the heuristic optimization procedure in two consecutive global iterations are less than 1%.

Figure 3.

Mean equivalent global (top) and local (bottom) muscle forces among the control (CTL) and lower limb amputation (LLA) groups. Data are presented as functions of percent gait cycle; that is, from right (intact) heel strike to subsequent right (intact) heel strike.

Figure 4.

Mean compression forces at mid-plane of the L4-L5 (top) and L5-S1 (bottom) intervertebral discs among the control (CTL) and lower limb amputation (LLA) groups. Data are presented as functions of percent gait cycle; that is, from right (intact) heel strike to subsequent right (intact) heel strike.

Figure 5.

Mean shear forces at the mid-plane of the L5-S1 intervertebral disc in lateral (top) and antero-posterior (bottom) directions among the control (CTL) and lower limb amputation (LLA) groups. Positive shear force in lateral direction indicates force toward the right (intact) limb for controls (LLA) and positive shear force in antero-posterior direction indicate anterior direction. Data are presented as functions of percent gait cycle; that is, from right (intact) heel strike to subsequent right (intact) heel strike.

Table 1

Participant characteristics for the control (CTL) and lower limb amputation (LLA) groups (Hendershot and Wolf, 2014).

Variable	CTL (n=20)	LLA (n=20)
Age (year)	28.1 (4.8)	29.20 (6.70)
Stature (cm)	181.00 (6.10)	176.20 (6.70)
Body mass (kg)	83.90 (8.60)	80.60 (12.20)

Table 2Mean (SD) estimated maximum muscle forces and resultant spinal loads.

	Control (n=20)	<u>Transfemoral Amputation (n=20)</u>	
Variable		Intact Stance	Prosthetic Stance
<u>MUSCLE FORCES</u>			
Equivalent global – Ipsilateral heel strike (N/kg)	7.7 (2.5)	10.4 (5.0) *	7.8 (3.0) †
Equivalent global – Contralateral toe off (N/kg)	7.0 (2.6)	12.6 (5.2) *	9.0 (4.1) †
Equivalent local – Ipsilateral heel strike (N/kg)	8.4 (2.0)	8.9 (2.1)	8.1 (1.7)
Equivalent local – Contralateral toe off (N/kg)	7.8 (1.4)	10.3 (3.1) *	9.3 (2.3) *
<u>SPINAL LOADS</u>			
Compression – Ipsilateral heel strike (N/kg)	18.2 (3.4)	23.0 (5.8) *	19.6 (4.1) †
Compression – Contralateral toe off (N/kg)	16.8 (3.3)	25.4 (7.0) *	21.5 (4.8) *
Lateral Shear (N/kg)	5.5 (1.1)	8.8 (1.6) *†	4.5 (1.2)
Posterior Shear (N/kg)	3.7 (0.8)	2.4 (0.8) *	1.9 (0.6) *
Anterior Shear (N/kg)	4.2 (1.0)	6.0 (1.1) *	4.6 (0.9) †

* Significant difference relative to control

† Significant difference between intact vs. prosthetic

Table 3
Reported values of maximum compression force $(*(1/\text{body weight}))$ at the lower spinal level.

Study	Walking Speed (m/s)						
	0.90	1.00	1.20	1.35	1.50	1.70	2.20
Typical walking	Current study			1.85			
	Cappozzo, 1983	1.20		1.50		1.90	2.50
	Cheng et al., 1998	2.28	2.53		2.95		
	Khoo et al., 1995		1.71				
	Yoder et al., 2015	1.0					
Atypical walking	Current study			2.60			
	(Cappozzo and Gazzani, 1982) (amputation)	2.00		2.70	3.00		
	(Cappozzo and Gazzani, 1982) (knee ankylosis)	1.80			2.10		
	Yoder et al., 2015	1.0					