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

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 (Devan et al., 2014; Reiber et al., 2010).

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 (Adams et al., 2007; Arjmand and Shirazi-Adl 2005; Calisse et al., 1999; Cholewicki and Mcgill, 1996; 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 equilibrium[1] and stability[2] 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, 1983a, 1983b; Khoo et al., 1995; Cheng et al., 1998; Callaghan et al., 1999; Yoder et al., 2015). All but

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[1] The balance of internal tissue forces and net mechanical demand of activity.
[2] The ability to return to the equilibrium condition after perturbation.

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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 and 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 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.

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)

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, 2006; Bazrgari et al., 2007, 2008a, 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 (Fig. 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, 2006; Bazrgari et al., 2008a,b).

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 (Fig. 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