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Trunk Kinetic Effort during Step Ascent and Descent in Patients with Transtibial Amputation using Angular Momentum Separation

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Abstract

Background—Patients with transtibial amputation adopt trunk movement compensations that alter effort and increase the risk of developing low back pain. However, the effort required to achieve high-demand tasks, such as step ascent and descent, remains unknown.

Methods—Kinematics were collected during bilateral step ascent and descent tasks from two groups: 1) seven patients with unilateral transtibial amputation and 2) seven healthy control subjects. Trunk kinetic effort was quantified using translational and rotational segmental moments (time rate of change of segmental angular momentum). Peak moments during the loading period were compared across limbs and across groups.

Findings—During step ascent, patients with transtibial amputation generated larger sagittal trunk translational moments when leading with the amputated limb compared to the intact limb ($P = 0.01$). The amputation group also generated larger trunk rotational moments in the frontal and transverse planes when leading with either limb compared to the healthy group ($P = 0.01$, $P < 0.01$, respectively). During step descent, the amputation group generated larger trunk translational and rotational moments in all three planes when leading with the intact limb compared to the healthy group ($P < 0.017$).

Interpretation—This investigation identifies how differing trunk movement compensations, identified using the separation of angular momentum, require higher kinetic effort during stepping tasks in patients with transtibial amputation compared to healthy individuals. Compensations that

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Conflict of Interest

None.

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produce identified increased and asymmetric trunk segmental moments, may increase the risk of the development of low back pain in patients with amputation.

Keywords

Transfemoral amputation; high-demand tasks; step ascent/descent; angular momentum; kinetic effort

1. Introduction

Currently in the United States, over 80% of all lower-limb amputations result from neurovascular pathologies (e.g. diabetes mellitus), and this number is increasing due to an aging population and growing prevalence of patients with diabetes mellitus (Dillingham et al., 2002; Davies & Datta, 2003). Patients with dysvascular amputation have additional comorbidities that are associated with poor physical function such as aging, osteoarthritis, obesity and low back pain (LBP) (Ehde et al., 2001; Norvell et al., 2005; Ziegler-Graham et al., 2008; Morgenroth et al., 2011; Rosenberg et al., 2012). Because of these comorbidities, 40–50% of patients with dysvascular transfemoral amputation (TTA) fail to achieve community ambulation one year following amputation (Davies & Datta, 2003), and the majority report difficulty achieving high-demand tasks such as step ambulation (De Laat et al., 2013).

To compensate for the loss of active ankle plantarflexion, patients with TTA adopt exaggerated trunk movements, which assist forward progression in the sagittal plane and help maintain balance/stability in the frontal plane during ambulation (Sagawa et al., 2011). Because the trunk and abdomen account for over a third of the body mass (Winter, 1990), even seemingly small trunk movement compensations can lead to high loads in the low back and increased risk developing LBP (Kumar, 2001; McGill, 2007). In the sagittal plane, patients with amputation use an exaggerated forward trunk lean (Anzel et al., 1994; Goujon-Pillet et al., 2008), and generate large anterior momentum during the pre-swing phase of walking (Gaffney et al., 2016). The anterior position of the body COM relative to the knee requires less ankle propulsion during pre-swing to translate the body COM forward with respect to the stance foot, and reduces quadriceps demand on the amputated limb (Torburn et al., 1990; Anzel et al., 1994; Powers et al., 1998). In the frontal plane, patients with TTA exaggerate the lateral trunk lean toward the amputated limb during weight acceptance (compensated Trendelenburg gait pattern) (Molina-Rueda et al., 2014; Gaffney et al., 2016). Because these patients have similar hip abductor strength in the amputated and intact limbs (Nadollek et al., 2002), the compensated Trendelenburg pattern is likely a strategy used to maintain balance on the amputated limb. Although the development of LBP remains idiopathic, these patterns are likely associated with detrimental low back loading (Devan et al., 2014; Hendershot & Wolf, 2014; Esposito & Wilken, 2014).

A clear understanding of trunk compensations adopted by patients with TTA during high-demand tasks may help identify potential risk factors that can be used to improve the efficacy of movement retraining following amputation. During initial movement retraining, patients with TTA are often instructed to compensate by ascending steps leading with the intact limb and descend steps leading with the amputated limb (Jones et al., 2005, 2006; Schmalz et al., 2007; Alimusaj et al., 2009; Barnett et al., 2014). Step ascent and descent are

common high-demand tasks required for community ambulation that are more difficult for patients with TTA than their healthy counter parts (De Laat et al., 2013). Because step ambulation places higher demand on the musculoskeletal system (higher joint loading and muscle activation) compared to over-ground walking, compensations adopted during these tasks likely increase the risk of sustaining additional disabling comorbidities (McFadyen & Winter, 1988; Nadeau et al., 2003; Protopapadaki et al., 2007; Reeves et al., 2008). These risks likely increase when patients with TTA ambulate steps in a manner different than that taught in movement retraining (e.g. ascend with the amputated limb) because the effect of training alters the motor control and movement strategies adopted (Lay et al., 2002). However, chronic movement compensations adopted by patients with TTA when ambulating steps with either limb are not well-understood (Molina-Rueda et al., 2015). In addition, despite movement retraining inclusion of walking on various surfaces and step negotiation, many biomechanics investigations of movement compensations in patients with TTA focus only on level-ground walking.

The objective of this investigation was to evaluate trunk compensations and the associated kinetic effort needed for patients with unilateral TTA to perform step ascent and descent tasks, and compare to healthy control subjects. We define kinetic effort as the net segment moment calculated through Euler's Laws in rotational form that is created by gravity, intersegmental joint forces, and muscles forces used to generate and arrest segmental angular momentum throughout movement. We compared kinetic effort across groups by identifying translational and rotational trunk moments, which were calculated from segmental momenta. We previously identified differences in the generation and arresting of trunk translational (TAM) and rotational angular momentum (RAM) in patients with unilateral dysvascular TTA during over-ground walking (Gaffney et al., 2016). In the current study, we hypothesized that patients with TTA would require larger trunk kinetic effort when stepping onto the amputated limb compared to the intact limb or healthy controls for both step ascent and descent. By establishing the link between trunk movement compensations and kinetic effort, we can gain additional insight into consequential effects of compensatory movement patterns, and can provide more targeted movement retraining following TTA.

2. Methods

2.1 Participants

Seven male patients with unilateral TTA and seven male healthy control (HC) participants were enrolled (Table 1). For the TTA group, all prostheses were passive with total contact carbon fiber sockets and dynamic elastic response feet. Each participant provided written, informed consent in accordance with the Colorado Multiple Institutional Review Board prior to the start of the experimental session. Each participant visited the laboratory for one data collection session, in which whole body kinematics were collected during step ascent and descent tasks.

2.1 Motion Analysis

Each participant was instrumented with 63 reflective markers to obtain whole-body kinematics during the step ascent and descent tasks. Three-dimensional positions of the

markers with respect to the inertial origin were recorded from eight near-infrared cameras (100 Hz sampling frequency) (Vicon, Centennial, CO). Each participant performed three bilateral step ascent and descent trials onto a 20-cm platform. The TTA group was instructed to perform the ascent and descent tasks leading with the limb instructed during movement retraining (ascent: lead with intact limb; descent: lead with amputated limb (Barnett et al., 2014)), then instructed to perform the tasks leading with the contralateral limb. The HC group was instructed to perform the tasks with the dominant limb (right limb for all participants), then instructed to perform the tasks leading with the contralateral limb. No instructions were provided to the participants regarding the speed at which to complete each task. Data were averaged across the three trials and used for between-limb and between-group comparisons.

2.2 Data Analysis

Marker position data were low-pass filtered with a 4th order Butterworth filter (6 Hz cutoff frequency). A 15-segment subject model was created in Visual 3D (C-Motion, Germantown, MD) (Gaffney et al., 2016). Intact segment inertial parameters were based on regression equations of segment geometry (Dempster, 1955) and inertial parameters of the prosthetic shank (residual limb + prosthetic socket) and prosthetic foot were measured using a reaction board and oscillation method (Smith et al., 2014).

The total segmental angular momentum of the trunk with respect to the leading stance foot is described as:

$${}^I\mathbf{h}_{\text{Foot}} = {}^I\mathbf{h}_{\text{Trunk/Foot}} + {}^I\mathbf{h}_{\text{Trunk}} \quad (1)$$

where ${}^I\mathbf{h}_{\text{Trunk/Foot}}$ is the trunk translational angular momentum (TAM) with respect to the leading stance foot and ${}^I\mathbf{h}_{\text{Foot}}$ is the trunk rotational angular momentum (RAM) about the trunk COM. Based on the principle of angular momentum separation, TAM and RAM can be separated into two independent components (Kadin & Paley, 2011). Trunk TAM with respect to the leading stance foot is described as:

$${}^I\mathbf{h}_{\text{Trunk/Foot}} = \mathbf{r}_{\text{Trunk/Foot}} \times m_{\text{Trunk}} {}^I\mathbf{v}_{\text{Trunk/Foot}} \quad (2)$$

where $\mathbf{r}_{\text{Trunk/Foot}}$ is the position vector of the trunk COM relative to the stance foot COM, ${}^I\mathbf{v}_{\text{Trunk/Foot}}$ is the velocity of the trunk COM relative to the stance foot COM as observed in an inertial reference frame I , and m_{Trunk} is the mass of the trunk (Figure 1). The time derivative of trunk TAM is an expression of Euler's 1st Law in angular momentum form:

$$\frac{{}^I d}{dt} ({}^I\mathbf{h}_{\text{Trunk/Foot}}) = \mathbf{M}_{\text{Trunk/Foot}} - (\mathbf{r}_{\text{Trunk/Foot}} \times m_{\text{Trunk}} {}^I\mathbf{a}_{\text{Foot}}) \quad (3)$$

where $\mathbf{r}_{\text{Trunk/Foot}} \times m_{\text{Trunk}} {}^I \mathbf{a}_{\text{Foot}}$ is the corrective inertial moment of the trunk relative to the stance foot and is required to satisfy Euler's laws when the foot accelerates during the task. The translational trunk segmental moment about the foot, expressed as:

$$\mathbf{M}_{\text{Trunk/Foot}} = \mathbf{r}_{\text{Trunk/Foot}} \times \mathbf{F}_{\text{Trunk}}^{\text{Ext}} \quad (4)$$

where $\mathbf{F}_{\text{Trunk}}^{\text{Ext}}$ is the net of all external forces applied to the trunk due to the force of gravity, intersegmental joint forces, and the forces applied to a segment due to muscle force actuators (Figure 1) (Gaffney et al., 2017).

Trunk RAM is described as:

$${}^I \mathbf{h}_{\text{Trunk}} = \mathbf{I}_{\text{Trunk}} \cdot \boldsymbol{\omega}_{\text{Trunk}} \quad (5)$$

where $\mathbf{I}_{\text{Trunk}}$ and $\boldsymbol{\omega}_{\text{Trunk}}$ are the inertial tensor and angular velocity of the trunk, respectively. The time derivative of trunk RAM is an expression of Euler's 2nd Law:

$$\frac{{}^I d}{dt} ({}^I \mathbf{h}_{\text{Trunk}}) = \mathbf{M}_{\text{Trunk}}^{\text{Ext}} \quad (6)$$

The right hand side of Equation 6 is the rotational trunk moment expressed as:

$$\mathbf{M}_{\text{Trunk}} = \sum_{i=1}^N \mathbf{M}_i^{\text{Ext}} + \sum_{i=1}^N (\mathbf{r}_{i/\text{Trunk}} \times \mathbf{F}_i^{\text{Ext}}) \quad (7)$$

where i is the distal and proximal locations of forces and moments applied to the trunk. $\mathbf{M}_{\text{Trunk}}$ is the total trunk moment that is used to solve for the joint moments by adding the sum of the applied (external) proximal and distal moments and moments about the trunk COM due to intersegmental joint forces (Figure 1) (Gaffney et al., 2017).

Segmental kinetic effort is defined as the net segment moment calculated through Euler's Laws in rotational form that is created by gravity, intersegmental joint forces, and muscles forces that are used to generate and arrest segmental angular momentum throughout movement. Kinetic effort was interpreted based on the relation between the underlying segmental kinetics derived via Euler's Laws in rotational form to joint kinetics calculated using the more common iterative Newton-Euler method via inverse dynamics. Joint kinetics calculated from inverse dynamics are common clinical descriptors of joint demand or effort because they represent the total agonist and antagonist muscle activity spanning a joint (Carollo & Matthews, 2009). Therefore, a change in segmental translational or rotational moments represent a change in kinetic effort, in which the joint kinetics contribute (Gaffney et al., 2017)

To facilitate anatomically planar analyses, all momenta and moment vectors were expressed in a basis with respect to the path of the body COM: $\mathbf{e}_{\text{frontal}}$ (tangent to the horizontal path of the body COM), $\mathbf{e}_{\text{transverse}}$ (opposite direction of the gravity), and $\mathbf{e}_{\text{sagittal}}$ ($\mathbf{e}_{\text{frontal}} \times \mathbf{e}_{\text{transverse}}$).

All trunk translational and rotational moments were normalized by time during the loading period (leading limb foot initial contact (0%) to trailing limb foot initial contact (100%)). The functional sub-phases of the step ascent and descent used for qualitative description of timing were based on the sub-phases defined by Zachazewski et al., (1993) and normalized to the loading period (Figure 2).

2.3 Statistical Analysis

To quantify the effort needed to generate or arrest trunk angular momentum we identified global minima and maxima of the trunk translational and rotational moments in all three planes during the step ascent and descent trials (dependent variables). Three one-way mixed-factor models were used for each dependent variable: *between* subjects (leading with amputated limb vs. HC (leading with dominant (right) limb) and leading with intact vs. HC (leading with dominant (right) limb) and *within* subjects (leading with amputated limb vs. leading with intact limb) while controlling for differences in height and mass (covariates). For the HC group, only trials performed on the dominant (right) limb were used for comparison because no differences existed across limbs. Therefore, we analyzed 12 trials for the TTA group (3 trials, 2 step conditions, 2 limbs) and six trials performed on the dominant (right) limb were used for comparison with the HC group (3 trials, 2 step conditions, 1 limb). When statistically significant differences were found, pairwise comparisons with a Bonferroni adjustment for multiple comparisons were used ($\alpha_B = 0.05/3 = 0.017$). To quantify the amount of change between dependent variables, Hedges' g effect size and bootstrapped 95% confidence intervals were calculated (Hedges, 1981; Hentschke & Stuttgen, 2011; Halsey et al., 2015) and categorized as small effect ($g \leq 0.2$), medium effect ($0.2 < g < 0.8$), or large effect ($g \geq 0.8$). Only confidence intervals that did not cross zero were considered to be statistically significant (Curran-Everett, 2009).

3. Results

3.1 Step Ascent

In the sagittal plane, the peak posterior translational trunk moment (positive) was larger in the TTA group with the amputated limb than the intact limb during vertical thrust of ascent ($P = 0.01$, $g = 1.52$ [1.16–2.64]) (Figure 3).

In the frontal plane, the peak mediolateral translational trunk moment toward the leading stance foot (positive) was larger in the TTA group when leading with the amputated limb compared to the HC group during weight acceptance ($P = 0.01$, $g = 1.8$ [1.17–3.53]) (Figure 3). Peak mediolateral rotational trunk moment toward the leading stance foot (positive) was larger in the TTA group when leading with the intact limb compared to the HC group at the beginning of weight acceptance ($P < 0.01$, $g = 1.56$ [0.89–3.52]) (Figure 4).

In the transverse plane, peak axial translational moment toward the leading stance foot (negative) was larger in the TTA group when leading with either the amputated or intact limb compared to the HC group during weight acceptance ($P < 0.01$, $g = 2.36$ [1.62–5.01]; $P = 0.01$, $g = 1.47$ [1.01–2.94], respectively) (Figure 3). The peak axial rotational moment away from the leading stance foot (positive) was larger in the TTA group when leading with either the amputated or intact limb compared to the HC group during vertical thrust ($P = 0.015$, $g = 1.50$ [1.18–3.46]; $P < 0.01$, $g = 2.52$ [1.85–5.59] respectively) (Figure 4).

3.2 Step Descent

In the sagittal plane, peak anterior (negative) and posterior (positive) translational trunk moments were larger in the TTA group when leading with the intact limb compared to the HC group ($P < 0.01$, $g = 1.83$ [1.48–3.70]; $P = 0.01$, $g = 1.16$ [0.40–3.16], respectively) (Figure 5). Peak anterior rotational trunk moment (negative) was larger in the TTA group when leading with the intact limb than the HC group during weight acceptance ($P = 0.017$, $g = 0.99$ [0.10–3.42]) (Figure 6).

In the frontal plane, peak translational moment away from the leading stance foot (negative) was larger in the TTA group when leading with either the amputated or intact limb compared to the HC group during weight acceptance ($P < 0.01$, $g = 1.70$ [1.14–3.55]; $P < 0.01$, $g = 1.51$ [0.53–4.93], respectively). The peak translational moment toward the leading stance foot (positive) was larger in the TTA group when leading with the intact limb than the HC group at the beginning of forward continuance ($P = 0.01$, $g = 1.16$ [0.40–3.16]). The peak mediolateral rotational trunk moment away from the leading stance foot (negative) was larger in the TTA group when leading with the intact limb compared to the HC group ($P < 0.01$, $g = 3.52$ [1.99–11.86]). The peak mediolateral rotational trunk moment toward the leading stance foot (positive) was larger in the TTA group when stepping onto either the amputated or intact limb compared to the HC group ($P < 0.01$, $g = 2.11$ [1.19–5.07]; $P < 0.01$, $g = 2.06$ [1.77–8.87], respectively) (Figure 6).

In the transverse plane, peak axial rotational trunk moment toward the leading stance foot (negative) was larger in the TTA group when stepping onto either the amputated or intact limb compared to the HC group ($P = 0.01$, $g = 1.13$ [0.01–3.78]; $P = 0.017$, $g = 1.45$ [0.30–4.28], respectively) (Figure 6).

4. Discussion

The purpose of this study was to determine how movement compensations altered the required trunk kinetic effort during step ascent and descent in patients with unilateral transtibial amputation and healthy controls by analysis of translational and rotational segmental moments. Changes in segmental moments are created by simultaneous changes in gravitational moments, joint reaction forces, and joint moments (moments created by muscle forces). We measured differences between groups in trunk translational and rotational segmental moments in three anatomical planes during step ascent and descent tasks across limbs and across groups, which represent differences in the effort required to successfully complete the task. During initial training following amputation, patients are often instructed to ascend steps leading with the intact limb and descend steps leading with the amputated

limb (Jones et al., 2005, 2006; Schmalz et al., 2007; Alimusaj et al., 2009; Barnett et al., 2014). These trained patterns may influence the long-term gait patterns during step ambulation. Our results indicate that patients with amputation adopt different strategies dependent upon which limb is leading during step ambulation compared to able-bodied individuals, which alters the kinetic effort of the trunk required for successful completion of the task.

4.1 Step Ascent

In the sagittal plane, patients with amputation demonstrated a larger posterior trunk translational moment when stepping up with the amputated limb than the intact limb, which may increase the demand on the low back extensor muscles (e.g. multifidus, erector spinae). The increase in posterior translational moment is a result of increased external joint reaction forces applied to the trunk at the low back that originate from the excessive plantarflexor forces from the intact limb (Schmalz et al., 2007). When stepping onto the amputated limb, patients with amputation adopt a ‘hip-extensor dominant’ strategy in the intact limb to elevate the body COM (Powers et al., 1997; Yack et al., 1999; Schmalz et al., 2007). This strategy creates larger anteriorly directed net external joint forces at the hip and low back and posterior motion of the trunk later in the vertical thrust phase (Figure 3), which is consistent with previous findings during amputee gait (Hendershot & Wolf, 2014). A large posterior trunk translational moment when stepping up with the amputated limb makes the peak posterior moment late in the vertical thrust phase (Figure 3a) necessary to arrest the trunk momentum and maintain balance.

Asymmetric low back loading is associated with an increased risk of LBP in the frontal plane (Davis & Marras, 2000); therefore, compensations identified during step ambulation may have potential long-term adverse effects on the low back through repetitive increased and asymmetric loading. Patients with amputation demonstrated compensations in the trunk translational and rotational moments that are likely used for stability (Jones et al., 2005). In addition, patients with amputation use the momentum of the trunk, generated by increasing concentric muscle activation of the intact limb hip abductors (Nadeau et al., 2003), to elevate the pelvis and avoid contact between the amputated swing limb and the step. When stepping up with the amputated limb, the translational trunk moment toward the stance limb was larger in comparison to healthy controls, which is consistent with previous findings of high laterally-directed low back joint reaction shear forces measured in patients with amputation (Hendershot & Wolf, 2014). When stepping onto the intact limb, patients with amputation demonstrated larger rotational trunk moments toward the stance limb early during early weight acceptance. The rotational segment moments are used in iterative Newton-Euler inverse dynamic analyses to calculate joint moments; therefore, the current findings are likely consistent with our previous findings (Murray et al., *In Review*), which found increased lateral bend moments coupled with increased frontal plane trunk displacement directed toward the stance limb during step ambulation in the current amputation group (Figure 4). Increased trunk displacement toward the stance limb is consistent with a compensated Trendelenburg pattern and is hypothesized to improve lateral balance in patients with lower-limb amputation (Michaud et al., 2000; Tura et al., 2010; Molina-Rueda et al., 2014). However, the compensated Trendelenburg pattern increases demand on the

spine and surrounding musculature which may increase the risk of developing LBP (Hendershot et al., 2013).

In the transverse plane, patients with amputation ascend steps with bilateral movement strategies that require increased trunk kinetic effort compared to healthy subjects. Patients with amputation increased axial rotational moments away from the stance limb, which may be a strategy required to arrest momentum. However, the timing of peak axial rotational moment away from the stance limb differs between limbs, indicating limb-dependent strategies. When stepping onto the intact limb, peak axial rotational trunk moment away from the stance foot occurs earlier in the vertical thrust phase compared to the amputated limb, which is likely an aggregate effect of the ‘hiking strategy’ accomplished by increasing the axial rotation of the trunk during early weight acceptance (Figure 4). This strategy is likely required to elevate the body COM onto the step in the absence of the active ankle plantar flexion from the trailing (amputated) limb. However, when stepping onto the amputated limb, the peak axial rotational trunk moment away from the stance limb occurs later in the vertical thrust phase, which is likely result in the delayed and increased ground reaction forces created from the trailing (intact) limb (Schmalz et al., 2007) which results in increased axial rotation of the pelvis when loading the amputated limb (Gaffney et al., 2016).

4.2 Step Descent

In the sagittal plane, patients with amputation demonstrated a higher posterior trunk translational moment when stepping onto the intact limb than the amputated limb which provides insight into the kinetic strategies used to achieve forward progression. This higher translational moment is caused by a higher vertical ground reaction force (GRF) that propagates up the kinetic chain, and is associated with ‘falling’ onto the intact limb with limited control. This strategy is commonly adopted by patients with amputation when stepping onto the intact limb because of reduced ability to actively control the lowering of the body COM with the amputated limb (Schmalz et al., 2007). During weight acceptance, the amputation group had greater trunk rotational moment when stepping onto either the amputated or intact limbs in comparison to the healthy group, which is consistent with a strategy to generate momentum that results in anterior trunk lean strategy over the leading stance limb (Figure 6). This strategy reduces the demand on the trailing limb quadriceps muscles during loading, but may have long-term detrimental effects due to increased demand placed on the trunk and hip extensor musculature (Hendershot & Wolf, 2014).

In the frontal plane, patients with amputation increased trunk rotational moments in the direction toward the stance limb during weight acceptance compared to healthy controls. This is likely a similar strategy to that observed during the step ascent, in which patients with amputation increase the generation of frontal plane momentum of the trunk towards the stance foot during loading (compensated Trendelenburg) (Figure 6). However, in contrast to step ascent, this strategy occurred when leading with either the amputated or intact limb. Patients with amputation may employ this strategy bilaterally during step descent to compensate for hip abductor weakness (Molina-Rueda et al., 2014) and maximize stability during this highly destabilizing task (Michaud et al., 2000; Tura et al., 2010). Although the

compensated Trendelenburg pattern assists with maintaining balance, the excessive and abrupt loading may have negative long-term consequences on the spine and low back musculature.

In the transverse plane, patients with amputation demonstrated 74% (amputated limb) and 42% (intact limb) larger trunk rotational moments toward the stance limb during weight acceptance onto either limb compared to healthy controls. This difference demonstrates that the loading strategies adopted by patients with amputation increase the kinetic effort required for successful completion, regardless of the limb being loaded. When stepping onto the amputated limb, the increased trunk rotational moment is likely an effect of the inability to arrest axial rotation with the prosthetic limb (Figure 6). When stepping onto the intact limb, the increased trunk rotational moment is likely an effect of the increased momentum from 'falling' off of the step and landing abruptly, which causes exaggerated axial rotation of the trunk (Figure 6). Lack of axial control of trunk rotation is frequently linked to the development of LBP (Van Dieën et al., 2003; van den Hoorn et al., 2012), and therefore indicates that these adaptations may have potential long term adverse effects.

4.3 Limitations

Several limitations to this investigation should be considered. First, this investigation included a small sample of individuals with dysvascular TTA, which may limit generalizability to individuals with other types of TTA (e.g. traumatic, oncologic, congenital). Second, the TTA group was not screened for LBP at the time of testing; therefore, we cannot conclude if compensatory movement patterns observed were habitual or a result of LBP. However, we do not expect this to have a confounding effect on the present results because, when asked "Did you experience any pain during that exercise", no participant reported LBP at the time of testing. The reporting occurred following a maximum isometric contraction of the low back muscles, which can induce pain in patients with chronic LBP (Kankaanpää et al., 1998). Third, the length of time using a prosthesis was not included in our analysis, which, in addition to typical comorbidities, may affect the compensations adopted by an individual with amputation. Finally, the step ascent and descent trials did not consist of alternating stairs; therefore, the compensatory movement patterns during each task may not indicate strategies that are required for ascending and descending stairs.

5. Conclusion

This investigation identified the trunk movement compensations that alter effort during step ascent and descent in patients with unilateral transtibial amputation by identifying the translational and rotational trunk moments. Trunk compensations are required to successfully ascend and descend steps without active plantarflexion, but may have long-term adverse effects through the increased and asymmetric demand placed on the low back musculature. It remains unclear what level of trunk movement compensation can be used to compensate for the loss of active ankle plantarflexion without having potential adverse effects through increased demand on the low back.

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Highlights

- Kinetic effort interpreted using segmental moments derived using Euler's Laws.
- Kinetic effort compared between amputees and controls during stepping tasks.
- Movement compensations adopted by transtibial amputees alter kinetic trunk effort.
- High trunk kinetic effort potentially linked to consequential low back loading.

Figure 1.

(a) The trunk translational moment ($\mathbf{M}_{\text{Trunk/Foot}}$) is the total segment moment due to all external forces applied to the trunk $\mathbf{F}_{\text{Trunk}}$ due to the force of gravity, Trunk intersegmental joint forces, and forces due to muscle force actuators with respect to the stance foot ($\mathbf{r}_{\text{Trunk/Foot}}$). (b) The trunk rotational moment ($\mathbf{M}_{\text{Trunk}}$) is the total segment moment about the trunk COM and is dependent on external moments and moments due to intersegmental joint forces. The trunk translational and rotational moments are the time rate of change of the translational and rotational angular momentum of the trunk (Euler's Laws in angular momentum form), which are foundational in the iterative Newton-Euler methods used in inverse dynamics.

Figure 2.

Tasks (double and single limb support) and functional phases of the step ascent and descent expressed as a percentage of the loading period (leading limb foot initial contact to trailing limb foot initial contact), as defined by Zachazewski et al., (1993).

Figure 3.

Trunk *translational* moment about the leading stance foot (a) mean ensemble averages and (b) mean (1 SD) peak (minimum and maximum) during the *step ascent* on the amputated limb (red), intact limb (blue), and right limb of healthy controls (black). Significant differences ($P < 0.017$) within and across groups are as follows: amputated vs. intact limbs (▲), amputated limb vs. healthy controls (*), and intact limb vs. healthy controls (+).

Figure 4.

Trunk *rotational* moment about the leading stance foot (a) mean ensemble averages and (b) mean (1 SD) peak (minimum and maximum) during the *step ascent* on the amputated limb (red), intact limb (blue), and right limb of healthy controls (black). Significant differences ($P < 0.017$) within and across groups are as follows: amputated vs. intact limbs (▲), amputated limb vs. healthy controls (*), and intact limb vs. healthy controls (+).

Figure 5.

Trunk *translational* moment about the leading stance foot (a) mean ensemble averages and (b) mean (1 SD) peak (minimum and maximum) during the *step descent* on the amputated limb (red), intact limb (blue), and right limb of healthy controls (black). Significant differences ($P < 0.017$) within and across groups are as follows: amputated vs. intact limbs (▲), amputated limb vs. healthy controls (*), and intact limb vs. healthy controls (+).

Figure 6.

Trunk *rotational* moment about the leading stance foot (a) mean ensemble averages and (b) mean (1 SD) peak (minimum and maximum) during the *step descent* on the amputated limb (red), intact limb (blue), and right limb of healthy controls (black). Significant differences ($P < 0.017$) within and across groups are as follows: amputated vs. intact limbs (▲), amputated limb vs. healthy controls (*), and intact limb vs. healthy controls (+).

Mean (1 SD) participant characteristics for patients with dysvascular unilateral transtibial amputation (TTA) and healthy control (HC) groups. Functional performance was quantified using the stair climb test (SCT) (Powers et al., 1997; Bean et al., 2007; Schmalz et al., 2007; Bennell et al., 2011).

Table 1

Group	Age (Years)	BMI (kg/m ²)	Time since Amputation (Months)	Residual Limb Length (cm)	SCT – Ascent Time (s)	SCT – Total Time (s)
TTA	56.3 (4.5)	28.3 (2.7)	16.7 (5.2)	14.4 (2.9)	11.3 (3.3)	23.0 (6.7)
HC	64.6 (5.5)	27.4 (3.3)	-	-	4.8 (0.8)	9.0 (1.6)