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3D intersegmental knee loading in below-knee amputees across steady-state walking speeds

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ABSTRACT

Background: Unilateral below-knee amputees often develop comorbidities that include knee joint disorders (e.g., intact leg knee osteoarthritis), with the mechanisms leading to these comorbidities being poorly understood. Mechanical knee loading of non-amputees has been associated with joint disorders and shown to be influenced by walking speed. However, the relationships between amputee knee loading and speed have not been identified. This study examined three-dimensional mechanical knee loading of amputees across a wide range of steady-state walking speeds.

Methods: Fourteen amputees and 10 non-amputee control subjects were analyzed at four overground walking speeds. At each speed, intersegmental joint moment and force impulses (i.e., time-integrals over the stance phase) were compared between the control, intact and residual knees using repeated-measures ANOVAs.

Findings: There were no differences in joint force impulses between the intact and control knees. The intact knee abduction moment impulse was lower than the non-amputees at 0.6 and 0.9 m/s. The intact knee flexion moment impulses at 0.6, 1.2 and 1.5 m/s and knee external rotation moment impulses at all speeds were greater than the residual knee. The residual knee extension moment and posterior force impulses were insensitive to speed increases, while these quantities increased in intact and control knees.

Interpretation: These results suggest the intact knees of asymptomatic and relatively new amputees are not overloaded during walking compared to non-amputees. Increased knee loads may develop in response to prolonged prosthesis usage or joint disorder onset. Further study is needed to determine if the identified bilateral loading asymmetries across speeds lead to diminished knee joint health.

1. Introduction

Limb loss prevalence in the U.S. is projected to dramatically increase over the next 40 years, with the primary causes related to dysvascular disease and traumatic injuries (Ziegler-Graham et al., 2008). Unilateral below-knee amputees often develop comorbidities that include joint disorders in both the intact and residual legs (e.g., Gailey et al., 2008) and chronic pain in their legs (e.g., Ephraim et al., 2005) and lower backs (e.g., Burke et al., 1978; Ephraim et al., 2005). Pain is most frequent in the intact leg knee joint, and long-term prosthetic use is associated with increased prevalence of joint disorders (for review, see Gailey et al., 2008). For example, the prevalence of osteoarthritis in unilateral amputees is higher in the intact knee compared to the residual knee (Burke et al., 1978). In addition, the development of intact-knee osteoarthritis in older unilateral below-knee amputees (83%) is higher than in non-amputees (50%) (Lemaire and Fisher, 1994). However, the mechanisms leading to the development of such disorders are unclear.

Previous studies seeking to identify these mechanisms in amputees have examined a variety of factors including bone mineral density (Royer and Koenig, 2005), strength (Lloyd et al., 2010) and mechanical loading (Lloyd et al., 2010; Royer and Koenig, 2005; Royer and Wasilewski, 2006) of the knee joint, which were generally greater in the intact leg compared to the residual leg and non-amputees. For example, sagittal plane knee strength of amputees was found to be asymmetric compared to non-amputees (Lloyd et al., 2010). However, frontal plane knee moments at self-selected walking speeds (SSWS) and hip abduction strength were not significantly asymmetric compared to non-amputees (Lloyd et al., 2010). Previous studies investigating joint disorders in non-amputees have also examined mechanical loading of the knee in addition to other factors such as joint laxity, varus alignment and serum hyaluronan level (Miyazaki et al., 2002; Sharma et al., 1998). These studies have shown that increased knee loading is associated with knee joint disorders (Miyazaki et al., 2002; Sharma et al., 1998) and is proportional to disease severity (for review, see Foroughi et al., 2009; Mundermann et al., 2004). In addition, other factors such as specific walking patterns

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(Fregly et al., 2009), gait retraining (Barrios et al., 2010) and altered walking speed (Mundermann et al., 2004) have been shown to influence knee loading in non-amputees and may offer effective non-invasive treatment interventions for amputees. However, the relationships between these quantities, specifically knee loading and walking speed in amputees, remain unclear. Since unilateral amputees commonly walk using passive prosthetic feet on their residual leg, which cannot perform net positive work, compensation from other muscles to provide body support and propulsion and differences in walking mechanics as walking speed increases may influence the joint loads that develop in amputees.

Studies analyzing sagittal plane mechanical loading during amputee walking have identified bilateral loading asymmetry between the residual and intact legs. Greater intact leg ground reaction forces (GRFs) and stance time (e.g., Sanderson and Martin, 1997; Silverman et al., 2008) as well as intersegmental joint moments and powers (e.g., Beyaert et al., 2008; Gitter et al., 1991) have been reported. Some studies have suggested that reduced residual leg loads compared to the intact leg are indicative of a mechanism to protect the residual leg (e.g., Beyaert et al., 2008; Nolan et al., 2003), which may have long-term detrimental consequences in the intact leg. Other studies examining intersegmental joint forces between legs have produced conflicting results. One study found a significantly smaller peak horizontal force and no difference in vertical force of the intact knee compared to non-amputees (Hurley et al., 1990), while others found no significant difference in peak vertical and horizontal forces of the intact knee compared to non-amputees, but reported a significantly greater vertical force impulse (Lemaire and Fisher, 1994). Thus, while studies have shown that the intact leg experiences greater sagittal plane GRFs and joint moments and powers, consistently greater intact leg intersegmental forces have not been observed.

Previous analyses of frontal plane loading during amputee walking have shown the intact knee abduction moment peak magnitude to be significantly greater (~56%) than in the residual leg at their SSWS (Royer and Koenig, 2005; Royer and Wasilewski, 2006). However, there was no significant difference in knee loads compared to non-amputees (Royer and Koenig, 2005). Others have shown up to a 44% difference in the abduction moment magnitude between intact and residual knees, although they were not statistically compared (Underwood et al., 2004).

Previous observational studies examining amputee knee loading have emphasized loading magnitudes and typically do not control walking speed. Recent studies analyzing below-knee amputee walking have shown that differences in GRFs, joint work and muscle activity between residual, intact and control legs are not always consistent across walking speeds (Fey et al., 2010; Silverman et al., 2008), which will influence both the magnitude and duration of knee loading characteristics. In this study, we analyzed amputees and non-amputees walking at different speeds to identify the influence of speed on knee loading. Since the loading magnitude and duration change with speed, both factors would influence the mechanical loading dose and response characteristics. Therefore, in order to quantify this relationship, knee loading impulses (i.e., the time-integral of the corresponding load) were compared during stance of both amputees and non-amputees.

The overall objective of this study was to examine three-dimensional mechanical knee loading across a wide range of steady-state walking speeds in below-knee amputees and non-amputees. Four hypotheses were evaluated at each walking speed: (1) net intersegmental joint moment impulses and (2) force impulses of the intact knee of amputees will be greater than non-amputee control subjects, and (3) net intersegmental joint moment impulses and (4) force impulses of the intact knee of amputees will be greater than in the residual knee. In addition, we expected the influence of walking speed on moment and force impulses to be different between the residual and intact knees of amputees, and between the intact knee of amputees and non-amputee control subjects.

2. Methods

2.1. Subjects

Each subject provided informed consent to an Institutional Review Board approved protocol. Subjects were pain free and asymptomatic of joint disorders. Fourteen unilateral, below-knee amputees (11 traumatic, 3 vascular; 13 males, 1 female; 45 (9) years mean (standard deviation); 90.6 (18.6) kg; 1.76 (0.10) m) and 10 control non-amputees (7 males, 3 females; 33 (12) years; 70.9 (13.6) kg; 1.76 (0.11) m) were analyzed. Time from amputation was $\geq$1.5 years and averaged 6 (3) years. The vascular and traumatic amputee groups did not substantially differ in terms of body weight or activity level, and all amputee subjects were at least limited community ambulators. Each amputee used their own prosthesis (9 energy storage and return, ESAR, and 5 solid ankle cushioned heel, SACH, prosthetic feet). Prosthesis alignment and fit were verified by a licensed prosthetist.

2.2. Procedures

Each subject walked along a 10-m walkway at four randomly-ordered, steady-state speeds of 0.6, 0.9, 1.2 and 1.5 m/s. For each walking trial, speed was measured using infrared timing gates. To be included in the analysis, the speed was required to be within 0.06 m/s of the target speed. Bilateral kinematic marker data (Silverman et al., 2008) were measured at 120 Hz using an eight-camera motion capture system (Vicon, OMG plc, Oxford, UK). Reflective markers were placed on the C-7 vertebrae and bilaterally on the acromion, iliac crest, posterior superior iliac spine, anterior superior iliac spine, greater trochanter, lateral and medial femoral condyles, lateral and medial malleoli, heel, dorsal foot, and the first, second and fifth metatarsal heads. To minimize skin movement artifact, marker clusters were also placed bilaterally on the shank and thigh. Markers on the foot of the residual leg were placed symmetric to the intact leg. GRF data were measured at 1200 Hz using four embedded force plates (Advanced Mechanical Technology, Inc., Watertown, MA, USA). Repeated trials were collected until at least five complete gait cycles per foot were measured at each speed.

2.3. Data analyses

Inverse dynamics analysis was performed to calculate net intersegmental joint moments and forces using Visual3D (C-Motion, Inc., Germantown, MD, USA), which were expressed in the local orthogonal femoral frame (Fig. 1). Kinematic data were low-pass filtered using a 4th-order Butterworth filter with a cut-off frequency of 6 Hz, and GRF data were low-pass filtered with a cut-off frequency of 20 Hz. Inertial properties of the residual leg were adjusted for the amputees and were based on previous measurements of residual limb mass and center-of-mass location for below-knee amputees (Mattes et al., 2000). Residual shank mass was reduced to 50% of the intact leg shank mass (2.3% body weight). Residual shank center-of-mass location was moved more proximally to 25% of the total shank length measured relative to the knee. Joint moments and forces were normalized by each subject's body mass.

Knee intersegmental joint moment and force impulses (i.e., time integral of the corresponding moment or force) were computed over the stance phase (ipsilateral heel strike to toe off). Knee extension, flexion, abduction, internal rotation and external rotation moment impulses and lateral (positive x-direction), anterior (positive y-direction), posterior (negative y-direction) and distal (negative z-direction) force impulses of the residual, intact and control legs were compared at each walking speed. Control subject data from the left and right legs were averaged for the analyses.

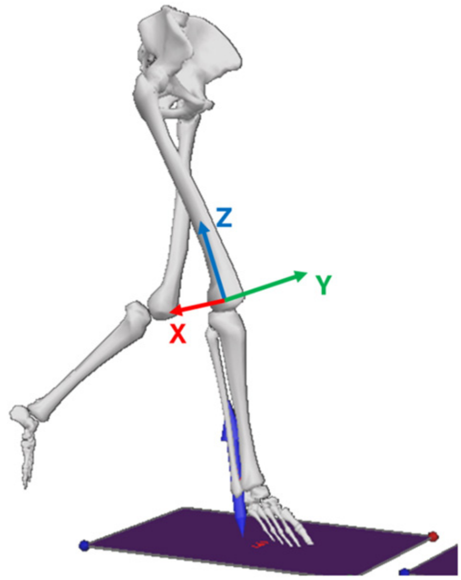

Fig. 1. The local femoral reference frame used to express intersegmental moments and forces was defined as positive x-direction oriented laterally, positive z-direction oriented proximally and axially along the femur, and positive y-direction orthogonal to x- and z-direction axes and oriented anteriorly.

2.4. Statistical analyses

For each impulse tested, two, two-factor repeated-measures ANOVAs were performed to compare quantities across legs, group and speed (SPSS 16.0 GP, SPSS, Inc., Chicago, IL, USA). The two ANOVAs included: (1) intact and residual legs (two-factor, leg and speed), and (2) intact and control legs (two-factor, group and speed). The leg factor consisted of two levels (residual leg, intact leg). The group factor consisted of two levels (amputee, non-amputee). The speed factor consisted of four levels (0.6, 0.9, 1.2 and 1.5 m/s). If a significant main or interaction effect was found ($\alpha = 0.05$), pairwise comparisons were made using a Bonferroni adjustment for multiple comparisons to determine which factors were significantly different. To further assess how these impulses varied with respect to walking speed, a polynomial decomposition was evaluated in each ANOVA to determine if there was a significantly different linear trend with walking speed in the intact, residual and control legs ($\alpha = 0.05$).

3. Results

3.1. Joint moments — intact to control comparison

The hypothesis that the intact knee moment impulses would be greater than in the control subjects was not supported in any of the moment impulses tested. In contrast, the knee abduction moment impulse was significantly lower (group, $P = 0.039$; speed∗group, $P < 0.001$) in the intact knee compared to control subjects at 0.6 and 0.9 m/s ($P \leq 0.031$) and showed a significantly different linear relationship between intact and control legs as walking speed increased ($P < 0.001$, Figs. 2 and 3, Table 1).

3.2. Joint moments — intact to residual comparison

The hypothesis that the intact knee moment impulses would be greater than in the residual knee was supported by the knee flexion moment impulse (leg, $P = 0.001$; leg∗speed, $P = 0.033$) at 0.6, 1.2 and 1.5 m/s ($P \leq 0.009$) (Figs. 2 and 3, Table 1). This hypothesis was also supported by the knee external rotation moment impulse (leg, $P = 0.001$; leg∗speed, $P = 0.025$), which was larger in the intact knee compared to the residual knee at all speeds ($P \leq 0.041$). In addition, the knee extension moment impulse had a significant leg∗speed interaction effect ($P = 0.015$) with a significantly different linear

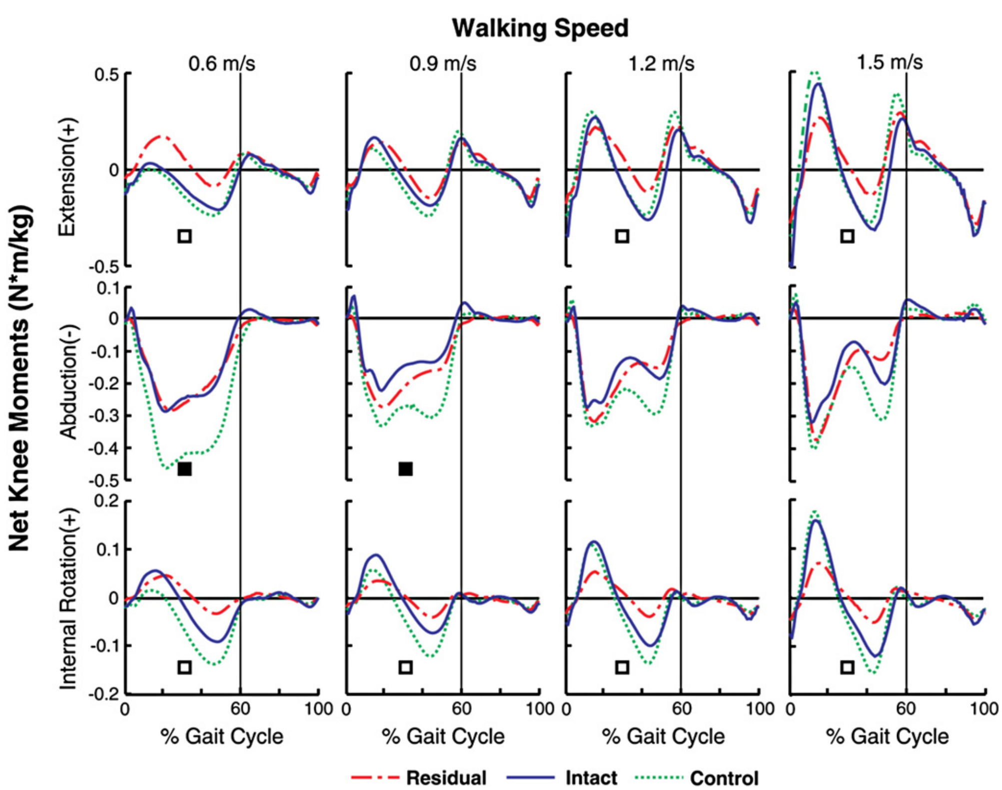

Fig. 2. Group average net internal joint moments (N∗m/kg) of the residual, intact and control knees across steady-state walking speeds. Significant pairwise differences between the intact and control knees (■) and between the intact and residual knees (□) are indicated.

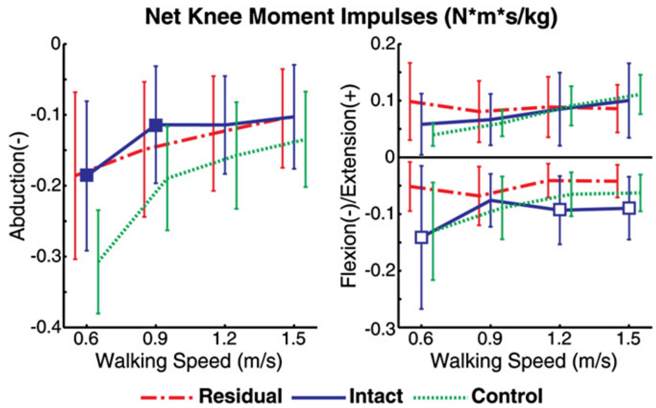

Fig. 3. Group average net internal joint moment impulses (N∗m∗s/kg) of the residual, intact and control knees across steady-state walking speeds. Vertical bars represent one standard deviation of the experimental data. Significant pairwise differences between the intact and control knees (■) and between intact and residual knees (□) are indicated.

relationship as speed increased between the intact and residual legs ($P=0.003$, Fig. 3). The residual knee extension moment impulse did not significantly change with speed, while the intact knee extension moment impulse increased as speed increased (0.6 to 1.5 m/s, $P=0.043$; 0.9 to 1.5 m/s, $P=0.019$). However, there were no significant pairwise differences between intact and residual knee extension moment impulse. In addition, the internal rotation moment impulse had a significantly different linear relationship as speed increased between the intact and residual legs ($P=0.025$).

3.3. Intersegmental forces — intact to control comparison

The hypothesis that the intact knee force impulses would be greater than in the control subjects was not supported, as there were no significant differences found between the intact and control knees at any speed (Fig. 4, Table 2).

3.4. Intersegmental forces — intact to residual comparison

The hypothesis that the intact knee joint force impulses would be greater than in the residual knee was not supported in any of the force impulses. Differences in lateral force impulses approached significance (leg, $P=0.062$) between intact and residual knees (Fig. 4, Table 2). Posterior force impulse had a significant leg∗speed interaction effect ($P<0.001$) and significantly different linear relationships as speed increased between the intact and residual legs ($P=0.002$, Fig. 4, Table 2), but there were no differences between the intact and residual legs at any speed. A significant speed effect ($P=0.001$) was found in the posterior force impulse. However, as walking speed increased, the posterior force impulse did not significantly change in the residual knee and significantly increased with walking speed in the intact knee (0.6 to 1.2 m/s, $P<0.001$; 0.6 to 1.5 m/s, $P<0.001$; 0.9 to 1.2 m/s, $P<0.001$; 0.9 to 1.5 m/s, $P<0.001$; 1.2 to 1.5 m/s, $P=0.021$).

4. Discussion

We expected that the amputee intact knee loading would be greater than in the non-amputee subjects. In contrast, intact knee abduction moment impulse was actually smaller compared to the control subjects at 0.6 and 0.9 m/s (Figs. 2 and 3, Table 1) and no differences were found in the intersegmental forces of the amputee and non-amputee knees. Previously, an increased internal knee abduction (external knee adduction) moment magnitude was associated with osteoarthritis disease progression (Miyazaki et al., 2002) and severity (Mundermann et al., 2004) in non-amputees. However, the non-amputee subjects were either already experiencing osteoarthritis and/or were substantially older (Miyazaki et al., 2002; Mundermann et al., 2004; Sharma et al., 1998). Thus, it is not clear whether the increased knee abduction moment of osteoarthritic non-amputees contributes to disease onset or is an adaptation resulting from disease onset and progression. An increased intact knee abduction moment peak magnitude at SSWS was previously shown in amputee subjects (Royer and Koenig, 2005; Royer and Wasilewski, 2006) that were free of joint pain and of similar age, 41 (10) years mean (standard deviation), to our amputee subjects, 45 (9) years. However, the post-amputation times for their subjects were substantially larger compared to our amputee subjects (17 (11) years compared to 6 (3) years). These previous studies along with our current knee abduction moment results suggest that prolonged prosthetic usage and/or aging may lead to frontal plane knee loading asymmetries over time.

In the sagittal plane, the larger intact knee flexor moment impulse compared to the residual knee at 0.6, 1.2 and 1.5 m/s supported our third hypothesis (Figs. 2 and 3, Table 1). Previously, these same subjects showed increased residual leg quadriceps activity across walking speeds (Fey et al., 2010), which would act to reduce the residual knee flexion moment and may provide additional body support in the absence of the ankle plantar flexors (e.g., Neptune et al., 2004). Increased residual leg hamstring activity was also shown in these

Table 1
Group average, mean (standard deviation), net internal joint moment impulses (N∗m∗s/kg) of the residual, intact and control knees during stance across steady-state walking speeds. Significant pairwise differences (α = 0.05) between intact and control knees and intact and residual knees are indicated by the shaded regions.

		0.6 m/s	0.9 m/s	1.2 m/s	1.5 m/s
Extension (N∗m∗s/kg)	Residual	0.099 (0.068)	0.081 (0.054)	0.089 (0.053)	0.086 (0.042)
	Intact	0.058 (0.054)	0.067 (0.045)	0.085 (0.064)	0.100 (0.066)
	Control	0.040 (0.020)	0.060 (0.023)	0.091 (0.035)	0.111 (0.035)
Flexion (N∗m∗s/kg)	Residual	− 0.051 (0.043)	− 0.068 (0.052)	− 0.041 (0.030)	− 0.042 (0.029)
	Intact	− 0.141 (0.126)	− 0.076 (0.047)	− 0.093 (0.060)	− 0.090 (0.055)
	Control	− 0.130 (0.086)	− 0.089 (0.055)	− 0.065 (0.039)	− 0.063 (0.032)
Abduction (N∗m∗s/kg)	Residual	− 0.186 (0.118)	− 0.149 (0.096)	− 0.126 (0.081)	− 0.110 (0.070)
	Intact	− 0.186 (0.105)	− 0.114 (0.083)	− 0.114 (0.069)	− 0.103 (0.073)
	Control	− 0.307 (0.073)	− 0.190 (0.073)	− 0.157 (0.075)	− 0.135 (0.067)
Internal rot. (N∗m∗s/kg)	Residual	0.025 (0.017)	0.019 (0.015)	0.017 (0.010)	0.019 (0.014)
	Intact	0.026 (0.017)	0.026 (0.016)	0.027 (0.019)	0.030 (0.020)
	Control	0.012 (0.008)	0.015 (0.008)	0.020 (0.010)	0.025 (0.009)
External rot. (N∗m∗s/kg)	Residual	− 0.020 (0.013)	− 0.019 (0.015)	− 0.012 (0.008)	− 0.014 (0.009)
	Intact	− 0.044 (0.034)	− 0.026 (0.015)	− 0.029 (0.020)	− 0.029 (0.016)
	Control	− 0.067 (0.042)	− 0.039 (0.024)	− 0.033 (0.017)	− 0.030 (0.014)

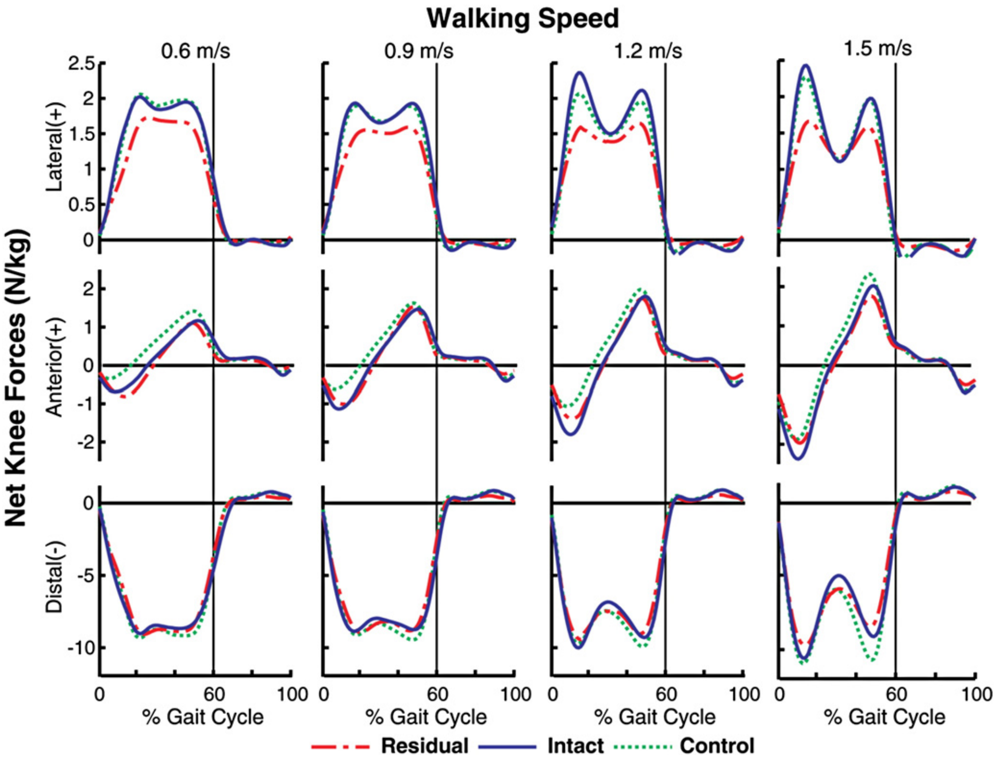

Fig. 4. Group average net internal joint forces (N/kg) of the residual, intact and control knees across steady-state walking speeds.

subjects across all speeds (Fey et al., 2010), which would act to increase hip extension and knee flexion moments. Based on these results, increased residual leg hamstring activity may increase hip extensor moments (e.g., Gitter et al., 1991; Silverman et al., 2008) and have a smaller effect at the knee. Further work is needed to examine if the increased intact knee flexor moment impulse is unfavorable to intact knee health.

Significantly different linear relationships as walking speed increased were found in the residual knee extension, abduction and internal rotation moment impulses and posterior force impulse compared to the intact knee. In addition, residual knee extension moment impulse (Figs. 2 and 3, Table 1) and posterior force impulse (Fig. 4, Table 2) did not significantly change as walking speed increased, while they did increase in the intact and control knees. It is unclear whether these different relationships with walking speed result in less favorable loading conditions in the intact knee as amputees modulate speed. Future studies analyzing amputees during accelerated/decelerated walking or typical maneuvering tasks such as standing

Table 2
Group average, mean (standard deviation), net internal joint force impulses (N∗s/kg) of the residual, intact and control knees during stance across steady-state walking speeds. There were no significant differences between legs at any speed.

		0.6 m/s	0.9 m/s	1.2 m/s	1.5 m/s
Lateral	Residual	1.26 (0.44)	1.01 (0.34)	0.88 (0.31)	0.77 (0.28)
(N∗s/kg)	Intact	1.50 (0.35)	1.19 (0.27)	1.09 (0.25)	0.95 (0.20)
	Control	1.54 (0.49)	1.19 (0.37)	1.02 (0.34)	0.86 (0.28)
Anterior	Residual	0.40 (0.31)	0.41 (0.31)	0.36 (0.24)	0.38 (0.21)
(N∗s/kg)	Intact	0.51 (0.42)	0.47 (0.30)	0.44 (0.24)	0.45 (0.16)
	Control	0.69 (0.46)	0.58 (0.36)	0.54 (0.28)	0.50 (0.21)
Posterior	Residual	−0.36 (0.27)	−0.33 (0.21)	−0.35 (0.18)	−0.40 (0.18)
(N∗s/kg)	Intact	−0.27 (0.24)	−0.30 (0.23)	−0.39 (0.23)	−0.46 (0.22)
	Control	−0.14 (0.17)	−0.17 (0.16)	−0.23 (0.14)	−0.32 (0.15)
Distal	Residual	−6.81 (0.83)	−5.82 (0.63)	−4.98 (0.44)	−4.41 (0.30)
(N∗s/kg)	Intact	−7.07 (0.85)	−5.92 (0.49)	−5.12 (0.43)	−4.56 (0.42)
	Control	−7.30 (0.82)	−6.17 (0.64)	−5.33 (0.56)	−4.71 (0.54)

and hopping on their intact leg may provide insight since ~40% of day to day walking in adults located in urban environments are short duration bouts of less than twelve steps (Orendurff et al., 2008). This is especially important for below-knee amputees since they are commonly classified as limited community ambulators, which are characterized by a reduced walking range (Johnson et al., 1995). Thus, amputees may experience increased intact leg loading demands during these types of non-steady-state activities or perform these activities with greater regularity, which may be detrimental to knee joint health.

No differences in intersegmental joint force impulses were found between legs. An important limitation of analyzing intersegmental joint forces is that they underestimate net joint contact forces by not accounting for co-contraction of antagonistic muscles or compressive forces due to ligaments (e.g., Zajac et al., 2002). Therefore, differences in knee contact forces may exist. However, intersegmental forces do quantify the contributions of GRFs and body segment inertial forces to the net knee contact force, and therefore provide useful insight into knee loading. Since studies have identified increases in intact leg GRFs compared to the residual leg (e.g., Nolan et al., 2003; Silverman et al., 2008), it was reasonable to expect larger knee intersegmental forces would develop. Previous studies examining intersegmental joint forces of amputees between intact and control legs have produced conflicting results and found either a smaller intact knee intersegmental horizontal force and no difference in vertical forces when testing younger amputees (36 (7) years; Hurley et al., 1990), or found no significant difference in peak vertical and horizontal forces of the intact knee and a significantly greater vertical force impulse when testing elderly amputees (72 (4) years; Lemaire and Fisher, 1994). These previous studies and our intersegmental force results again suggest that longer prosthetic use and/or aging may play a role in developing higher intact knee loading over time.

An interesting finding was that as walking speed increased, we observed the highest degree of symmetry between the residual and intact knee moments and forces at 0.9 m/s (Fig. 2, Table 1). This finding is consistent with a previous work (Nolan et al., 2003) that tested amputees walking at 0.5, 0.9 and 1.2 m/s and found that the highest degree of vertical ground reaction force symmetry occurred at 0.9 m/s. Other studies have shown that amputees exhibit a slower SSWS (e.g., Waters et al., 1976) and an increased metabolic cost at a given walking speed compared to non-amputees, which is minimized at their slower SSWS (e.g., Genin et al., 2008; Waters et al., 1976). Since the highest degree of loading symmetry occurs at a slower walking speed compared to the SSWS (~1.3 m/s) of non-amputees (e.g., Waters et al., 1976), unilateral amputees may prefer slower walking speeds to maintain a higher degree of bilateral knee loading symmetry in addition to reducing metabolic cost.

A potential limitation in this study is the amputees analyzed included multiple etiologies (traumatic, n = 11; vascular, n = 3). Also, amputees walked using their own clinically prescribed prostheses which included ESAR (n = 9) and SACH (n = 5) feet. To assess whether this influenced our results, the analyses were performed on two subsets of amputees (i.e., only traumatic amputees and only amputees that used ESAR feet). We found that the statistical results were similar and our overall conclusions remained unchanged.

5. Conclusions

In summary, intact knee mechanical loading during steady-state walking was similar or even lower at some speeds compared to the control subjects. Thus, in terms of intersegmental joint moments and forces, it does not appear that the intact knee is at a higher risk of overloading compared to non-amputees. However, there were sagittal and transverse plane bilateral loading asymmetries in the flexion and external rotation moment impulses, which were larger in the intact knee at multiple speeds. In addition, the residual knee extension moment impulse and posterior force impulse were insensitive to walking speed increases, while these quantities increased in the intact and control knees. Future modeling studies are needed to quantify the contributions of muscles, ligaments and external loads to the knee contact forces (e.g., Winby et al., 2009) to determine if these bilateral asymmetries across speed lead to less favorable intact knee loading that contributes to joint disorders over time. In addition, factors other than mechanical loading and walking speed such as non-steady-state task demands and frequency, bone mineral density (Royer and Koenig, 2005), joint laxity and varus alignment (Miyazaki et al., 2002; Sharma et al., 1998), joint contact geometry and localized stress distribution (Koo and Andriacchi, 2007), and specific walking patterns (Fregly et al., 2009) may also play important roles in the development of joint disorders in lower-limb amputees that warrants further research.

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References

Barrios, J.A., Crossley, K.M., Davis, I.S., 2010. Gait retraining to reduce the knee adduction moment through real-time visual feedback of dynamic knee alignment. J. Biomech. 43 (11), 2208–2213.

Beyaert, C., Grumillier, C., Martinet, N., Paysant, J., Andre, J.M., 2008. Compensatory mechanism involving the knee joint of the intact limb during gait in unilateral below-knee amputees. Gait Posture 28 (2), 278–284.

Burke, M.J., Roman, V., Wright, V., 1978. Bone and joint changes in lower limb amputees. Ann. Rheum. Dis. 37 (3), 252–254.

Ephraim, P.L., Wegener, S.T., MacKenzie, E.J., Dillingham, T.R., Pezzin, L.E., 2005. Phantom pain, residual limb pain, and back pain in amputees: results of a national survey. Arch. Phys. Med. Rehabil. 86 (10), 1910–1919.

Fey, N.P., Silverman, A.K., Neptune, R.R., 2010. The influence of increasing steady-state walking speed on muscle activity in below-knee amputees. J. Electromyogr. Kinesiol. 20 (1), 155–161.

Foroughi, N., Smith, R., Vanwanseele, B., 2009. The association of external knee adduction moment with biomechanical variables in osteoarthritis: a systematic review. Knee 16 (5), 303–309.

Fregly, B.J., D'Lima, D.D., Colwell Jr., C.W., 2009. Effective gait patterns for offloading the medial compartment of the knee. J. Orthop. Res. 27 (8), 1016–1021.

Gailey, R., Allen, K., Castles, J., Kucharik, J., Roeder, M., 2008. Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use. J. Rehabil. Res. Dev. 45 (1), 15–29.

Genin, J.J., Bastien, G.J., Franck, B., Detrembleur, C., Willems, P.A., 2008. Effect of speed on the energy cost of walking in unilateral traumatic lower limb amputees. Eur. J. Appl. Physiol. 103 (6), 655–663.

Gitter, A., Czerniecki, J.M., DeGroot, D.M., 1991. Biomechanical analysis of the influence of prosthetic feet on below-knee amputee walking. Am. J. Phys. Med. Rehabil. 70 (3), 142–148.

Hurley, G.R., McKenney, R., Robinson, M., Zadravec, M., Pierrynowski, M.R., 1990. The role of the contralateral limb in below-knee amputee gait. Prosthet. Orthot. Int. 14 (1), 33–42.

Johnson, V.J., Kondziela, S., Gottschalk, F., 1995. Pre and post-amputation mobility of trans-tibial amputees: correlation to medical problems, age and mortality. Prosthet. Orthot. Int. 19 (3), 159–164.

Koo, S., Andriacchi, T.P., 2007. A comparison of the influence of global functional loads vs. local contact anatomy on articular cartilage thickness at the knee. J. Biomech. 40 (13), 2961–2966.

Lemaire, E.D., Fisher, F.R., 1994. Osteoarthritis and elderly amputee gait. Arch. Phys. Med. Rehabil. 75 (10), 1094–1099.

Lloyd, C.H., Stanhope, S.J., Davis, I.S., Royer, T.D., 2010. Strength asymmetry and osteoarthritis risk factors in unilateral trans-tibial, amputee gait. Gait Posture 32 (3), 296–300.

Mattes, S.J., Martin, P.E., Royer, T.D., 2000. Walking symmetry and energy cost in persons with unilateral transtibial amputations: matching prosthetic and intact limb inertial properties. Arch. Phys. Med. Rehabil. 81 (5), 561–568.

Miyazaki, T., Wada, M., Kawahara, H., Sato, M., Baba, H., Shimada, S., 2002. Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. Ann. Rheum. Dis. 61 (7), 617–622.

Mundermann, A., Dyrby, C.O., Hurwitz, D.E., Sharma, L., Andriacchi, T.P., 2004. Potential strategies to reduce medial compartment loading in patients with knee osteoarthritis of varying severity: reduced walking speed. Arthritis Rheum. 50 (4), 1172–1178.

Neptune, R.R., Zajac, F.E., Kautz, S.A., 2004. Muscle force redistributes segmental power for body progression during walking. Gait Posture 19 (2), 194–205.

Nolan, L., Wit, A., Dudzinski, K., Lees, A., Lake, M., Wychowanski, M., 2003. Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees. Gait Posture 17 (2), 142–151.

Orendurff, M.S., Bernatz, G.C., Schoen, J.A., Klute, G.K., 2008. Kinetic mechanisms to alter walking speed. Gait Posture 27 (4), 603–610.

Royer, T., Koenig, M., 2005. Joint loading and bone mineral density in persons with unilateral, trans-tibial amputation. Clin. Biomech. 20 (10), 1119–1125.

Royer, T.D., Wasilewski, C.A., 2006. Hip and knee frontal plane moments in persons with unilateral, trans-tibial amputation. Gait Posture 23 (3), 303–306.

Sanderson, D.J., Martin, P.E., 1997. Lower extremity kinematic and kinetic adaptations in unilateral below-knee amputees during walking. Gait Posture 6, 126–136.

Sharma, L., Hurwitz, D.E., Thonar, E.J., Sum, J.A., Lenz, M.E., Dunlop, D.D., et al., 1998. Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis. Arthritis Rheum. 41 (7), 1233–1240.

Silverman, A.K., Fey, N.P., Portillo, A., Walden, J.G., Bosker, G., Neptune, R.R., 2008. Compensatory mechanisms in below-knee amputee gait in response to increasing steady-state walking speeds. Gait Posture 28 (4), 602–609.

Underwood, H.A., Tokuno, C.D., Eng, J.J., 2004. A comparison of two prosthetic feet on the multi-joint and multi-plane kinetic gait compensations in individuals with a unilateral trans-tibial amputation. Clin. Biomech. 19 (6), 609–616.

Waters, R.L., Perry, J., Antonelli, D., Hislop, H., 1976. Energy cost of walking of amputees: the influence of level of amputation. J. Bone Joint Surg. Am. 58 (1), 42–46.

Winby, C.R., Lloyd, D.G., Besier, T.F., Kirk, T.B., 2009. Muscle and external load contribution to knee joint contact loads during normal gait. J. Biomech. 42 (14), 2294–2300.

Zajac, F.E., Neptune, R.R., Kautz, S.A., 2002. Biomechanics and muscle coordination of human walking. Part I: introduction to concepts, power transfer, dynamics and simulations. Gait Posture 16 (3), 215–232.

Ziegler-Graham, K., MacKenzie, E.J., Ephraim, P.L., Travison, T.G., Brookmeyer, R., 2008. Estimating the prevalence of limb loss in the united states: 2005 to 2050. Arch. Phys. Med. Rehabil. 89 (3), 422–429.