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Biomechanical risk factors for knee osteoarthritis when using passive and powered ankle–foot prostheses☆

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ABSTRACT

Background: Gait compensations following transtibial amputation negatively affect sound limb loading and increase the risk of knee osteoarthritis. Push-off assistance provided by new powered prostheses may decrease the demands on the sound limb. However, their effects in a young population in the early stages of prosthetic use are still unknown. The purpose of this study was to compare limb loading between 1. passive and powered ankle–foot prostheses, 2. sound and amputated limbs, and 3. individuals with amputations in the relatively early stages of prosthetic use and controls.
Methods: Ten young, active individuals with unilateral transtibial amputation and 10 controls underwent biomechanical gait analysis at three speeds. The peak external knee flexor and adductor moments, adductor moment's angular impulse, peak vertical ground reaction force and loading rate were calculated. Repeated measures ANOVAs compared between limbs, prostheses, and groups.
Findings: The powered prosthesis did not decrease the sound limb's peak adduction moment or its impulse, but did decrease the external flexor moment, peak vertical force and loading rate as speed increased. The powered prosthesis decreased the loading rate from controls. The sound limb did not display a significantly greater risk for knee osteoarthritis than the intact limb or than controls in either device.
Interpretation: In the early stages of prosthetic use, young individuals with transtibial amputation display few biomechanical risk factors for knee osteoarthritis development. However, a powered ankle–foot prosthesis still offers some benefits and may be used prophylactically to mitigate potential increases of these variables with continued prosthetic use over time.

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1. Introduction

Of the more than 1600 amputations which resulted from recent military efforts, nearly half involved the lower leg (Stinner et al., 2010). Lower limb amputations not only limit mobility, but may also predispose individuals to secondary musculoskeletal impairments and disability. Deviations from normal movement patterns are used to compensate for loss of the biologic limb function, resulting in asymmetrical motion and loading (Czerniecki, 1996; Engsberg et al., 1993; Lloyd et al., 2010; Sanderson and Martin, 1997; Seroussi et al., 1996). Specifically, the sound limb bears a greater portion of the load during walking than the amputated limb (Bateni and Olney, 2002; Sanderson and Martin, 1997). Increased reliance on the sound limb relates to greater resultant forces and knee moments, which have been proposed to increase the risk of sound limb musculoskeletal injury, joint degeneration and pain (Norvell et al., 2005; Royer and Koenig, 2005; Royer and Wasilewski, 2005).

Individuals with transtibial amputation (TTA) have a greater susceptibility to knee osteoarthritis (OA) than the general population (Burke et al., 1978; Gailey et al., 2008; Norvell et al., 2005; Royer and Wasilewski, 2005; Struyf et al., 2009) with 65% displaying evidence of knee OA in their sound limb (2001). These rates are disproportionately higher even after accounting for age, body weight, and history of knee trauma in the sound limb (2005). In long-time prosthetic users, knee pain in the sound limb is the primary complaint (Mussman et al., 1983). While individuals with TTA are 17 times more likely to develop knee OA in the sound limb compared to age-matched non-amputees (Struyf et al., 2009), they are five times less likely to develop pain in their amputated limb (Norvell et al., 2005). The asymmetrical loading between the limbs and subsequent development of knee pain suggest a potentially causative role of mechanical factors.

Accordingly, biomechanical risk factors for knee OA are greater in individuals with amputations than the general population. The external knee adduction moment (EKAM) is a widely-reported risk factor for

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knee osteoarthritis due to its relationship with the internal load on the medial compartment of the knee joint (Kutzner et al., 2013; Schipplein and Andriacchi, 1991). The magnitude of the EKAM and its angular impulse are associated with knee OA severity (Astephen et al., 2008; Mundermann et al., 2005; Thorp et al., 2006) and greater values at baseline are predictive of disease development and progression in a non-amputee population (Bennell et al., 2011; Miyazaki et al., 2002; Sharma et al., 1998). Royer and Wasilewski (2005) found significantly greater peak EKAMs in the sound limb of individuals with TTA compared to the amputated limb. This finding has been confirmed by other reports of greater EKAMs (Lloyd et al., 2010; Royer and Koenig, 2005) and prevalence of pain (Norvell et al., 2005) in the sound knee compared to the amputated limb. Previous investigations have also highlighted the importance of not only accounting for discrete parameters, such as the peak EKAM, but also other measures, such as the EKAM angular impulse, which assesses joint loading across the gait cycle (Bennell et al., 2011).

Large EKAMs and impulses are not the only biomechanical risk factors for knee osteoarthritis. The external knee flexor moment (EKFM) is related to the overall load on the knee and large external flexor moments have been shown to increase the frontal plane moment (Kulmala et al., 2013). The peak vertical ground reaction force (GRF) and loading rate are also greater during walking in individuals with knee OA (Mundermann et al., 2005) and with amputations (Grabowski and D'Andrea, 2013). Specifically, Mundermann et al. (2005) found that patients with knee OA had 50% greater loading rates than asymptomatic individuals. Hobara et al. (2014) reported greater loading rates in the intact sides than affected sides of individuals with TTA during running and Grabowski and D'Andrea (2013) reported greater loading rates on the intact side of amputees compared to controls at moderate walking speeds. When the articular cartilage that serves to protect and cushion the knee joint is loaded at increasingly faster rates, its ability to dissipate contact forces lessens (Radin and Paul, 1971), potentially contributing to degenerative changes.

Once this biomechanical degradation of the cartilage has been initiated, few treatments are available to impede the course of this disease. Advancements in prosthetic technology are constantly being sought to improve gait symmetry and decrease sound limb loading to delay or prevent disease development. Morgenroth et al. (2011) and Underwood et al. (2004) reported that increased push-off from the prosthetic limb can reduce the magnitude of the peak EKAM in the sound limb. While these authors tested the benefits of newer energy storage and return (ESR) prosthetic feet against more traditional models, these devices were still all passive in nature and limited in the amount of energy return they were able to provide at push-off.

In contrast to passive prostheses, currently-available powered ankle–foot prostheses provide greater push-off assistance during step-to-step transitions and have been shown to normalize many biomechanical parameters to those of able-bodied individuals (Herr and Grabowski, 2012). Grabowski and D'Andrea (2013) investigated the effect of powered prostheses on the external knee adduction moment and loading rate in individuals with amputation and found that powered assistance decreased the EKAM in the sound limb at fast walking speeds and the peak GRF at relatively slower speeds, but not the loading rates. The EKAM profiles suggested that, while the powered device provided a beneficial reduction in the peak EKAM at some speeds, the cumulative impulse may be negatively affected at others. At an average of 45 years, their subjects were younger overall than the general population of individuals with amputation (Centers for Disease and Prevention, 2001) but they had an average of 21 years of experience walking with prosthetic devices. They exhibited several biomechanical risk factors for knee OA compared to control subjects and it is possible that degenerative changes had already begun. It is currently unknown if the gait mechanics adopted by individuals in the early stages of prosthesis use increase their risk for disease development and if these biomechanical risk factors could be lessened with powered prostheses.

The high rates of knee OA following amputation are particularly concerning for young individuals with traumatic TTA because they will experience decades of prosthetic use combined with the general risk of OA with increasing age (Busija et al., 2010). Early recognition of risk factors and development of potential preventative strategies, such as powered prostheses, may reduce the risk of long-term cumulative disability. Therefore, the purpose of this study was to compare knee joint loading 1. between passive and powered ankle–foot prostheses across a range of walking speeds, 2. between the sound and prosthetic limbs, and 3. between young, asymptomatic individuals with TTA and able-bodied controls. We hypothesized that knee joint loading would be 1. less when individuals with TTA used the powered prosthesis than passive, 2. less on the prosthetic limb than the sound limb, and 3. greater on the sound limb of individuals with TTA than able-bodied controls.

2. Methods

2.1. Subjects

Ten service members with traumatic unilateral TTA provided written informed consent to participate in the protocol approved by the Brooke Army Medical Center Institutional Review Board (Table 1). All

Table 1
Subject Characteristics

	Gender	Age (yrs)	Height (m)	Leg length (m)	Mass + ESR (kg)	Mass + BiOM (kg)	Months ambulating	ESR prosthesis
1	Male	32	1.93	1.08	102.0	105.0	23	O. VSP
2	Male	38	1.80	1.01	99.0	100.0	9	O. VSP
3	Male	29	1.93	1.13	108.9	106.1	37	O. FlexFoot
4	Male	29	1.78	0.95	93.2	90.9	12	O. VSP
5	Male	38	1.87	1.04	96.4	97.3	11	F. Renegade
6	Male	26	1.70	0.93	97.7	100.5	9	F. Renegade
7	Female	34	1.65	0.86	85.5	85.2	33	F. Renegade
8	Male	29	1.83	0.93	93.2	94.0	25	O. VSP
9	Male	25	1.93	1.07	97.5	97.5	12	W. Pathfinder
10	Male	26	1.75	0.99	84.3	87.5	14	O. VSP
TTA		**30.2 (5.3)**	**1.82 (0.10)**	**1.00 (0.08)**	**95.8 (7.3)**	**96.4 (7.0)**	**18.5 (10.3)**	
Control		**28.7 (6.1)**	**1.80 (0.06)**	**0.93 (0.04)**	**81.4 (8.9)***			

Subject and prosthesis characteristics. O. — Ossür (Reykjavíc, Iceland), F. — Freedom Innovations (Irvine, CA), W. — Ohio Willow Wood (Mt. Sterling, OH). The different ESR prosthetic foot masses were: O. VSP: 709 g, O. FlexFoot: 700 g, F. Renegade: 358 g, W. Pathfinder: 785 g. Bold values indicate means (standard deviations).
* Indicates a statistically significant difference between groups.

subjects were in the relatively early stages of prosthetic use compared with prior literature, were relatively young, active, and could ambulate at a K3 Medicare Functional Classification Level or higher. All subjects wore their prescribed, passive ESR prosthesis for activities of daily living and had no prior experience with long-term use of powered devices. Ten able-bodied service members served as a control group.

2.2. Prosthetic device

The passive-dynamic ESR prosthetic foot absorbs shock and stores energy through a compressible heel and elastic keel when loaded and then releases stored energy at push-off. The powered prosthesis (BiOM, iWalk, Bedford, MA, USA) utilizes active, motorized components and passive ESR components to emulate biological foot function and provide powered push-off in terminal stance. BiOM ankle parameters, including ankle joint power production, timing of push-off, toe-off ankle angle, net non-conservative work, and foot stiffness, were tuned by a company representative. Parameters were maintained within two standard deviations of biological ankle data. Detailed descriptions of the BiOM can be found in the literature (Au and Herr, 2008; Eilenberg et al., 2010; Markowitz et al., 2011). Subjects had three weeks accommodation time with the BiOM prior to assessment. During this time, subjects wore the BiOM as their primary prosthesis for walking activities and minor adjustments to tuning parameters were made when requested by the subject.

2.3. Experimental setup

The experimental setup for overground walking consisted of a 26-camera motion capture system (120 Hz; Motion Analysis, Santa Rosa, CA, USA). The system tracked three-dimensional marker locations from 57 markers placed on segments and anatomical landmarks of the body (Wilken et al., 2012). A digitization pointer was used to identify 20 bilateral anatomical body landmarks (C-Motion, Inc. Germantown, MD, USA). The ankle joint center of the BiOM was calculated from digitized points on the medial and lateral aspects of the BiOM's axis of rotation. For the ESR device, the position of the ankle joint center in the intact foot coordinate system was mirrored into the foot coordinate system of the prosthetic limb resulting in comparable anterior/posterior, vertical, and medial/lateral positioning. The laboratory setup of five force platforms embedded in the floor and positioned in series recorded three-dimensional GRFs at the foot–ground interface (1200 Hz; AMTI, Watertown, MA, USA). Data were collected from all five force platforms along the walkway but only used when a full footstrike was made with one of the platforms. Steps in which the stance phase spanned two platforms were disregarded.

2.4. Protocol

Participants walked at three controlled walking velocities based on leg length and dimensionless Froude numbers of 0.10, 0.16 and 0.23 (Vaughan and O'Malley, 2005). Auditory cues based on the forward progression of a marker on the seventh cervical vertebrae provided feedback to the subject when they attained the intended speed. Subjects walked at each of these three speeds in both of the prosthetic device conditions: ESR then BiOM. Five strides from each limb were selected for analysis.

Subjects also completed a Prosthetics Evaluation Questionnaire (PEQ) (Roorda et al., 1996) for each prosthetic condition and sections relating to ambulation ability were analyzed.

2.5. Data analysis

Marker data were initially tracked in Cortex (Motion Analysis, Santa Rosa, CA, USA) and marker and analog data were exported to Visual3D (C-Motion, Inc., Germantown, MD, USA) for analysis. A 13-segment model was created and scaled to height and the sum of biological body mass and prosthesis mass. The prosthetic foot and shank were modeled using the properties of a biological intact ankle because it has previously been observed that the modified mass and center of mass location of the prosthetic foot and shank have minimal influence on joint moments during stance (Miller, 1987). Interpolated marker trajectories and GRF data were filtered using a low-pass, Butterworth, 4th order filter with cutoff frequencies of 6 and 50 Hz, respectively. The net EKFM and EKAM were calculated using inverse dynamics and expressed in the proximal segment's coordinate system. The peak values were identified as the greatest local maxima during the first half of stance. The impulse of the EKAM was calculated as the integral of the positive portion of the data during the stance phase of gait. The peak vertical GRF was obtained from the first peak during the first half of stance. Loading rates were calculated as the slope of the vertical GRF from 20 to 80% of the GRF profile from heel strike to the instant of peak force (Milner et al., 2006). GRFs and moments were scaled to body weight and mass, respectively, and differences in mass between prosthesis conditions were accounted for. Data were then time-normalized to 100% of the stance phase.

2.6. Statistical analysis

Statistical analyses were performed in SPSS. A three-way ANOVA (prosthesis × speed × limb) assessed statistical significance of the main effects. Significant main effects or interactions were separated with post-hoc paired t-tests Bonferroni–Holm corrections. The Bonferroni–Holm method uses a step-down approach to account for multiple comparisons by arranging P-values from the smallest to the largest and comparing them to sequential significance cutoffs (Holm, 1979). The unadjusted criterion for statistical significance was set at $P < 0.05$ and the appropriate alpha value related to the Bonferroni–Holm correction is presented for all post-hoc tests. A correction factor accounting for 12 comparisons was applied to three-way interactions (the smallest P-value is $0.05/12 = 0.0042$), a correction factor accounting for six comparisons was applied to two-way prosthesis × speed or limb × speed interactions (the smallest P-value is $0.05/6 = 0.0083$), and a correction factor accounting for four comparisons was applied to two-way prosthesis × limb interactions (the smallest P-value is $0.05/4 = 0.0125$). Comparisons to the control group were performed using a one-way ANOVA at each speed with Dunnett's post-hoc tests ($P < 0.05$).

3. Results

3.1. Subject characteristics

Subject characteristics are presented in Table 1. There were no significant differences between groups for standardized walking velocities based on leg lengths ($P = 0.0585$). Calculated slow, moderate, and fast walking velocities were, on average, 0.99 (0.04), 1.25 (0.05) and 1.50 (0.06) m/s for the TTA group, respectively, and 0.96 ± 0.02, 1.21 ± 0.03 and 1.45 ± 0.03 for the controls, respectively. Self-selected walking velocities were not significantly different between prosthesis conditions (ESR: 1.36 (0.05) m/s; BiOM: 1.44 (0.14) m/s; $P = 0.060$) and subject rating of ambulation ability using the PEQ was high in both devices (ESR: 79/100 (13), BiOM: 85/100 (13), $P = 0.2459$) (for additional PEQ results, see Ferris et al., 2012).

3.2. Knee moments

The peak EKAM increased with increasing speed ($P = 0.0032$) with both prosthetic devices and for both the sound and amputated limbs. There was a significant prosthesis × limb interaction ($P = 0.0034$) for the peak EKAM and subjects experienced a 15% increase in the peak EKAM on the sound side relative to the prosthetic side at the slowest

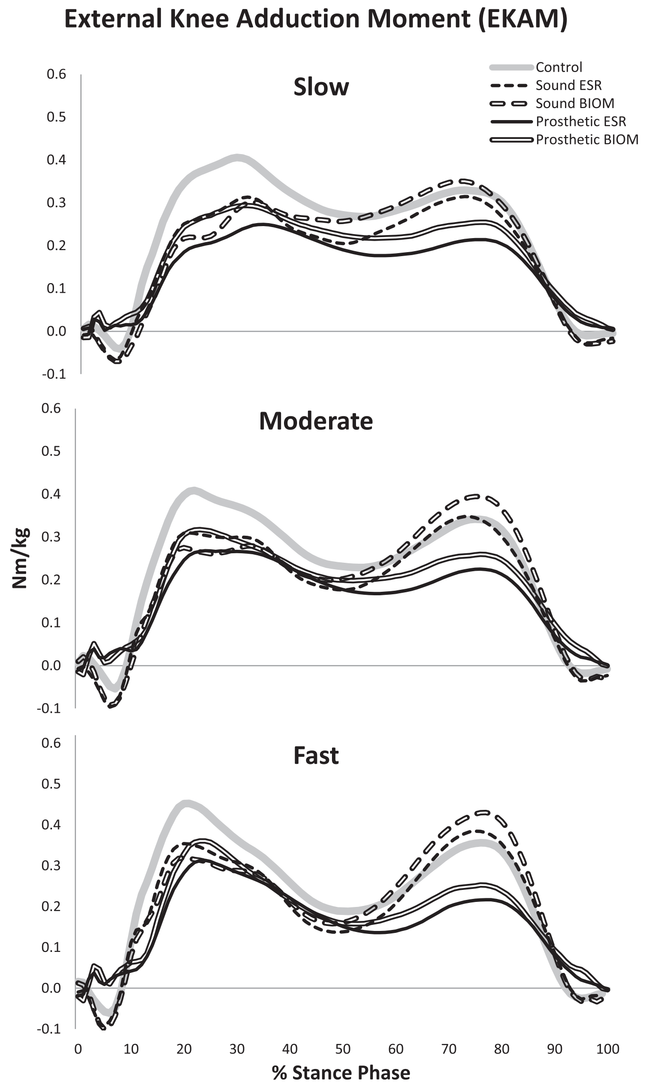

Fig. 1. Mean external knee adduction moment (EKAM) profiles across the stance phase.

Table 2
Kinetic Measures

	ESR		BIOM		Control
	Sound	Prosthetic	Sound	Prosthetic	
Peak EKAM (Nm/kg)					
Slow	0.33 (0.10)	0.28 (0.06)[ac]	0.34 (0.13)	0.32 (0.10)	0.44 (0.13)
Moderate	0.35 (0.11)	0.30 (0.07)	0.34 (0.12)	0.35 (0.12)	0.43 (0.10)
Fast	0.39 (0.15)	0.39 (0.09)	0.37 (0.14)	0.39 (0.12)	0.47 (0.11)
EKAM impulse (Nm/kg ∗ s)					
Slow	0.17 (0.06)	0.12 (0.05)	0.18 (0.08)	0.16 (0.06)	0.20 (0.07)
Moderate	0.16 (0.06)	0.12 (0.06)	0.16 (0.05)	0.14 (0.07)	0.18 (0.06)
Fast	0.15 (0.06)	0.13 (0.05)	0.16 (0.07)	0.12 (0.05)	0.17 (0.05)
Peak EKFM (Nm/kg)					
Slow	0.44 (0.23)	0.37 (0.11)	0.30 (0.08)	0.36 (0.16)	0.26 (0.09)
Moderate	0.58 (0.26)[c]	0.41 (0.15)	0.41 (0.06)	0.40 (0.13)	0.39 (0.10)
Fast	0.78 (0.21)[c]	0.42 (0.16)[a]	0.59 (0.08)[b]	0.42 (0.14)[a]	0.53 (0.14)
Peak GRF (BW)					
Slow	1.03 (0.04)	1.04 (0.03)	0.99 (0.04)	1.00 (0.03)	1.01 (0.02)
Moderate	1.09 (0.06)	1.05 (0.03)	1.02 (0.04)[b]	1.05 (0.05)	1.05 (0.03)
Fast	1.19 (0.06)	1.05 (0.03)	1.11 (0.03)[b]	1.12 (0.05)[a]	1.10 (0.02)
Loading rate 20–80% (BW/s)					
Slow	47.0 (6.2)	48.8 (8.6)	41.3 (7.7)	46.3 (8.2)	49.3 (9.5)
Moderate	56.8 (9.2)	51.4 (7.5)	43.8 (8.5)[bc]	57.6 (10.7)[a]	60.8 (10.2)
Fast	65.8 (10.7)	53.5 (6.8)	54.0 (10.9)[bc]	83.4 (17.8)[ab]	72.5 (17.0)

Mean dependent measures (standard deviations). Significance is based on post-hoc comparisons. Effects of speed are presented in the main text and are not accounted for in the table.
[a] Indicates significant differences between limbs.
[b] Indicates significant difference in the BiOM from the ESR prosthesis for the same limb.
[c] Indicates significant differences from controls.

speed ($P = 0.0047$, $\alpha = 0.0125$) in the ESR (Fig. 1, Table 2). In the BiOM, however, the sound and prosthetic sides were not significantly different. Across the three speeds, the peak EKAM was not different between BiOM and ESR ($P > 0.065$ for all comparisons). While the peak EKAM represents a discrete instance of loading, its impulse provides a measure of loading across stance, however, there were no significant differences between prosthetic devices. There was a significant interaction between speed and limb ($P = 0.0382$) such that the impulse decreased with increasing speed in all conditions except the prosthetic limb of the ESR condition.

For the EKFM (Fig. 2, Table 2) there were significant prosthetic × limb ($P = 0.0120$) and prosthetic × speed interactions ($P < 0.001$). As speed increased, the EKFM increased more in the sound limbs ($P < 0.001$) than the prosthetic limbs. At the fast speed, the EKFM was significantly less on the sound limb in the BiOM compared to the ESR ($P = 0.0190$, $\alpha = 0.0250$) and the sound limbs were greater than the prosthetic in both conditions (BiOM: $P = 0.0014$, $\alpha = 0.0167$, ESR: $P = 0.0001$, $\alpha = 0.0125$).

3.3. Peak ground reaction force (GRF)

The peak GRF was significantly affected by all three independent factors of prosthesis, speed and limb ($P = 0.0020$ for the three-way interaction) (Fig. 3, Table 2). First, comparisons between prosthetic devices indicated that the BiOM significantly reduced the sound limb peak GRF 7% at both the moderate ($P = 0.0043$, $\alpha = 0.0050$) and fast ($P = 0.0025$, $\alpha = 0.0042$) speeds, but did not reach the corrected level for statistical significance at the slowest speed ($P = 0.0175$, $\alpha = 0.0063$). At the fastest speed, the sound limb had a 12% greater peak GRF relative to the prosthetic limb in the ESR condition ($P = 0.0013$, $\alpha = 0.0042$); in the BiOM, sound and prosthetic limbs were not different.

3.4. Loading rate

There was a significant three-way interaction among prosthesis, limb, and speed for the loading rate ($P = 0.0037$) (Table 2). Loading rates in the sound limb were lower when participants used the BiOM than the ESR prosthesis at the moderate and fast speeds, but not slow speed (moderate: $P = 0.0003$, $\alpha = 0.0042$, fast: $P = 0.0028$, $\alpha = 0.0056$). Also, the loading rate in the sound side limb of the BiOM condition was significantly less than the prosthetic side at the moderate ($P = 0.0018$, $\alpha = 0.0050$) and fast ($P = 0.0010$, $\alpha = 0.0045$) speeds but there were no between limb differences in the ESR prosthesis. The interaction was next separated to compare differences across speeds; loading rate increased more in the sound limb of the BiOM than in the ESR with increasing speed.

3.5. Comparison to controls

The sound limb in the BiOM had lower loading rates than controls at the two faster speeds (moderate: $P = 0.0007$, fast: $P = 0.0164$). Neither prosthetic condition exhibited a clear advantage over the other in terms of normalizing sound limb loading to controls for the peak EKAM, its impulse, or the peak GRF and values were not significantly different from controls. On the prosthetic ESR limb, the peak EKAM was 35% lower than controls at the slowest speed ($P = 0.0029$). At the two faster speeds, the sound limb's peak EKFM in the ESR condition was significantly greater than controls (moderate: $P = 0.0320$, fast: $P = 0.0020$).

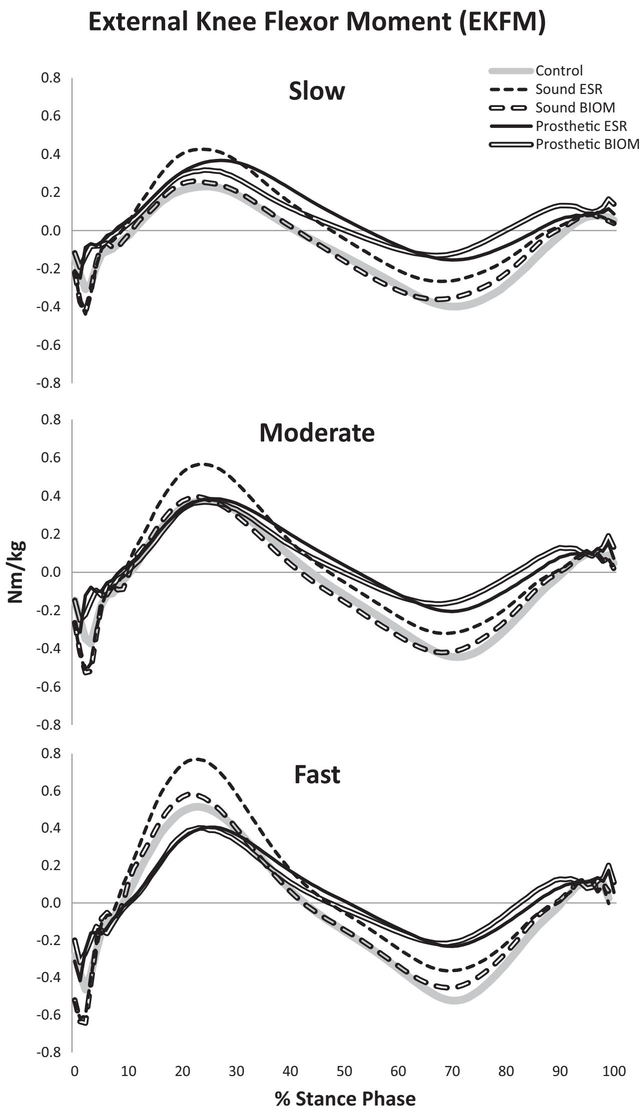

Fig. 2. Mean external knee flexor moment (EKFM) profiles across the stance phase.

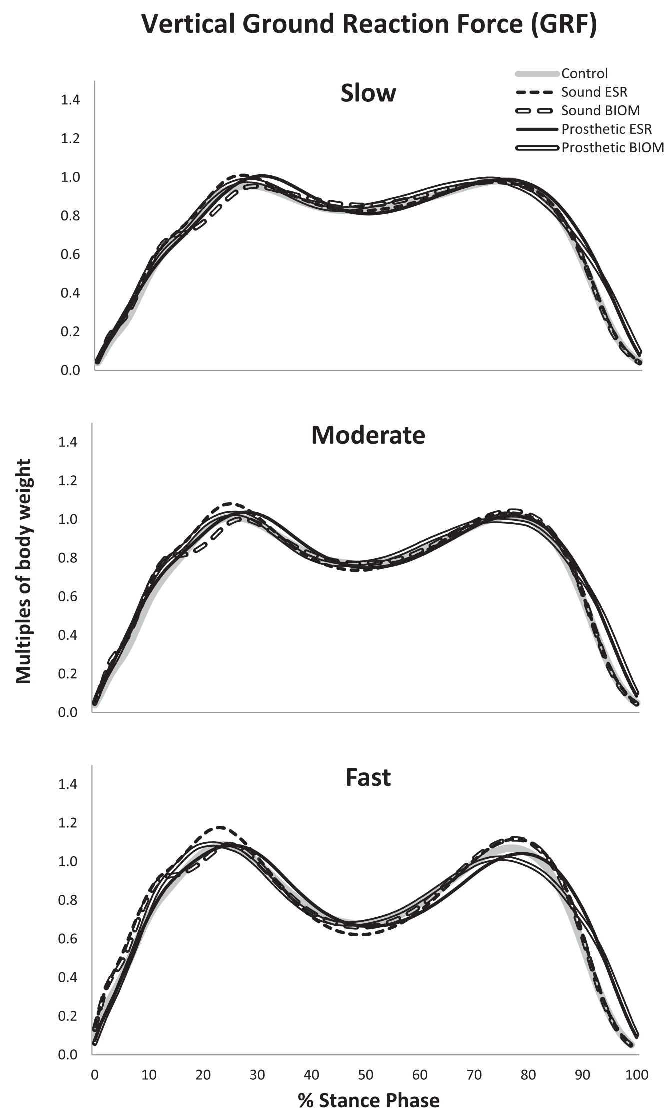

Fig. 3. Mean vertical ground reaction force (GRF) profiles across the stance phase.

4. Discussion

Individuals with TTA are at increased risk for knee OA disease development in their sound limb and this study investigated biomechanical risk factors associated with disease development. Specifically, this study compared risk factors between powered and passive ankle–foot prostheses, between sound and prosthetic limbs, and between individuals with TTA and able-bodied controls across a range of three walking speeds.

In partial support of the first hypothesis, the powered BiOM ankle–foot prosthesis reduced the sound limb's peak GRF, loading rate and peak EKFM, but only at faster speeds. Although these variables are all measures of loading and have been shown to relate to the risk of knee OA, they do not always directly relate to the peak EKAM (Creaby et al., 2013). It was expected that the peak EKAM and EKAM impulse would also decrease with the use of the powered device but this was not the case. Although Grabowski and D'Andrea (2013) also found no difference in the peak EKAM of the sound limb at similar slow and moderate speeds, they found a 20% decrease at a similar fast speed. It is likely that peak EKAMs and impulses on the sound limb did not decrease with the use of a powered ankle–foot prosthesis in the present study because they already fell within a normal range of values reported in previous literature from healthy, able-bodied subjects with no history of knee OA (e.g. Fey and Neptune, 2012; Royer and Koenig, 2005).

Contrary to the second hypothesis, individuals with TTA did not display a greater risk for knee OA in their sound limb than their prosthetic limb and, contrary to the third hypothesis, their risk was not greater than able-bodied individuals. This is not the first study to show that individuals with TTA do not consistently exhibit risk factors for knee OA in their sound limb. Fey and Neptune (2012) found that, at a similar slow speed, peak sound limb EKAMs and impulses were lower in individuals with TTA than controls and at similar moderate and fast speeds there were no differences between groups for either variable. Rueda et al. (2013) also found that peak EKAMs on the sound limb were not different from controls. A common factor in both of these studies and the present study is that the participants largely consisted of individuals with amputation due to trauma. Additionally, they may have been in relatively early stages of prosthetic use. In Fey and Neptune (2012), subjects averaged six years from amputation and in Rueda et al. (2013) inclusion criteria required at least one year of prosthetic use, although the authors did not specify mean time. However, Royer et al. (2005) studied long-time prosthetic users (average 16.7 years) and found no significant differences between the sound limb of TTA and controls for proximal tibia bone mineral density or the peak EKAM. In addition, Lloyd et al. (2010) included eight TTA who ranged in prosthetic use time from six months to 23 years and found no difference between the peak sound limb EKAM in the TTA and controls. While the peak EKAM and its impulse are widely recognized risk factors for knee OA in able-bodied populations, they may not adequately relate to the risk in individuals with TTA during walking.

Instead, Lloyd et al. (2010) suggested that strength asymmetries about the knee joint may relate to the risk for knee OA in individuals with TTA as greater asymmetries resulted in greater loading rates. The BiOM effectively reduced the loading rate in the sound limb as speed increased but increased the loading rate on the prosthetic side. This increase in the loading rate may not be problematic as there is no recognized risk of knee OA in the prosthetic limbs of individuals with TTA and all other biomechanical risk factors were not different between limbs in the BiOM condition. The greater loading rates on the sound limb in the ESR compared to the BiOM may be attributed partly to the greater peak GRF on the sound limbs in the ESR condition. GRFs were similar to those reported on passive and powered ankle–foot prosthetic users (Grabowski and D'Andrea, 2013; Sanderson and Martin, 1997), and displayed the initial impact peak on the sound limb observed in Sanderson and Martin (1997).

Direct comparisons of these data to other previous studies on knee OA may be limited due to the comparatively younger population included in the present study. However, knee OA is not limited to the elderly. By age 40, the majority of individuals will see some evidence of cartilage deterioration and OA can develop in adults of all ages within 10 years of a major joint injury (Roos et al., 1995). There are, however, preventative measures individuals at risk for OA can take. Muscular strength, particularly quadriceps strength, stabilizes the knee joint and protects the articular surfaces from high force loads (Cicuttini et al., 2004; Mikesky et al., 2006; Slemenda et al., 1998). General physical activity may also have a protective effect on knee cartilage (White et al., 1993). While specific measures of physical activity and function were not recorded, the service members involved in this study participated in rigorous physical therapy and were active and fit prior to injury. They had also attained self-selected walking velocities that were comparable to able-bodied individuals, which may be a functional indicator of recovery in this population (Baker and Hewison, 1990).

Although the powered prosthesis did not yield dramatic reductions in limb loading across all variables, small changes may be important. Interventions that introduce small changes in limb loading (5–7%) have been argued to be clinically relevant over the accumulation of thousands of repetitions of loading throughout a typical day (Kerrigan et al., 2002). At the moderate speed we found a 16% decrease in the peak EKAM and, although this difference did not reach statistical significance at $P = 0.054$, it may be important when considering other factors. For example, weight gain is common following amputation (Rosenberg et al., 2013) and increases the risk of knee OA (Messier et al., 2005), but small reductions in the peak EKAM with a powered prosthesis may offset a portion of its expected increase with increases in mass. If the small changes provided by the BiOM can offset other factors associated with amputation, individuals with TTA may be able to mitigate their overall risk for disease development and offer the sound limb a small protective effect over time. At the time of the study participants did not display biomechanical risk factors associated with knee OA but it is possible that years of prosthetic use may result in gait mechanics that increase the risk for disease development. However, the time course of knee OA onset in an initially asymptomatic population of young individuals with TTA is currently unknown.

A limitation to this study was that baseline cartilage volume and knee alignment radiographs were not taken to identify current evidence of OA. It was expected that, due to the young age of the subjects, this study compared individuals with amputations who were otherwise healthy. In addition, other risk factors for knee OA (e.g. obesity (Felson et al., 1988) race (Spector and MacGregor, 2004)) were not considered as part of this within-subject comparison. Also, the prevalence of medial and lateral knee OA has not been widely reported in an amputee population (Melzer et al., 2001) and the peak EKAM is only related to medial compartment loading. The authors can only assume greater medial knee OA rates as previous research indicates greater and more prevalent risk factors (Lloyd et al., 2010; Melzer et al., 2001; Royer and Koenig, 2005; Royer and Wasilewski, 2005). Lastly, only level ground walking was investigated. Other activities, such as hopping, which is commonly used when moving without the prosthesis, may also contribute to the development of OA.

5. Conclusions

Young, active individuals with TTA did not display significantly greater biomechanical risk factors for knee OA than able-bodied individuals during walking. Surrogate measures of knee joint loading were within the ranges of reports from the literature on healthy, able-bodied individuals with no history of OA. In addition, the sound knee did not display an overall greater risk of knee OA development than the prosthetic knee. The powered ankle–foot prosthesis reduced the peak GRF, loading rate and peak EKFM as speed increased, but did not reduce the peak EKAM or its impulse. The improvement in some of the factors of limb loading with the powered prosthesis indicates that it may be of value for the prophylactic reduction of risk factors linked to disease development. These preventative strategies may be particularly important as patients transition away from the daily regimen of the controlled rehabilitation environment. Continued research into the etiology of disability and the development of preventative strategies to mitigate risk factors are essential for maintaining the health of the intact limb and maximizing quality of life for individuals with amputations.

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