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Vacuum level effects on knee contact force for unilateral transtibial amputees with elevated vacuum suspension

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ABSTRACT

The elevated vacuum suspension system (EVSS) has demonstrated unique health benefits for amputees, but the effect of vacuum pressure values on knee contact force (KCF) is still unclear. The objective of this study was to investigate the effect of vacuum levels on KCF for unilateral transtibial amputees (UTA) using the EVSS. Three-dimensional gait was modeled for 9 UTA with five vacuum levels (0–20 inHg [67.73 kPa], 5 inHg [16.93 kPa] increments) and 9 non-amputees based on kinematic and ground reaction force data. The results showed that the vacuum level effects were significant for peak axial KCF, which had a relatively large value at 0 and 20 inHg (67.73 kPa). The intact limb exhibited a comparable peak axial KCF to the non-amputees at 15 inHg (50.79 kPa). At moderate vacuum levels (5 inHg [16.93 kPa] to 15 inHg [50.79 kPa]), co-contraction of quadriceps and hamstrings at peak axial KCF was similar for the intact limb, but was smaller for the residual limb comparing with the non-amputees. The intact limb showed a similar magnitude of quadriceps and hamstrings force at 15 inHg (50.79 kPa) to the non-amputees, but the muscle coordination patterns varied between the residual and intact limbs. These findings indicate that a proper vacuum level may partially compensate for the lack of ankle plantarflexor and reduce the knee loading. Of the tested vacuum levels, 15 inHg (50.79 kPa) appears most favorable, although additional analyses with more amputees are suggested to confirm these results prior to establishing clinical guidelines.

1. Introduction

There are an estimated 600,000 people with lower-limb amputations in the United States and approximately 80% of these are transtibial amputation (McGimpsey and Bradford, 2008). Previous studies report that unilateral transtibial amputees (UTA) show a shorter step length, slower speed and cadences when compared to non-amputees (Sagawa et al., 2011; Schmalz et al., 2002). Additionally, the residual limb displays smaller ankle sagittal motion due to the lack of plantarflexion, but a larger knee sagittal motion to reduce the possibility of a fall (Greenland, 2012; Sagawa et al., 2011). Gait asymmetries are common in UTA and are evident by a relatively longer step length, stance phase and single support duration on the intact limb than residual limb (Greenland, 2012; Isakov et al., 2000).

The biomechanical factors related to use prosthesis are considered to be one factor associated with osteoarthritis (OA) in amputees (Gailey et al., 2008; Struyf et al., 2009). In UTA, the intact limb showed a larger vertical ground reaction force (GRF) and greater peak adduction moments in the hip and knee than the residual limb (Greenland, 2012; Lloyd et al., 2010; Sagawa et al., 2011). A significantly higher risk to develop joint disorders, such as knee OA, was found in the intact limb relative to the residual limb and non-amputees, which is partially attributed to increased joint loading and moments (Norvell et al., 2005; Struyf et al., 2009).

Noninvasive measurements of knee contact forces (KCF) are currently impossible. Computational modeling has been used to determine the KCF during gait for different population and commonly reported peak axial KCF ranges from 1.8 to 6.7 body weight (BW) (Fregly et al., 2012; Kim et al., 2009; Richards and Higginson, 2010; Taylor et al., 2004; Winby et al., 2009). Relatively few studies investigated the KCF in amputee gait. A study conducted by Silverman and Neptune (2014) reported that the axial peak KCF was about 3.1 and 4.1 BW for the residual and intact limbs

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respectively using musculoskeletal simulation for 14 UTA with varied prosthesis.

The elevated vacuum suspension system (EVSS) have demonstrated positive effects for amputees providing a better fitting socket and a superior linkage between the residual limb and prosthesis. Researchers report that the EVSS could effectively reduce the vertical pistoning in the socket, increase the rotational stability and prevent volume loss of the residual limb (Board et al., 2001; Goswami et al., 2003; Klute et al., 2011). All of these effects improve the force transfer to the prosthesis and help reduce the compensation from the intact limb which is beneficial to the gait symmetry (Sagawa et al., 2011).

Although the number of amputees using the EVSS is growing rapidly, few research focused on the effect of different vacuum pressure settings including what is a preferred vacuum level. To our knowledge, only one study investigated the effect of 10 inHg (33.86 kPa) and 15 inHg (50.79 kPa) vacuum levels with a single UTA using the EVSS and found similar percent changes in residual limb volume in these two levels (Gerschutz et al., 2010). Therefore, the objective of this study was to investigate the effect of vacuum levels on KCF in both residual and intact limbs for UTA using muscle-driven gait simulations. The knowledge gained may benefit amputees and clinicians to better understand the effect of vacuum level on amputee gait and help prevent early onset of knee OA for amputees.

2. Methods

2.1. Experimental data

Previously collected experimental kinematic and GRF data were used (Greenland, 2012). Briefly, the data were collected from 9 male UTA using the EVSS and 9 male non-amputees while walking over ground at 1.20 to 1.40 m/s (Table 1). The non-amputee group was younger on average than the amputee group, but were not significantly different in height, weight, and BMI. All amputees had dynamic response prosthetic feet and were free from musculoskeletal disorders and leg pain, and did not require assistive devices for walking. All non-amputees were free from limb injuries or other disorders which would affect their gait. Institutional review board approval was obtained and all participants signed an informed consent document. Five trials were collected for UTA at each of five vacuum levels (0 –20 inHg [67.73 kPa], 5 inHg [16.93 kPa] increments) and also for the non-amputees. Kinematics and GRF data were collected at 100 Hz and 1000 Hz with low-pass filter at 6 Hz and 20 Hz respectively.

2.2. Musculoskeletal model and gait simulation

The musculoskeletal model used in this study consisted of 12 rigid body segments, 23 degrees of freedom (DOF) and 54 muscles (Fig. 1). Body segments included a right and left thigh, shank, talus, calcaneus and toe, as well as a head-torso segment and pelvis. 23 DOFs included 6 DOFs between pelvis and ground, 3 DOFs between torso and pelvis, 3 DOFs for the hip joint, and 1 DOF for the ankle, subtalar and metatarsalphalangeal joints respectively. The knee was represented as a 1 DOF joint with anteroposterior translation occurring as a function of knee flexion.

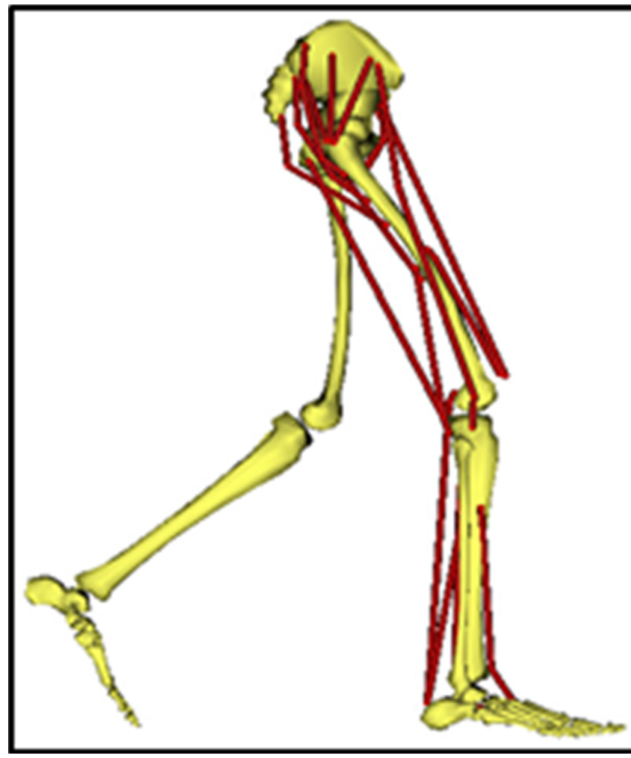

Fig. 1. The lower limb muscles in the musculoskeletal model. The muscle groups included in the model were GMAX (gluteus maximus and adductor magnus), GMED (anterior, middle and posterior portions of gluteus medius), IL (psoas and iliacus), VAS (three vastii muscles), RF (rectus femoris), HAM (medial hamstrings, gracilis, biceps femoris long head), BFsh (biceps femoris short head), SAR (sartorius), TFL (tensor fascia lata), QF (quadratus femoris and pectineus), SOL (soleus and tibialis posterior), GAS (medial and lateral gastrocnemius), TA (tibialis anterior).

OpenSim (Delp et al., 2007) was used to generate the muscle-driven gait simulation. The model was first scaled to match each subject. Segment lengths were scaled based on the marker position from the static trial. The segment masses were scaled by the subject's body mass and anthropometric data. Computed muscle control algorithm was used to compute muscle excitations and forces to drive the musculoskeletal model towards the experimental kinematics (Thelen and Anderson, 2006; Thelen et al., 2003). In order to model UTA, the muscles crossing the ankle were removed. The mass and inertial properties of the shank were modified for the residual limb and the ankle joint torque was modeled using a second-order dampened torsional spring to replicate ankle torque obtained from inverse dynamics, which were similar to previous research (Silverman and Neptune, 2012).

2.3. Calculation of KCF and co-contraction index

KCF were determined using the vector sum of the knee reaction force and the force from muscles crossing the knee joint, which

Table 1
Mean (standard deviation) amputee and non-amputee group characteristics.

	Age (years)	Height (m)	Weight (kg)	BMI (kg/m^2)	Time since amputation (years)	Time since using EVSS (years)	Etiology
Amputees (n = 9)	51.1 (16.1)	1.83 (0.06)	94.8 (12.1)	28.3 (4.0)	14.4 (11.4)	5.2 (3.1)	5 Traumatic 4 Vascular and others
Non-amputees (n = 9)	27.8 (3.7)	1.80 (0.05)	82.9 (17.7)	25.5 (4.8)	N/A	N/A	N/A

was quantified by the peak value and expressed on the tibia in the anteroposterior (AP), axial and mediolateral (ML) directions.

Muscle co-contraction, expressed as co-contraction index (CCI), was used to describe the simultaneous activity of antagonist-agonist muscle groups crossing the knee. CCI was calculated at the time where peak axial KCF occurred using Eq. (1) (Karakostas et al., 2003).

$$CCI = \frac{\sum F^M_{Antagonist}}{\sum F^M_{Agonists}} \quad (1)$$

$\sum F^M_{Antagonist}$ represented the total force of antagonist muscle groups, which were quadriceps (RF and VAS) during knee flexed and $\sum F^M_{Agonist}$ represented the total force of agonist muscle groups, which were hamstrings (HAM and BFsh) during knee flexed. Additionally, the muscle forces were decomposed into orthogonal components and the component along the long axis of the tibia to evaluate the contributions of hamstrings and quadriceps to the axial KCF.

The peak KCF and muscle forces were normalized by the subject's body weight. For UTA, the variables were averaged across trials for each subject at each vacuum level. For the non-amputees, the variables were averaged across five within-subject trials and then across subjects to obtain group-averaged data.

2.4. Statistical analysis

Statistics were performed using SPSS 20.0 (IBM Corporation, Armonk, NY). Dunnett's t-test was used to compare the intact and residual limbs at each vacuum level with the non-amputees. Two-factor repeated ANOVA was used to determine the effects of vacuum level and limb within UTA. If the assumption of sphericity was violated, the Greenhouse-Geisser correction was used. Post-hoc tests were performed using the Bonferroni adjustment for multiple comparisons. The results were considered significant when $p < 0.05$.

3. Results

3.1. Peak knee contact force

A significant vacuum level effect ($F(4,32) = 11.36$, $p < 0.001$, $\eta_p^2 = 0.694$) was found for the peak axial KCF (Table 2, Fig. 2a), Post-hoc test indicated that peak axial KCF was smaller at 15 inHg (50.79 kPa) compared with 5 inHg (16.93 kPa) ($p = 0.033$). Additionally, the change in vacuum level affected the peak axial KCF similarly across all nine amputees (see Supplementary Material). UTA showed a relatively large peak axial KCF at 0 and 20 inHg (67.73 kPa), but a decreased peak axial KCF from 5 inHg (16.93 kPa) to 15 inHg (50.79 kPa). No significant limb effect and interaction effect were found ($p = 0.057$ and 0.420, respectively). When compared with the non-amputees, UTA showed a larger peak axial KCF at 0 inHg ($p < 0.001$), 5 inHg (16.93 kPa) ($p = 0.006$), 10 inHg (33.86 kPa) ($p = 0.014$) and 20 inHg (67.73 kPa) ($p < 0.001$) on the intact limb and a smaller peak axial KCF at 15 inHg (50.79 kPa) ($p = 0.028$) on the residual limb.

For the peak AP KCF, the limb effect was significant ($F(1,8) = 27.80$, $p = 0.003$, $\eta_p^2 = 0.848$) and the magnitude was larger on the intact limb than residual limb (Table 2 and Fig. 2b). The vacuum level effect and interaction effect were not significant ($p = 0.267$ and 0.562, respectively). Additionally, the peak AP KCFs were also larger on the intact limb than the non-amputees at 5 inHg (16.93 kPa) ($p = 0.049$), 10 inHg (33.86 kPa) ($p = 0.038$) and 15 inHg (50.79 kPa) ($p = 0.048$, Table 2). For the peak ML KCF, no significant vacuum level and limb effects were found ($p = 0.742$ and 0.977, respectively) and the values at all vacuum levels were similar to non-amputees on intact and residual limbs (all $p > 0.05$) (Table 2, Fig. 2c).

3.2. Muscle co-contraction at the peak axial KCF

A significant limb effect was found for quadriceps-hamstrings co-contraction at the peak KCF ($F(1,8) = 15.73$, $p = 0.011$, $\eta_p^2 = 0.759$) and CCI was smaller on the residual limb than the intact limb (Table 3). No significant vacuum level effect and interaction effect were found ($p = 0.115$ and 0.073, respectively). Additionally, CCI was smaller on the intact limb at 20 inHg (67.73 kPa) ($p = 0.045$) and also smaller on the residual limb at 5 inHg (16.93 kPa) ($p = 0.022$), 10 inHg (33.86 kPa) ($p = 0.048$), 15 inHg (50.79 kPa) ($p = 0.031$) and 20 inHg (67.73 kPa) ($p = 0.020$) when compared with non-amputees.

For the force of muscle groups at the peak axial KCF, limb effect and vacuum level effect were significant for quadriceps ($F(1,8) = 27.80$, $p = 0.013$, $\eta_p^2 = 0.743$) and hamstrings ($F(4,32) = 5.70$, $p = 0.003$, $\eta_p^2 = 0.533$) respectively (Table 3). Post-hoc tests showed that quadriceps force was larger in the intact limb than residual limb, and hamstrings force was larger at 20 inHg (67.73 kPa) than 15 inHg (50.79 kPa) ($p = 0.035$). The vacuum level effect and the interaction effect were not significant for quadriceps ($p = 0.271$ and 0.391, respectively), and no significant limb effect and interaction effect were found for hamstrings ($p = 0.349$ and 0.066, respectively). When compared with the non-amputees, UTA showed a larger hamstrings force at 20 inHg (67.73 kPa) in the intact limbs ($p = 0.039$, Table 3), but no differences existed for quadriceps force between non-amputees and UTA at any vacuum levels (all $p > 0.05$).

3.3. Muscle contribution to the peak axial KCF

When comparing muscle contribution to the peak axial KCF, a significant limb effect existed for quadriceps ($F(1,8) = 10.95$, $p = 0.021$, $\eta_p^2 = 0.686$) which contributed more in the intact limb and also for hamstrings ($F(1,8) = 7.07$, $p = 0.045$, $\eta_p^2 = 0.586$) which contributed more in the residual limb (Fig. 3a). The vacuum level effect and interaction effect were not significant for both muscle

Table 2
Mean (standard deviation) peak KCF in BW for the intact and residual limbs and non-amputee. Positive was defined in the anterior, inferior and medial direction on the tibia.

Vacuum level	0 inHg		5 inHg		10 inHg		15 inHg		20 inHg		Non-amputee
	Intact	Residual	Intact	Residual	Intact	Residual	Intact	Residual	Intact	Residual	
Axial‡	4.40 (0.29)*	3.78 (0.46)	3.96 (0.25)*	3.54 (0.39)	3.94 (0.27)*	3.35 (0.47)	3.75 (0.27)	3.19 (0.47)*	4.40 (0.56)*	4.04 (0.53)	3.52 (0.38)
Anteroposterior †	1.77 (0.54)	0.88 (0.44)	1.79 (0.46)*	0.75 (0.51)	1.80 (0.44)*	0.95 (0.44)	1.68 (0.33)*	0.89 (0.33)	1.77 (0.38)	1.05 (0.57)	1.16 (0.46)
Mediolateral	0.27 (0.03)	0.21 (0.06)	0.27 (0.04)	0.22 (0.08)	0.26 (0.04)	0.22 (0.07)	0.27 (0.03)	0.21 (0.09)	0.26 (0.03)	0.21 (0.09)	0.23 (0.02)

† Significant limb effect in amputee group ($p < 0.05$).
‡ Significant vacuum level effect in amputee group ($p < 0.05$).
* Significant difference with non-amputees ($p < 0.05$).

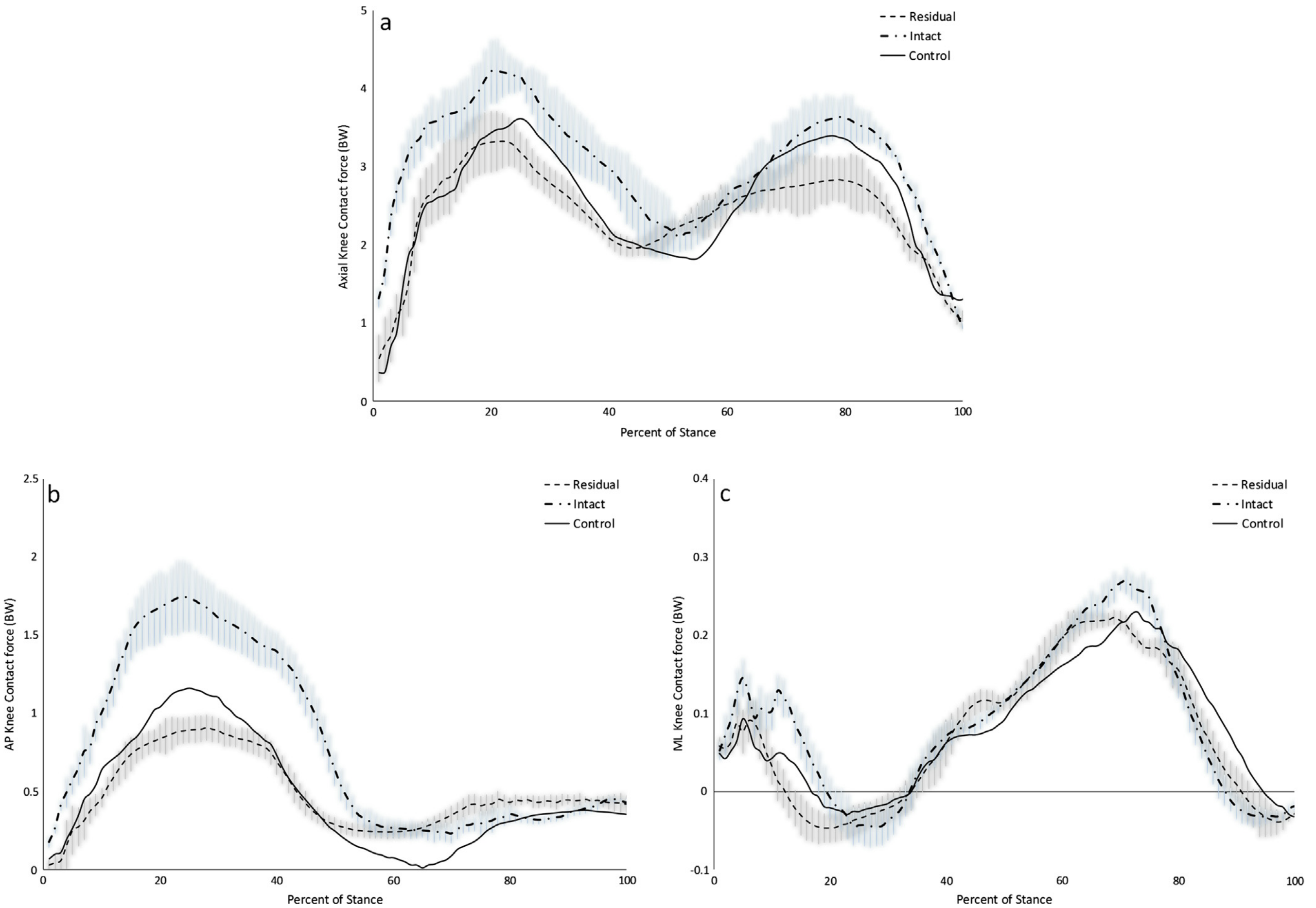

Fig. 2. Representative average KCF curves during stance phase in the axial (a), anteroposterior (b) and mediolateral (c) directions from one UTA and non-amputee. The shaded regions indicated one standard deviation.

Table 3
Mean (standard deviation) CCI and muscle force in BW at the peak axial KCF the intact and residual limbs and non-amputee.

	0 inHg		5 inHg		10 inHg		15 inHg		20 inHg		Non-amputee
	Intact	Residual	Intact	Residual	Intact	Residual	Intact	Residual	Intact	Residual	
CCI†	1.88 (1.02)	1.58 (1.18)	2.47 (0.89)	0.95 (0.54)*	2.41 (1.63)	0.94 (0.56)*	2.66 (1.17)	0.96 (0.49)*	1.20 (0.94)*	0.92 (0.45)*	2.93 (1.79)
Quadriceps (BW)†	2.07 (0.63)	1.33 (0.22)	2.00 (0.41)	1.11 (0.46)	1.99 (0.51)	1.09 (0.41)	1.64 (0.82)	1.03 (0.40)	1.65 (0.71)	1.31 (0.45)	1.57 (0.44)
Hamstrings (BW)‡	1.29 (0.45)	1.13 (0.52)	0.90 (0.33)	1.39 (0.55)	1.06 (0.50)	1.30 (0.33)	0.80 (0.42)	1.13 (0.17)	1.69 (0.58)*	1.54 (0.39)	0.90 (0.49)

† Significant limb effect in amputee group ($p < 0.05$).
‡ Significant vacuum level effect in amputee group ($p < 0.05$).
* Significant difference with non-amputees ($p < 0.05$).

groups ($p = 0.212$ and 0.217 for quadriceps, $p = 0.082$ and 0.074 for hamstrings). Additionally, there were no significant quadriceps and hamstrings contribution difference between non-amputees and UTA at any vacuum levels (all $p > 0.05$).

4. Discussion

The predicted muscle forces for amputees and non-amputees in this study showed agreement in pattern and magnitude compared with previous research using a similar musculoskeletal model (Richards and Higginson, 2010; Sasaki and Neptune, 2010; Silverman and Neptune, 2014). The average residual force and moment were 7.2 N and 19.8 Nm, which was less than the threshold values (10 N and 30 Nm) of a "good" CMC results for gait simulation. These results indicated that the present model is robust for simulation of UTA gait and estimation of peak KCF (Hicks et al., 2015).

The axial KCF had two major peak forces which occurred in early stance and late stance (Fig. 2a). This pattern was consistent with previous studies (Kim et al., 2009; Lin et al., 2010; Sasaki and Neptune, 2010; Silverman and Neptune, 2014). The first peak axial KCF was investigated in our study, but the second peak was not due to the lack of plantarflexors in the residual limb, which were the main contributor to the second peak in non-amputees (Anderson and Pandy, 2003; Sasaki and Neptune, 2010). The peak axial and AP KCFs on the intact limb were generally larger than the residual limb and non-amputees (Fig. 2a and b). This finding aligned with previous conclusions that UTA had an increased risk of developing knee OA on their intact limb (Lloyd et al., 2010; Sagawa et al., 2011).

At the vacuum level 15 inHg (50.79 kPa), the intact limb showed a comparable peak axial KCF (3.75 BW) with non-amputees (3.52 BW). Since excessive joint loading is a major factor in OA development, lower peak axial KCF may help reduce the

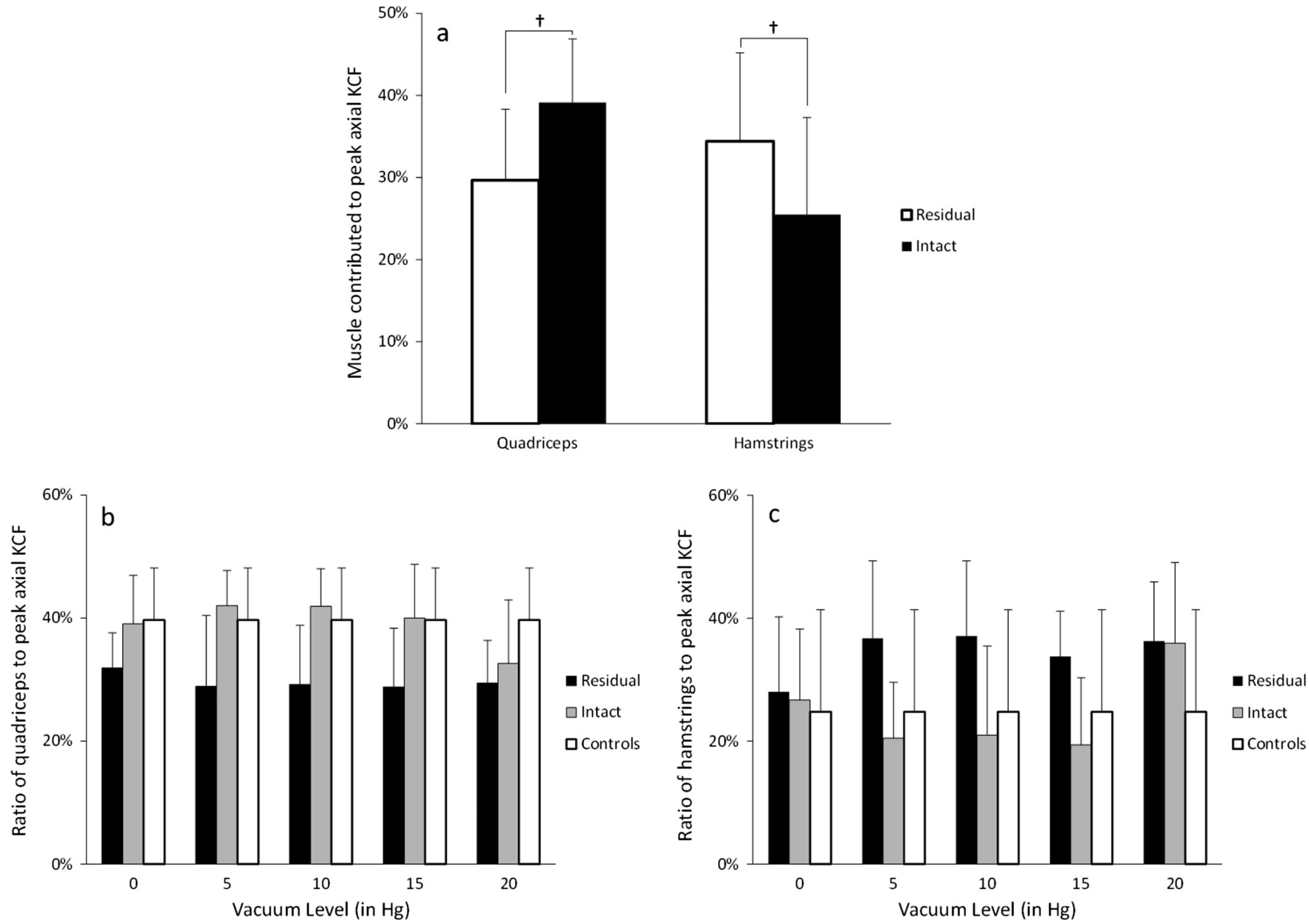

Fig. 3. Quadriceps and hamstrings contribute to the peak axial KCF by limb (a) and by vacuum level (b and c). † represented significant limb effect within UTA.

prevalence of knee OA in the intact limb. The intact and residual limbs showed a similar pattern in peak axial KCF at different vacuum levels, which indicated that UTA may use a whole body coordination strategy in response to a vacuum change.

In the extreme high vacuum level, 20 inHg (67.73 kPa), both intact and residual limbs showed an increased peak axial KCF. A possible explanation was the significantly increased peak external knee adduction torque at 20 inHg reported by previous study using the same data (Greenland, 2012). Since the positive correlation between external knee adduction torque and medial compartment knee loading exists (Erhart et al., 2010; Zhao et al., 2007b), a higher peak knee adduction torque could increase medial peak axial KCF and further for the peak axial KCF. Additionally, the extreme vacuum level affects pump energy efficiency (Gerschutz et al., 2011; Greenland, 2012), which could also increase the peak axial KCF for amputees.

Co-contraction of the quadriceps and hamstrings at peak axial KCF was lower in the residual limb than intact limb, especially at the moderate vacuum levels (5 inHg [16.93 kPa] to 15 inHg [50.79 kPa] in Table 3). This was caused by the smaller quadriceps force and larger hamstrings force on the residual limb compared to the intact limb, which was consistent with previous findings showing significant difference in VAS and RF between residual and intact limbs (Silverman and Neptune, 2012). The muscles in the upper leg atrophy in the residual limb (Moirenfeld et al., 2000) and UTA walk with reduced knee flexion on the residual limb during early stance (Greenland, 2012), which provide a possible explanation of the lower quadriceps forces on the residual limb. The hamstrings and gastrocnemius are the primary source of knee flexion in able-bodied gait. Increased hamstrings force may partially compensate for the lack of plantarflexor on the residual limb. The co-contraction pattern was comparable in the intact limb at moderate vacuum levels to the non-amputees. A high degree of co-contraction could help systematic distribution of compression forces, reduce external impact force and improve joint stability (Baratta et al., 1988; Jarić et al., 1995). Therefore, a moderate vacuum level may help reduce the risk of knee OA in the intact limb.

The magnitude of forces generated from quadriceps and hamstrings at 15 inHg (50.79 kPa) on the intact limb were closest to the forces generated by the non-amputees (Table 3). Larger hamstrings force at 20 inHg (67.73 kPa) and quadriceps force at 0 to 10 inHg (33.86 kPa) in the intact limb were observed comparing with non-amputees, which may increase the risk of muscle fatigue and joint injury during gait (Wade et al., 2010).

The contribution of quadriceps and hamstrings to the peak axial KCF was insensitive to the vacuum levels (Fig. 3b and c), but different contribution rate between these two muscle groups were observed on the intact and residual limbs (Fig. 3a). The vastii muscles were the primary contributors to the peak axial KCF, but controversial results exist for the contribution from other muscles, such as rectus femoris and hamstrings (Richards and Higginson, 2010; Sasaki and Neptune, 2010; Silverman and Neptune, 2014). Some of these differences are likely caused by the different musculoskeletal models and optimization algorithms. In our study, the contribution rates of quadriceps and hamstrings were quite similar between the intact limb and non-amputee, except for 20 inHg

(67.73 kPa). Although the prosthesis functioned similarly to the non-amputee soleus and partially compensated for the lack of gastrocnemius (Silverman and Neptune, 2012, 2014), the changed pattern of muscle groups still implied the hamstrings compensation on the residual limb. These results provide guidance about where to focus efforts to improve the design of EVSS and rehabilitation strategies for UTA.

The predicted KCF magnitudes in this study are larger than prior in vivo measurements, but consistent with prior simulation studies. The peak axial KCF reported by using an instrumented implant ranged from 1.8 to 3.0 BW for self-selected speed gait (Fregly et al., 2012; Kutzner et al., 2011). These in vivo loads are smaller than our computed results. However, the in vivo results cannot be extrapolated to larger populations due to the small sample size in the literature and different knee joint conditions. Research indicated the ability of using musculoskeletal models to predict physiological loading in the knee joint and the predicted peak axial KCF in our study are within the ranges of estimated values (2.7–4.4 BW) from previous musculoskeletal modeling (Kim et al., 2009; Silverman and Neptune, 2014; Taylor et al., 2004; Winby et al., 2009).

Vacuum level at 15 inHg (50.79 kPa) associates with a decreasing rate of volume fluctuation compared with 0 and 10 inHg (33.86 kPa) (Gerschutz et al., 2010). Self-report indicated amputees preferred higher vacuum levels, with most choosing 14 or 15 inHg (50.79 kPa) (Gerschutz et al., 2011). Our results agreed with these findings and showed the peak KCF and muscle coordination pattern on the intact limb was most similar to the non-amputees at 15 inHg (50.79 kPa). This may explain by the improvement of amputees' proprioception since UTA lack the ability to distinguish between different vacuum levels, especially below 14 inHg (47.41 kPa) (Gerschutz et al., 2011). The extreme high vacuum level (20 inHg [67.73 kPa]) should be considered carefully since the excessive pressure usually caused volume loss and improper fitting between the limb and socket (Street, 2006).

Several limitations existed in this study. First, some present results indicated notable changes but did not reach statistical significance, partially due to the small sample size. The varied individual movement strategies among UTA also reduces the ability of detecting these differences (Silverman et al., 2008). A larger sample size is needed to generalize the present results to UTA population with EVSS. Second, the knee joint was constrained as a single flexion-extension joint, but knee motions in the frontal and transverse planes also exists during gait though the motion ranges are relatively small (Xu et al., 2015). The lack of knee motions in these two planes may alter the muscle excitation pattern and underestimate the KCF (Glitsch and Baumann, 1997; Xiao and Higginson, 2008). Additionally, the distribution of knee loads on the medial and lateral compartments were not addressed in this study. The medial compartment load is reported 1.5–3.6 times greater than the lateral compartment in non-amputees (Mundermann et al., 2008; Shelburne et al., 2006; Zhao et al., 2007a). Further information about KCF distribution could help reveal the mechanisms of knee OA in UTA. Finally, our model did not include knee ligaments, which may generate force ranging from 0.2 to 0.7 BW during gait (Shelburne et al., 2006; Xu et al., 2014). A more detailed knee model may help study any change of muscle forces and the contribution of ligaments to the KCF.

5. Conclusions

This study investigated the effect of vacuum levels on KCF for UTA using the EVSS. The peak axial KCF was significantly affected by the vacuum level, which increased at the extreme vacuum levels (0 and 20 inHg [67.73 kPa]). The peak axial KCF on the intact limb was larger than the non-amputees except for 15 inHg (50.79 kPa). For the muscle co-contraction at peak axial KCF, the intact limb showed an increased CCI than the residual limb, but had a similar magnitude compared with the non-amputees except for 20 inHg (67.73 kPa). The hamstrings forces was significantly affected by the vacuum level, which was larger at 20 inHg (67.73 kPa) than 15 inHg (50.79 kPa). The quadriceps forces on the intact limb were most similar to the non-amputees at 15 inHg (50.79 kPa). Additionally, the pattern of muscle contribution to the peak axial KCF was also different on the residual limb than the intact limb.

The extreme vacuum levels (0 and 20 inHg [67.73 kPa]) is not suggested due to the large KCF, which may increase gait asymmetry and the risk of knee OA in the intact limb. The moderate vacuum levels (5 inHg [16.93 kPa] to 15 inHg [50.79 kPa]) are more favorable for the intact limb to generate muscle co-contraction patterns similar to non-amputees. Additionally, a proper vacuum setting may partially compensate for the lack of ankle plantarflexor and reduce the force required from hamstrings in the residual limb. While a specific "ideal" vacuum level was not determined, a moderate vacuum level at 15 inHg (50.79 kPa) is suggested based on the results of this study. Additional analyses with a larger sample size are needed to support these findings and to develop clinical guidelines for the ideal vacuum level with EVSS for UTA.

Conflict of interest statement

None.

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Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.jbiomech.2017.04.013.

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