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Three-dimensional knee joint contact forces during walking in unilateral transtibial amputees

Anne K. Silverman [a,*], Richard R. Neptune [b]

[a] Department of Mechanical Engineering, Colorado School of Mines, 1500 Illinois Street, Golden, CO 80401, USA
[b] Department of Mechanical Engineering, The University of Texas at Austin, Austin, TX 78712, USA

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ABSTRACT

Individuals with unilateral transtibial amputations have greater prevalence of osteoarthritis in the intact knee joint relative to the residual leg and non-amputees, but the cause of this greater prevalence is unclear. The purpose of this study was to compare knee joint contact forces and the muscles contributing to these forces between amputees and non-amputees during walking using forward dynamics simulations. We predicted that the intact knee contact forces would be higher than those of the residual leg and non-amputees. In the axial and mediolateral directions, the intact and non-amputee legs had greater peak tibio-femoral contact forces and impulses relative to the residual leg. The peak axial contact force was greater in the intact leg relative to the non-amputee leg, but the stance phase impulse was greater in the non-amputee leg. The vasti and hamstrings muscles in early stance and gastrocnemius in late stance were the largest contributors to the joint contact forces in the non-amputee and intact legs. Through dynamic coupling, the soleus and gluteus medius also had large contributions, even though they do not span the knee joint. In the residual leg, the prosthesis had large contributions to the joint forces, similar to the soleus in the intact and non-amputee legs. These results identify the muscles that contribute to knee joint contact forces during transtibial amputee walking and suggest that the peak knee contact forces may be more important than the knee contact impulses in explaining the high prevalence of intact leg osteoarthritis.

1. Introduction

Individuals with unilateral transtibial amputations have altered gait mechanics and muscle coordination patterns relative to non-amputees (e.g., Fey et al., 2010; Silverman and Neptune, 2012), which may lead to the onset of joint disorders with prolonged use. For example, transtibial amputees have an increased prevalence and early onset of osteoarthritis (OA) and pain in the intact leg knee joint relative to the residual leg and non-amputees (Burke et al., 1978; Lemaire and Fisher, 1994; Melzer et al., 2001; Norvell et al., 2005; Struyf et al., 2009). However, the biomechanical mechanisms that contribute to the increased prevalence remain unclear.

The etiology of OA is not completely understood, but is partially attributed to increased and/or atypical joint loading (Maly, 2009; Morgenroth et al., 2012). Thus, the increased prevalence of OA in the intact knee of amputees may be a result of greater joint loading relative to the residual leg and non-amputees. Studies of amputee walking have shown elevated intact leg ground reaction forces (GRFs) and joint kinetics relative to the residual leg (Nolan et al., 2003; Royer and Koenig, 2005; Sanderson and Martin, 1997; Silverman et al., 2008) and non-amputee subjects (Nolan and Lees, 2000). Amputees also often have greater stance times on the intact leg relative to the residual leg (e.g., Isakov et al., 2000; Nolan et al., 2003), which may result in greater force impulses in the intact leg knee joint. Identifying differences in knee joint loading in amputees relative to non-amputees is important for understanding the potential biomechanical mechanisms that contribute to the high prevalence of OA in this population.

Recent work has investigated knee joint intersegmental forces across a range of walking speeds and found no significant differences between the intact and residual legs or between the intact and non-amputee legs (Fey and Neptune, 2012). However, inverse dynamics-based intersegmental forces often underestimate joint contact forces, as they do not account for the compressive forces from muscles (Zajac et al., 2002). In addition, altered muscle coordination patterns (Fey et al., 2010; Powers et al., 1998; Winter and Sienko, 1988) and increased co-contraction of the residual leg vasti and hamstring muscles (e.g., Culham et al., 1986; Isakov et al., 2001; Pinzur et al., 1991) in transtibial amputee walking likely influence the knee joint contact forces.

* Corresponding author. Tel.: +1 303 384 2162; fax: +1 303 273 3602.
E-mail address: asilverm@mines.edu (A.K. Silverman).

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A significant challenge to investigating joint contact forces is the extreme difficulty, if not impossibility, of measuring them *in vivo*. In contrast, musculoskeletal modeling and simulation provide an ideal framework to estimate joint contact forces and quantify the contributions of individual muscles during dynamic movements (e.g., Sasaki and Neptune, 2010; Shelburne et al., 2006; Zajac et al., 2003). The gastrocnemius and soleus muscles have been shown to be large contributors to the knee joint contact force in non-amputee walking (Lin et al., 2010; Sasaki and Neptune, 2010; Shelburne et al., 2006), which may lead to asymmetric knee loading in unilateral transtibial amputees because they no longer have the functional use of these muscles in the residual leg.

The objective of this study was to investigate differences in knee joint contact forces in both the residual and intact legs relative to non-amputees during steady-state walking using three-dimensional musculoskeletal models and forward dynamics simulations. We expected that the peak knee contact forces and stance phase force impulses would be greater in the intact knee relative to the residual leg and non-amputees. In addition, individual muscle contributions to the knee contact forces were quantified to identify the potential biomechanical mechanisms that may contribute to the increased prevalence of OA in the intact knee.

2. Methods

2.1. Experimental data collection

Previously-collected kinematic, GRF and electromyographic (EMG) data were used to generate the forward dynamics simulations (Fey et al., 2010; Silverman et al., 2008). Briefly, the data were collected from 14 individuals with transtibial amputation and 10 non-amputees walking overground at 1.2 ± 0.06 m/s (Table 1). Subjects provided informed consent to participate in the experimental protocol approved by an Institutional Review Board. Kinematic data were collected at 120 Hz and GRF and EMG data were collected at 1200 Hz. EMG data were collected using surface electrodes from eight intact-leg muscles and five residual-leg muscles (Table 2). Kinematic data were low-pass filtered using a 4th-order Butterworth filter with a cutoff frequency of 6 Hz. GRF data were similarly filtered with a cutoff frequency of 20 Hz. EMG data were demeaned, rectified, high-pass filtered with a cutoff frequency of 40 Hz, and then low-pass filtered with a cutoff frequency of 4 Hz. Data were normalized to the gait cycle and averaged across subjects for both the amputee and non-amputee groups.

2.2. Musculoskeletal model

A three-dimensional musculoskeletal model was developed using SIMM (Musculographics, Inc.) and has been previously described in detail (Silverman and Neptune, 2012). The model had 14 rigid body segments and 23 degrees-of-freedom. Body segments included a right and left thigh, shank, patella, talus, calcaneus and toes, as well as a head-arms-trunk (HAT) segment and pelvis. Degrees-of-freedom included a six degree-of-freedom joint between the ground and pelvis, three rotational degrees-of-freedom between the pelvis and HAT, and three rotational degrees-of-freedom between the pelvis and each thigh. Single rotational degrees-of-freedom were defined at the knee, ankle, subtalar and metatarsalphalangeal joints. Passive torques were applied at each joint to represent ligament and passive tissue forces (Anderson, 1999; Davy and Audu, 1987). Foot-ground contact was modeled using 31 independent visco-elastic elements with coulomb friction on the bottom of each foot (Neptune et al., 2000). The dynamic equations-of-motion were generated using SD/FAST (PTC).

The model included 38 Hill-type musculotendon actuators per leg, with geometry based on Delp et al. (1990) and force-length-velocity relationships (Zajac, 1989). The excitation for each muscle was defined using a bimodal excitation pattern (e.g., Silverman and Neptune, 2012). Muscle activation/deactivation dynamics were modeled using a first-order differential equation (Raasch et al., 1997) with time constants based on Winters and Stark (1988). For the non-amputee model, symmetric muscle excitation patterns were used between legs.

To model the unilateral transtibial amputees, the mass and inertial properties of the shank were modified and muscles crossing the ankle joint were removed to represent the residual leg in an amputee (Silverman and Neptune, 2012). The prosthesis was modeled using a second-order torsional spring with damping at the ankle joint (Silverman and Neptune, 2012).

2.3. Optimization framework

A simulated annealing optimization framework (Goffe et al., 1994) was used to generate three forward dynamics simulations of the stance phase when the joint contact forces are the highest (Silverman and Neptune, 2012). The simulations consisted of amputee residual leg stance, amputee intact leg stance and non-amputee left leg stance. To generate simulations emulating the human subject walking mechanics, the optimization cost function (Eq. (1)) consisted of the

Table 2
Muscles included in the musculoskeletal model and corresponding muscle groups. '*' indicates muscles that were not included in the residual leg of the amputee model and 'Δ' indicates that electromyographic data were available for the muscle.

Muscle	Muscle group
Iliacus	IL
Psoas	
Adductor longus	AL
Adductor brevis	
Pectineus	
Quadratus femoris	
Adductor magnus (superior, middle, and inferior compartments)	AM
Sartorius	SAR
$^{\Delta}$ Rectus femoris	RF
Vastus medialis	VAS
Vastus intermedius	
$^{\Delta}$ Vastus lateralis	
$^{\Delta}$ Gluteus medius (anterior, middle and posterior compartments)	GMED
Gluteus minimus (anterior, middle and posterior compartments)	
Gemellus	
Piriformis	
Tensor fascia lata	TFL
$^{\Delta}$ Gluteus maximus (superior, middle and inferior compartments)	GMAX
Semimembranosus	HAM
Semitendinosus	
Gracilis	
$^{\Delta}$ Biceps femoris long head	
Biceps femoris short head	BFSH
* $^{\Delta}$ Gastrocnemius (medial and lateral heads)	GAS
* $^{\Delta}$ Soleus	SOL
*Tibialis posterior	
*Flexor digitorum longus	
* $^{\Delta}$ Tibialis anterior	TA
*Extensor digitorum longus	

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

	Age (years)	Body mass (kg)	Height (m)	Time since amputation (years)	Etiology	Prosthetic foot type
Amputees	45.1 (9.1)	90.6 (18.6)	1.76 (0.1)	5.6 (2.9)	11 Traumatic 3 Vascular	9 ESAR 5 SACH
Non-amputees	34.1 (13.0)	70.9 (13.6)	1.76 (0.1)	–	–	–