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Altering prosthetic foot stiffness influences foot and muscle function during below-knee amputee walking: A modeling and simulation analysis

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ABSTRACT

Most prosthetic feet are designed to improve amputee gait by storing and releasing elastic energy during stance. However, how prosthetic foot stiffness influences muscle and foot function is unclear. Identifying these relationships would provide quantitative rationale for prosthetic foot prescription that may lead to improved amputee gait. The purpose of this study was to identify the influence of altered prosthetic foot stiffness on muscle and foot function using forward dynamics simulations of amputee walking. Three 2D muscle-actuated forward dynamics simulations of unilateral below-knee amputee walking with a range of foot stiffness levels were generated, and muscle and prosthetic foot contributions to body support and propulsion and residual leg swing were quantified. As stiffness *decreased*, the prosthetic keel provided increased support and braking (negative propulsion) during the first half of stance while the heel contribution to support decreased. During the second half of stance, the keel provided decreased propulsion and increased support. In addition, the keel absorbed less power from the leg, contributing more to swing initiation. Thus, several muscle compensations were necessary. During the first half of stance, the residual leg hamstrings provided decreased support and increased propulsion. During the second half of stance, the *intact leg* vasti provided increased support and the residual leg rectus femoris transferred increased energy from the leg to the trunk for propulsion. These results highlight the influence prosthetic foot stiffness has on muscle and foot function throughout the gait cycle and may aid in prescribing feet of appropriate stiffness.

1. Introduction

Below-knee amputees commonly develop asymmetrical gait patterns and comorbidities in their residual and intact legs (Burke et al., 1978; Sanderson and Martin, 1997; Winter and Sienko, 1988). Prosthetic feet have been developed to minimize these asymmetries by utilizing elastic energy storage and return (ESAR) to help provide important walking subtasks including body support, forward propulsion and leg swing initiation, which are normally provided by the ankle plantar flexors in non-amputee walking (e.g., McGowan et al., 2009; Neptune et al., 2001). However, the influence of ESAR prosthetic foot stiffness on walking mechanics is not well-understood. One challenge to acquiring needed biomechanical data to identify this influence is the complexity of manufacturing custom feet with specific stiffness levels.

To overcome this challenge, we recently developed a manufacturing framework integrating selective laser sintering (SLS) to systematically vary a prosthetic foot design to assess the influence of foot stiffness on amputee kinematics, kinetics and muscle activity during walking (South et al., 2010; Fey et al., 2011). We found that decreasing foot stiffness increases the prosthesis range of motion, mid-stance energy storage and late-stance energy return that results in reduced residual leg hamstring activity. Thus, decreasing stiffness may enable ESAR prosthetic feet to provide additional forward propulsion and reduce the compensatory action of the hamstring muscles (Neptune et al., 2004). However, as stiffness decreased, a reduced residual leg vertical ground reaction force (GRF) and increased muscle activity of the residual leg vastus and gluteus medius, and intact leg vastus and rectus femoris were observed. These changes appear to be necessary to provide needed body support (Anderson and Pandy, 2003; Liu et al., 2006; Neptune et al., 2004). Thus, reduced residual leg hamstring contributions to forward propulsion during late-stance may be offset by needed muscle compensations to provide body support. Identifying the causal relationships between

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prosthetic foot stiffness and muscle and prosthetic foot function would provide quantitative rationale for prosthetic foot prescription that might be otherwise difficult to discern using experimental techniques.

The purpose of this study was to identify the influence of prosthetic foot stiffness on muscle and foot function by developing forward dynamics simulations of below-knee amputee walking with a range of foot stiffness levels. Previously, forward dynamics simulations have provided insight into muscle contributions to body support, forward propulsion and leg swing walking subtasks (e.g., Anderson and Pandy, 2003; Liu et al., 2006; McGowan et al., 2009; Neptune et al., 2004). Based on our previously-observed experimental findings, we tested the hypotheses that as stiffness decreases, foot contributions to forward propulsion and leg swing initiation would increase, and therefore muscle contributions to these subtasks would decrease. Also as stiffness decreases, we expected that foot contributions to body support would decrease, and therefore muscle contributions to body support would increase.

2. Methods

2.1. Bipedal musculoskeletal model

Forward dynamics simulations of unilateral below-knee amputee walking were generated using a planar bipedal musculoskeletal model (Fig. 1) developed using SIMM/Dynamics Pipeline (MusculoGraphics, Inc.). The model used was similar to previous analyses of walking (McGowan et al., 2009; Neptune et al., 2001) and consisted of rigid trunk, thigh and shank segments. Segments also represented the talus, calcaneus, mid-foot and toes of the intact leg foot. Musculoskeletal geometry was based on Delp et al. (1990). Residual leg shank inertial properties were based on Mattes et al. (2000). The trunk had two translational and one rotational degrees-of-freedom, while revolute joints modeled flexion/extension of the hip joints. The knees were modeled using planar joints, with a flexion/extension rotation and the two translational degrees-of-freedom prescribed as a function of knee flexion angle. Revolute joints modeled flexion/extension of the ankle, subtalar and metatarsophalangeal joints of the intact leg. Therefore, not including the prosthetic foot, the model had 10 degrees-of-freedom. Passive torques representing joint structures and tissues were applied (Anderson and Pandy, 1999; Davy and Audu, 1987), and 31 visco-elastic elements with coulomb friction were attached beneath each foot to model foot-ground contact (Neptune et al., 2000). The system equations of motion were generated using SD/FAST (PTC).

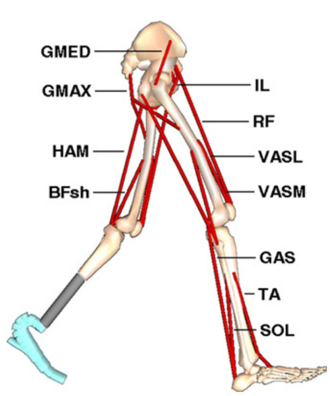

Fig. 1. The amputee musculoskeletal model was actuated by 25 Hill-type musculotendon actuators in the intact leg categorized into 14 muscle groups based on anatomical classification, with actuators in each group receiving the same excitation pattern. These muscle groups were: GMED (anterior and posterior compartments of the gluteus medius), GMAX (gluteus maximus, adductor magnus), HAM (biceps femoris long head, medial hamstrings), BFsh (biceps femoris short head), IL (psoas, iliacus), RF (rectus femoris), VASL (vastuslateralis, vastusintermedius), VASM (vastusmedialis), GAS (medial and lateral gastrocnemius), SOL (soleus, tibialis posterior), TA (tibialis anterior, peroneus tertius), PR (peroneus longus, peroneus brevis), FLXDG (flexor digitorumlongus, flexor hallucislongus) and EXTDG (extensor digitorumlongus, extensor hallucislongus). For figure clarity, the smaller muscle groups that control the foot (PR, FLXDG and EXTDG) are not shown. The residual leg had the same muscle groups except for the muscles that span the ankle (GAS, SOL, TA, PR, FLXDG and EXTDG) since these muscles are either removed or severely altered during a below-knee amputation.

2.2. Hill-type musculotendon actuators

Hill-type musculotendon actuators governed by intrinsic muscle force-length-velocity relationships (Zajac, 1989) were used to drive the model. Similar to previous work (Hall et al., 2011), excitation patterns were parameterized using a bimodal equation:

$$e(t)=\sum_{i=1}^{2}\begin{cases}\frac{a_i}{2}\left[1-\cos\left(\frac{2\pi(t-\text{onset}_i)}{\text{offset}_i-\text{onset}_i}\right)\right], & \text{onset}_i\le t\le \text{offset}_i\\ 0, & \text{otherwise}\end{cases}$$

where the excitation magnitude $e(t)$ depended on time (t) and amplitude (a_i), onset (onset$_i$), and offset (offset$_i$) of each mode (i). A first-order differential equation using time constants based on Winters and Stark (1988) was used to model the activation-deactivation dynamics (Raasch et al., 1997).

2.3. Energy storage and return prosthetic foot model

The prosthetic foot was modeled using 22 rigid segments, with the foot shape described by two spline curves (Fig. 2). The foot shape closely matched previously-manufactured SLS ESAR feet, which were based on a commonly prescribed commercial carbon fiber ESAR foot (Highlander, FS 3000, Freedom Innovations, LLC), and used in a human subject experiment with amputee participants (Fey et al., 2011; South et al., 2010).

At each rotational degree-of-freedom in the prosthetic foot model, a visco-elastic element applied a passive torque:

$$\tau_i=-k_i\theta_i-b\dot{\theta}_i$$

where each passive torque (τ_i) depended on element stiffness (k_i), angular displacement (θ_i) and angular velocity $\left(\dot{\theta}_i\right)$. A damping value b=5.73 N m s was used for each element (Fey et al., 2011). Three different prosthetic stiffness distributions (Fig. 3) were used that matched the stiffness levels from the previously manufactured nominal, compliant (50% less stiff) and stiff (50% more stiff) SLS ESAR prosthetic feet (Fey et al., 2011).

2.4. Dynamic optimization and experimental tracking data

Dynamic optimization was used to identify the muscle excitation parameters of each muscle group and the initial generalized velocities. A simulated annealing algorithm (Goffe et al., 1994) solved the optimal tracking problem to generate simulations of amputee walking with nominal, compliant and stiff feet. Group average experimental data of amputee subjects walking with each of the three SLS prosthetic feet were used as tracking data, while the objective functions also minimized the sum of squared individual muscle stresses. Subjects included 12 unilateral, below-knee amputees that walked overground at 1.2 m/s, while kinematic motion, GRF and surface electromyography (EMG) data were measured (see Fey et al., 2011).

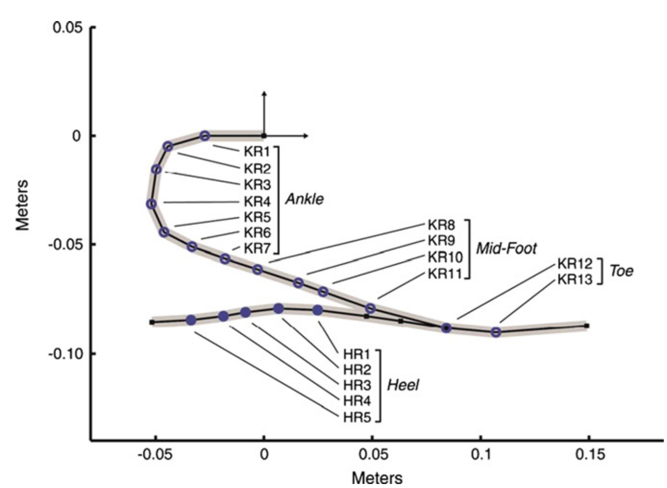

Fig. 2. The prosthetic foot model included 22 rigid segments connected in series by 13 keel (KR1-KR13, ○) and 5 heel (HR1-HR5, ●) revolute joints. Black squares indicate segment endpoints where rotational degrees-of-freedom are not present.