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The influence of energy storage and return foot stiffness on walking mechanics and muscle activity in below-knee amputees

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ABSTRACT

Background: Below-knee amputees commonly experience asymmetrical gait patterns and develop comorbidities in their intact and residual legs. Carbon fiber prosthetic feet have been developed to minimize these asymmetries by utilizing elastic energy storage and return to provide body support, forward propulsion and leg swing initiation. However, how prosthetic foot stiffness influences walking characteristics is not well-understood. The purpose of this study was to identify the influence of foot stiffness on kinematics, kinetics, muscle activity, prosthetic energy storage and return, and mechanical efficiency during amputee walking.
Methods: A comprehensive biomechanical analysis was performed on 12 unilateral below-knee amputees. Subjects walked overground at 1.2 m/s with three prosthetic feet of varying keel and heel stiffness levels, which were created using additive manufacturing.
Findings: As stiffness decreased, peak residual and intact leg ankle angles and residual leg knee flexion angle increased. The residual and intact leg braking ground reaction forces and knee extensor moments, residual leg vastus and gluteus medius activity, and intact leg vastus and rectus femoris activity also increased. The second vertical ground reaction force peak and hamstring activity in the residual leg and first vertical ground reaction force peak in the intact leg decreased. In addition, prosthetic energy storage and return increased and mechanical efficiency decreased as stiffness decreased.
Interpretation: Decreasing foot stiffness can increase prosthesis range of motion, mid-stance energy storage and late-stance energy return, but the net contributions to forward propulsion and swing initiation may be limited as additional muscle activity to provide body support becomes necessary.
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1. Introduction

The number of people living with limb loss in the U.S. is projected to dramatically increase over the next 40 years (Ziegler-Graham et al., 2008), with approximately 40% representing major lower limb amputations. Unilateral below-knee amputations often lead to temporal and ground reaction force bilateral asymmetries (Nolan et al., 2003; Sanderson and Martin, 1997; Silverman et al., 2008; Vrieling et al., 2008), altered residual leg muscle activity (e.g., Fey et al., 2010; Winter and Sienko, 1988), higher metabolic cost (Czerniecki, 1996; Genin et al., 2008; Waters et al., 1976) and reduced walking speed (Hermodsson et al., 1994; Perry et al., 1997; Powers et al., 1998; Robinson et al., 1977) relative to non-amputee walking. Many differences between amputee and non-amputee walking can be attributed to the absence of the ankle plantar flexor muscles, which provide needed body support, forward propulsion and swing initiation during non-amputee walking (Neptune et al., 2001). As a result, previous studies have identified a number of neuromotor adaptations used by amputees to provide these important biomechanical functions (e.g., Silverman et al., 2008; Winter and Sienko, 1988). To help minimize the necessary adaptations and improve amputee walking ability, proper prosthetic foot prescription and design is critical.

Some amputees use low or non-energy storage and return feet (e.g., solid ankle cushioned heel, SACH), which reduce ground contact forces during the initial loading response following heel-strike as well as provide body support. However, these prosthetic feet provide minimal energy storage and return over the stance phase (e.g., Ehara et al., 1993) due to their high stiffness and limited deflection, and therefore provide few biomechanical advantages (for review, see Hafner et al., 2002a). In an effort to improve performance, carbon fiber energy storage and return (ESAR) feet have been developed that store and release elastic energy during stance (Hafner et al., 2002a, 2002b) and provide body support, forward propulsion and leg swing initiation (Zmitrewicz et al., 2007). However, the appropriate ESAR stiffness level for a given amputee is not well-defined (Hafner et al., 2002b). Currently, clinicians base prosthetic prescription on a patient's body weight and activity level. The latter is evaluated

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subjectively using clinician experience as well as patient self-report and feedback, rather than objective biomechanical data. Thus, studies are needed to systematically vary foot stiffness to assess its influence on gait characteristics, which has the potential to improve amputee mobility by developing quantitative rationale for prosthetic foot prescription.

However, one challenge to such studies is the difficulty in generating ESAR feet of varying stiffness levels using traditional manufacturing techniques. One solution to overcome this difficulty is to use an additive manufacturing technology known as selective laser sintering (SLS), which has been used to fabricate ankle-foot orthoses (Faustini et al., 2008), below-knee prosthetic sockets (Faustini et al., 2005; Faustini et al., 2006a, 2006b; Rogers et al., 2007) and prosthetic ankles (Ventura et al., 2011). We recently developed a SLS-based framework to fabricate ESAR feet that replicated the geometry and dynamic response of a commercially available carbon fiber ESAR foot (South et al., 2010). The purpose of this study was to use this framework to identify the influence of ESAR foot stiffness on gait characteristics by manufacturing SLS ESAR feet with a range of stiffness levels and quantify their effect on below-knee amputee walking. Specifically, we investigated the influence of ESAR foot stiffness on gait kinematics, kinetics, muscle activity, prosthetic energy storage and return, and mechanical efficiency during overground walking.

2. Methods

Prosthetic foot stiffness was modified by altering keel and heel geometry (for details, see South et al., 2010) to yield three SLS feet: one that closely matched the nominal stiffness of a widely prescribed carbon fiber foot (Highlander^{TM}, FS 3000, Freedom Innovations, LLC), one that was 50% stiffer than this foot, and one that was 50% more compliant. The feet (Fig. 1) were mechanically tested using a uniaxial load cell (ATS, Inc.). Each foot was clamped to the base of the machine and loaded axially by a flat plate attached to the load cell in similar configurations as specified in the ISO 10328 standard for structural testing of lower-limb prostheses. The feet were tested in three different configurations: toe-only contact, heel-only contact and foot-flat (South et al., 2010). Foot mass increased as stiffness increased (compliant = 355 g, nominal = 426 g, stiff = 449 g). To assure all feet had identical mass, small amounts of lead were placed near the center-of-mass of the compliant and nominal feet. These feet were then used in the overground walking trials.

2.1. Subjects

Twelve unilateral below-knee amputees participated in the study (Table 1). All subjects were asymptomatic of musculoskeletal disorders and pain and were proficient walkers who did not use assistive devices.

Table 1
Below-knee amputee subject characteristics.

	Mean	Std. Dev.	Max	Min	#Subjects
Age (years)	51.9	17.1	72	25	–
Height (m)	1.78	0.06	1.91	1.66	–
Body mass (kg)	83.0	15.8	120.7	65.3	–
Post-amputation time (years)	13.9	14.4	42	0.8	–
Gender					
Male	–	–	–	–	12
Female	–	–	–	–	0
Cause of Amputation					
Traumatic injury	–	–	–	–	10
Secondary illness	–	–	–	–	2
Prescribed Prostheses					
Energy storage and return (ESAR)	–	–	–	–	12
Non-ESAR	–	–	–	–	0

2.2. Protocol

The experimental protocol was a crossover design in which amputees walked with the three different prosthetic feet in a randomized order. Prior to use, a certified and licensed prosthetist with 15 years clinical experience ensured proper fit and alignment. Subjects were allowed as much time as needed until they felt comfortable with each foot, which did not exceed five minutes for any subject. Subjects provided informed consent to an Institutional Review Board approved protocol prior to participation. Subjects walked overground along a 10-meter walkway consisting of four embedded force plates (Advanced Mechanical Technology, Inc. and Bertec Corporation) at 1.2 +/− 0.06 m/s, which was verified using infrared timing gates. Repeated trials were collected at each condition until at least five force plate contacts per leg were measured.

Ground reaction force (GRF) and kinematic marker data were measured at 1200 Hz and 120 Hz, respectively, using a motion capture system (Vicon Nexus, Oxford Metrics, Inc.). 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. Marker clusters (each consisting of four markers) were also placed bilaterally on the shank and thigh. Markers on the residual leg were placed symmetrically to the intact leg.

Electromyographic (EMG) data were collected at 2160 Hz using surface EMG electrodes (2, 10-mm disposable self-adhesive Ag/AgCl sensor contacts, 16-mm interelectrode distance; TeleMyo 900, Noraxon U.S.A., Inc.) from eight intact leg muscles including the tibialis anterior (TA), medial gastrocnemius (GAS), soleus (SOL), vastus medialis (VAS), rectus femoris (RF), biceps femoris long head (BF), gluteus medius (GMED) and gluteus maximus (GMAX), and from five residual leg muscles including VAS, RF, BF, GMED and GMAX. A ground electrode was placed on the sacrum. Common mode rejection and amplification characteristics included: effective 130 dB

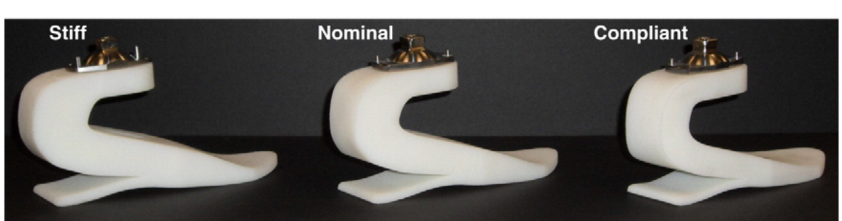

Fig. 1. ESAR prosthetic feet fabricated using SLS additive manufacturing technology that approximately matched the nominal carbon fiber stiffness (Nominal), and were 50% more stiff (Stiff) and 50% more compliant (Compliant). All feet were manufactured from Rilsan^{TM} D80 (Nylon 11 powder, Arkema) using a Vanguard HiQ Sinterstation (3D Systems, Inc.).

at DC (0 Hz), >100 dB at 50/60 Hz, and minimum 85 dB through 10–500 Hz operating range; <1.2 μV root-mean-square (RMS) noise level, 2000 fixed overall gain and >10 MΩ input impedance. Electrodes were placed on the muscle belly along the line of action between the origin and insertion points based on guidelines provided by Perotto and Delagi (1994). Each location was shaved and cleaned with alcohol prior to electrode placement.

2.3. Kinematics and kinetics

An inverse dynamics analysis was performed to calculate net intersegmental joint moments using Visual3D (C-Motion, Inc.). Kinematic data were low-pass filtered using a 4th-order Butterworth filter with a cut-off frequency of 6 Hz, while the GRF data were low-pass filtered with a cut-off frequency of 20 Hz. Inertial properties of the residual leg were adjusted in the inverse dynamics model based on previous work (Mattes et al., 2000). GRFs and joint moments were normalized by body weight and body mass, respectively. For the residual and intact legs, sagittal plane GRFs and joint angles and ankle, knee and hip moments were averaged across trials for each condition.

2.4. Electromyography

Raw EMG signals were filtered and processed in MATLAB (MathWorks, Inc.). Data were demeaned and then smoothed using a moving, symmetric 80-ms RMS window. Timing of the gait cycle events of each leg were exported using Visual3D. The smoothed EMG data were extracted in MATLAB based on the nearest time point for each event. The data from each gait cycle were then time-normalized to represent 100% of the gait cycle. The data for each gait cycle were then integrated within specific regions of the gait cycle (Perry, 1992) that corresponded to the loading response and mid-stance (0–30% gait cycle), terminal stance and pre-swing (30–60% gait cycle), swing (60–100% gait cycle) in addition to over the entire gait cycle (0–100% gait cycle). These integrated EMG (iEMG) quantities were averaged at each condition and then normalized for each subject by the 0–100% gait cycle iEMG magnitude from the nominal stiffness condition.

2.5. Prosthetic energy storage and return and mechanical efficiency

Residual leg ankle power was computed as the product of the residual leg ankle moment and angular velocity. Prosthetic foot energy storage and return characteristics were estimated by evaluating the time integrals of the residual leg ankle power. For each condition, the integrals of the residual leg negative ankle power (energy stored) and positive ankle power (energy returned) (J/kg) were computed during stance for each gait cycle and averaged across trials for each subject. Mechanical efficiency was calculated as the ratio of energy returned divided by the energy stored during stance.

2.6. Statistical analyses

All statistical analyses were performed using SPSS 16.0 GP (SPSS, Inc.). Differences in peak GRFs, sagittal-plane joint angles and moments, iEMG magnitudes, and prosthetic energy quantities were analyzed. Braking, propulsion and the first (V1) and second (V2) vertical peak GRFs were analyzed. Sagittal-plane joint angles including ankle plantarflexion, dorsiflexion, knee flexion and hip extension peaks were also compared. Joint moment comparisons included dorsiflexion, plantarflexion, knee extension in early (K1) and late (K3) stance, knee flexion (K2), hip extension (H1) and hip flexion (H2) peaks. The iEMG quantities over each phase of the gait cycle (0–30% gait cycle, 30–60% gait cycle, and swing) were also compared for each muscle. Finally, the prosthetic energy storage and return, and mechanical efficiency quantities were compared across stiffness levels. Differences in GRF, joint angle and moment, and bilateral iEMG quantities were compared using two-factor (prosthetic foot, leg) ANOVAs. Differences in unilateral iEMG and prosthetic energy

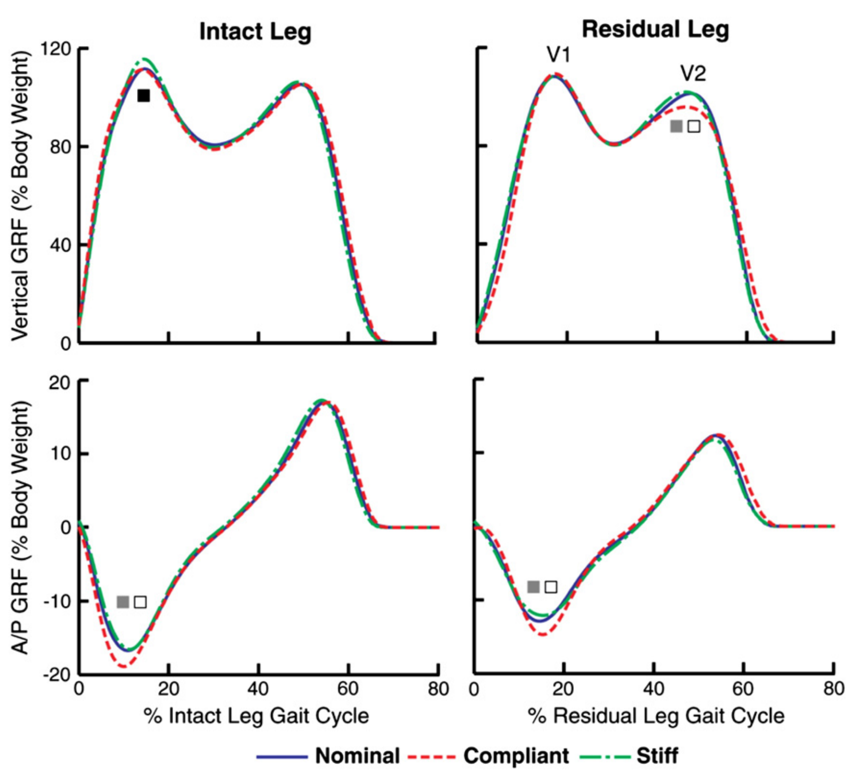

Fig. 2. Group average ground reaction force (GRF) data. GRF data of the residual and intact legs were normalized in time to their respective gait cycles. Indicated are vertical V1 and V2 GRFs. Significant differences associated with each peak quantity between nominal and compliant (■), compliant and stiff (□), and nominal and stiff (■) conditions are also indicated.

quantities were compared using one-factor (prosthetic foot) ANOVAs. The prosthetic foot factor consisted of three levels (nominal, compliant and stiff) and the leg factor consisted of two levels (residual and intact). If an ANOVA had a significant main or interaction effect, pairwise comparisons were evaluated using a Bonferroni adjustment for multiple comparisons, which reduced the statistical significance level from $\alpha = 0.05$. For presentation of these pairwise results, SPSS adjusted individual p-values were used for comparison to $\alpha = 0.05$. Leg main effects and pairwise differences between legs (i.e., residual leg compared to the intact leg) were not reported.

3. Results

3.1. Ground reaction forces

Altering prosthetic foot stiffness influenced the resulting peak GRF quantities in both residual and intact legs (Fig. 2, Table 2). Peak braking GRFs were larger in the compliant condition compared to the nominal and stiff conditions (prosthetic effect, $p<0.001$; compliant to nominal, $p \le 0.033$; compliant to stiff, $p \le 0.036$), while intact leg V1 GRF was larger in the stiff condition compared to the nominal condition (leg*prosthetic effect, $p=0.016$; stiff to nominal, $p=0.002$). In contrast, residual leg V2 GRF was reduced in the compliant condition relative to both the stiff and nominal conditions (Fig. 2, Table 2; prosthetic effect, $p<0.001$; leg*prosthetic effect, $p<0.001$; compliant to nominal, $p<0.001$; compliant to stiff, $p=0.001$).

Table 2
Mean (standard deviation) GRFs, joint angles, and joint moments of the residual and intact leg for each prosthetic foot condition.

		Compliant	Nominal	Stiff
GRFs (% Body Weight)				
V1 ○	Residual	112.0 (3.6)	110.3 (4.8)	109.7 (5.8)
	Intact ■	113.0 (5.6)	**113.1 (8.2)**	**117.4 (8.3)**
V2 ▲○	Residual ▩□	**96.5 (4.3)**	**102.1 (5.0)**	**103.4 (4.6)**
	Intact	106.2 (5.3)	105.9 (5.8)	107.0 (5.1)
Braking ▲	Residual ▩□	**−15.8 (3.1)**	**−13.9 (3.2)**	**−13.4 (2.5)**
	Intact ▩□	**−19.3 (3.1)**	**−17.3 (2.6)**	**−17.3 (2.5)**
Propulsion	Residual	12.6 (1.8)	12.7 (1.6)	12.4 (1.8)
	Intact	17.2 (2.2)	17.0 (2.2)	17.4 (2.0)
Joint Angles (°)				
Plantarflexion ▲○	Residual ▩□	**−14.4 (3.5)**	**−12.2 (3.0)**	**−11.6 (3.0)**
	Intact	−12.4 (2.9)	−12.3 (2.1)	−12.3 (2.4)
Dorsiflexion ▲○	Residual ▩□■	**13.1 (2.9)**	**8.6 (2.5)**	**5.7 (2.4)**
	Intact □	**11.0 (2.4)**	10.2 (2.3)	**9.2 (3.0)**
Knee flexion ▲	Residual □	**−58.8 (6.2)**	−57.2 (6.4)	**−56.7 (5.8)**
	Intact	−60.4 (5.4)	−60.3 (5.9)	−59.5 (6.2)
Hip extension	Residual	−13.2 (4.3)	−13.5 (4.0)	−13.1 (3.4)
	Intact	−11.0 (3.8)	−10.8 (3.8)	−10.3 (3.4)
Joint Moments (Nm/kg)				
Dorsiflexion ▲○	Residual ■	0.35 (0.14)	**0.36 (0.12)**	**0.42 (0.10)**
	Intact	0.29 (0.12)	0.27 (0.11)	0.27 (0.12)
Plantarflexion ▲○	Residual ▩□	**−0.85 (0.15)**	**−1.05 (0.17)**	**−1.04 (0.13)**
	Intact	−1.33 (0.13)	−1.32 (0.13)	−1.34 (0.13)
K1 ▲○	Residual □■	**0.24 (0.25)**	**0.27 (0.24)**	**0.33 (0.24)**
	Intact ▩	**0.56 (0.19)**	**0.44 (0.17)**	0.48 (0.17)
K2 ▲○	Residual ▩□■	**0.05 (0.15)**	**−0.10 (0.15)**	**−0.14 (0.16)**
	Intact □	**−0.16 (0.17)**	−0.20 (0.15)	**−0.21 (0.14)**
K3 ▲○	Residual ▩□	**0.34 (0.12)**	**0.28 (0.10)**	**0.27 (0.11)**
	Intact	0.30 (0.11)	0.28 (0.09)	0.29 (0.09)
H1 ▲	Residual	−0.68 (0.21)	−0.67 (0.18)	−0.65 (0.18)
	Intact □	**−0.60 (0.20)**	−0.58 (0.19)	**−0.54 (0.19)**
H2	Residual	0.78 (0.22)	0.83 (0.24)	0.82 (0.25)
	Intact	0.92 (0.15)	0.94 (0.14)	0.95 (0.15)

▲ denotes a significant prosthetic main effect.
○ denotes a significant leg*prosthetic interaction effect.
▩ denotes a significant nominal to compliant pairwise comparison.
□ denotes a significant compliant to stiff pairwise comparison.
■ denotes a significant nominal to stiff pairwise comparison.

3.2. Joint kinematics and kinetics

The most notable change in joint kinematics was residual leg dorsiflexion angle (Table 2), which systematically increased as stiffness decreased. Relative to nominal, residual leg dorsiflexion angle increased and decreased by 52% and 34% in the compliant and stiff conditions, respectively (prosthetic effect, $p<0.001$; leg*prosthetic effect, $p<0.001$; compliant to nominal, $p<0.001$; nominal to stiff, $p<0.001$; compliant to stiff, $p<0.001$).

Intact leg dorsiflexion angle was larger in the compliant condition relative to the stiff condition (Table 2; prosthetic effect, $p<0.001$; leg*prosthetic effect, $p<0.001$; compliant to stiff, $p=0.006$). Residual leg plantarflexion angle in the compliant condition was also larger compared to the nominal and stiff conditions (Table 2; prosthetic effect, $p=0.004$; compliant to nominal, $p=0.002$; compliant to stiff, $p=0.001$). During swing, residual knee flexion angle was larger in the compliant condition compared to the stiff condition (Table 2; prosthetic effect, $p=0.006$; compliant to stiff, $p=0.023$).

Residual leg ankle plantarflexion moment and knee moments in the residual (K2, K3) and intact (K1) legs showed the largest differences across stiffnesses (Fig. 3, Table 2). Residual leg plantarflexion moment in the compliant condition was reduced by 19% compared to the nominal and stiff conditions (Fig. 3, Table 2; prosthetic effect, $p<0.001$; leg*prosthetic effect, $p<0.001$; compliant to nominal, $p<0.001$; compliant to stiff, $p<0.001$). Compared to the nominal condition, residual leg K2 moment increased (became more flexor) in the stiff condition and decreased (became extensor) in the compliant condition, which represented 43% and 147% net changes, respectively (Fig. 3, Table 2; prosthetic effect, $p<0.001$; leg*prosthetic effect, $p<0.001$; compliant to nominal, $p<0.001$; nominal to stiff, $p=0.014$; compliant to stiff, $p<0.001$). In addition, residual leg K3 moment was 25% larger in the compliant condition compared to nominal and stiff conditions (prosthetic effect, $p<0.001$; leg*prosthetic effect; $p<0.001$; compliant to nominal, $p=0.002$; compliant to stiff, $p<0.001$). The increased residual leg K3 extensor moment in the compliant condition corresponded with a 29% larger intact leg K1 moment in the compliant condition compared to nominal (Fig. 3, Table 2; prosthetic effect, $p=0.008$; leg*prosthetic effect, $p<0.001$; compliant to nominal, $p=0.001$).

Intact leg K2 moment was reduced (prosthetic effect, $p<0.001$; leg*prosthetic effect, $p<0.001$; compliant to stiff, $p=0.048$) and H1 moment was larger (prosthetic effect, $p=0.035$; compliant to stiff, $p=0.010$) in the compliant condition compared to stiff (Fig. 3, Table 2). Finally, residual leg K1 moment was larger in the stiff condition compared to nominal and compliant conditions (Fig. 3, Table 2; prosthetic effect, $p=0.008$; leg*prosthetic effect, $p<0.001$; stiff to nominal, $p=0.011$; stiff to compliant, $p=0.012$) and residual leg dorsiflexion moment was larger in the stiff condition compared to nominal (prosthetic effect, $p=0.035$; leg*prosthetic, $p=0.006$; stiff compared to nominal, $p=0.016$).

3.3. Electromyography

No differences in muscle activity were found during the swing phase across all conditions (Fig. 4). However, differences in iEMG magnitudes were observed during stance in both residual and intact leg muscles, which generally increased activity as stiffness decreased.

From 0 to 30% of the gait cycle, the most apparent changes in muscle activity were observed in the residual leg GMED (prosthetic effect, $p=0.013$; leg*prosthetic effect, $p=0.006$; compliant to stiff, $p=0.003$) and intact leg VAS (prosthetic effect, $p=0.001$; compliant to nominal, $p=0.015$) and RF (prosthetic effect, $p=0.048$; compliant to nominal, $p=0.026$), which all increased in activity in the compliant condition as stiffness decreased (Fig. 4). Other differences were seen in GAS (prosthetic effect, $p=0.002$; compliant to nominal, $p=0.005$, compliant to stiff, $p=0.030$) and TA (prosthetic effect, $p=0.012$;

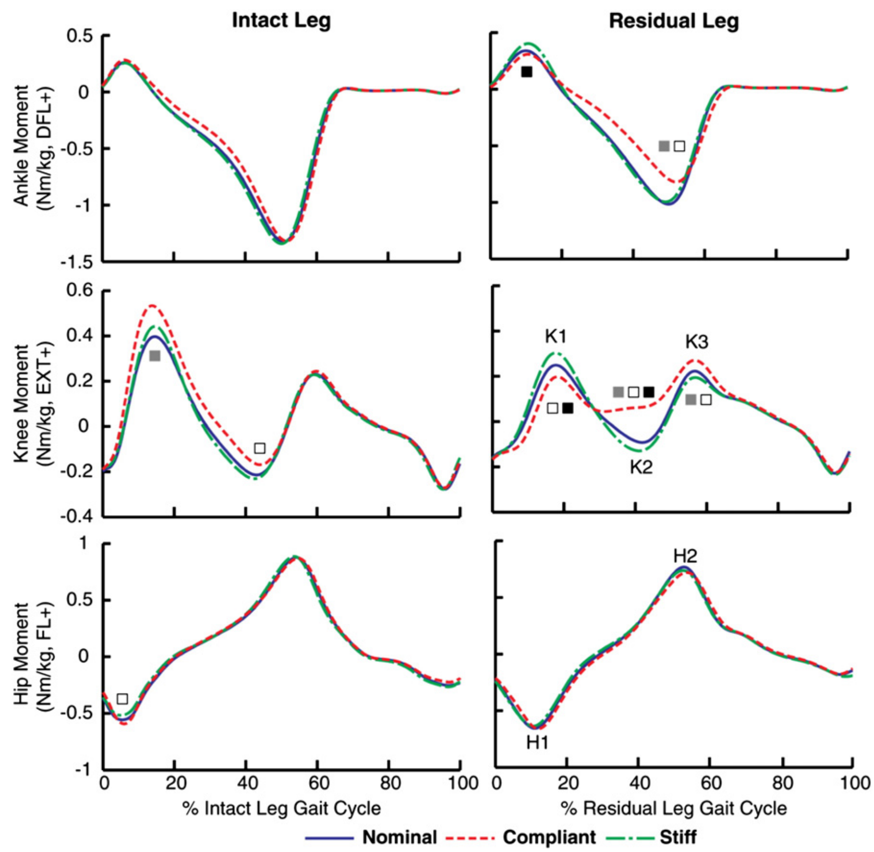

Fig. 3. Group average joint moment data of the residual and intact legs. Data for each leg (residual and intact) were normalized in time to their respective gait cycles. Indicated are knee K1, K2, and K3 moments and hip H1 and H2 moments. Significant differences associated with each peak quantity between nominal and compliant (■), compliant and stiff (□), and nominal and stiff (■) conditions are also indicated.

compliant to nominal, $p=0.015$), which had lower and higher muscle activities in the compliant condition, respectively (Fig. 4).

From 30 to 60% of the gait cycle, changes in muscle activity were observed in the residual leg GMED (prosthetic effect, $p<0.001$; leg*prosthetic effect, $p=0.031$; compliant to nominal, $p=0.037$; compliant to stiff, $p=0.007$) and VAS (prosthetic effect, $p=0.028$; compliant to nominal, $p=0.030$), which increased activity as stiffness decreased (Fig. 4). BF excitation also decreased as stiffness decreased (leg*prosthetic effect, $p<0.001$; compliant to stiff, $p=0.016$). Also, from 30 to 60% of the gait cycle, GMAX activity had a prosthetic main effect ($p=0.006$) and appeared to increase as stiffness decreased. Pairwise differences of residual leg GMAX activity during this region approached significance (compliant to nominal, $p=0.070$). In addition, increased residual leg RF activity approached significance as stiffness decreased (Fig. 4; prosthetic effect, $p=0.054$), although it was highly variable across subjects.

3.4. Prosthetic energy storage and return and mechanical efficiency

Prosthetic energy storage and return, and mechanical efficiency values followed consistent trends as stiffness was altered (Fig. 5). Energy stored showed a significant prosthetic effect ($p<0.001$), with energy values increasing as stiffness decreased (nominal compared to stiff, $p<0.001$; compliant to stiff, $p<0.001$). Energy returned also increased as stiffness decreased (prosthetic effect, $p=0.003$; nominal to stiff, $p=0.001$; compliant to stiff, $p=0.014$). Energy stored values were all larger than their respective energy returned values such that efficiency ranged between 49 and 59%. In contrast to energy stored and returned, mechanical efficiency decreased as stiffness decreased (Fig. 5).

4. Discussion

As a result of the changes in foot stiffness, significant differences in gait mechanics and muscle activity were observed in both the residual and intact legs, which primarily occurred in the muscle contributions to body support and forward propulsion. In terms of body support as stiffness decreased, the residual leg plantarflexion moment and V2 GRF were reduced (Figs. 2 and 3), and changes in joint kinematics were most apparent during single-leg support and terminal double-leg support of the residual leg stance and consistent with a more flexed body posture (Table 2). At the knee, the residual leg K2 and K3 moments became more extensor (Fig. 3, Table 2), which corresponded with increased VAS and reduced BF activity during the second half of stance (Fig. 4). At the hip, residual leg GMED activity increased throughout stance (Fig. 4). The changes in residual leg GMED and VAS were consistent with their functional roles to provide body support (Anderson and Pandy, 2003; Liu et al., 2006; McGowan et al., 2009; Neptune et al., 2004). Thus, as less support was provided by the prosthetic feet as stiffness decreased, these muscles appeared to compensate by increasing their activity to provide additional body support. Previously, increased residual VAS activity during the first half of stance had been identified in below-knee amputee walking using clinically prescribed prostheses (e.g., Fey et al., 2010; Powers et al., 1998). The present results suggest that decreased stiffness prolongs this compensatory mechanism into the second half of residual leg stance.

The changes in joint moments and muscle activity to provide additional support as stiffness decreased were not limited to the residual leg. During late stance of the residual leg (early stance of the intact leg), intact leg K1 extensor moment increased (Fig. 3, Table 2), consistent with its role to provide support during this region in non-

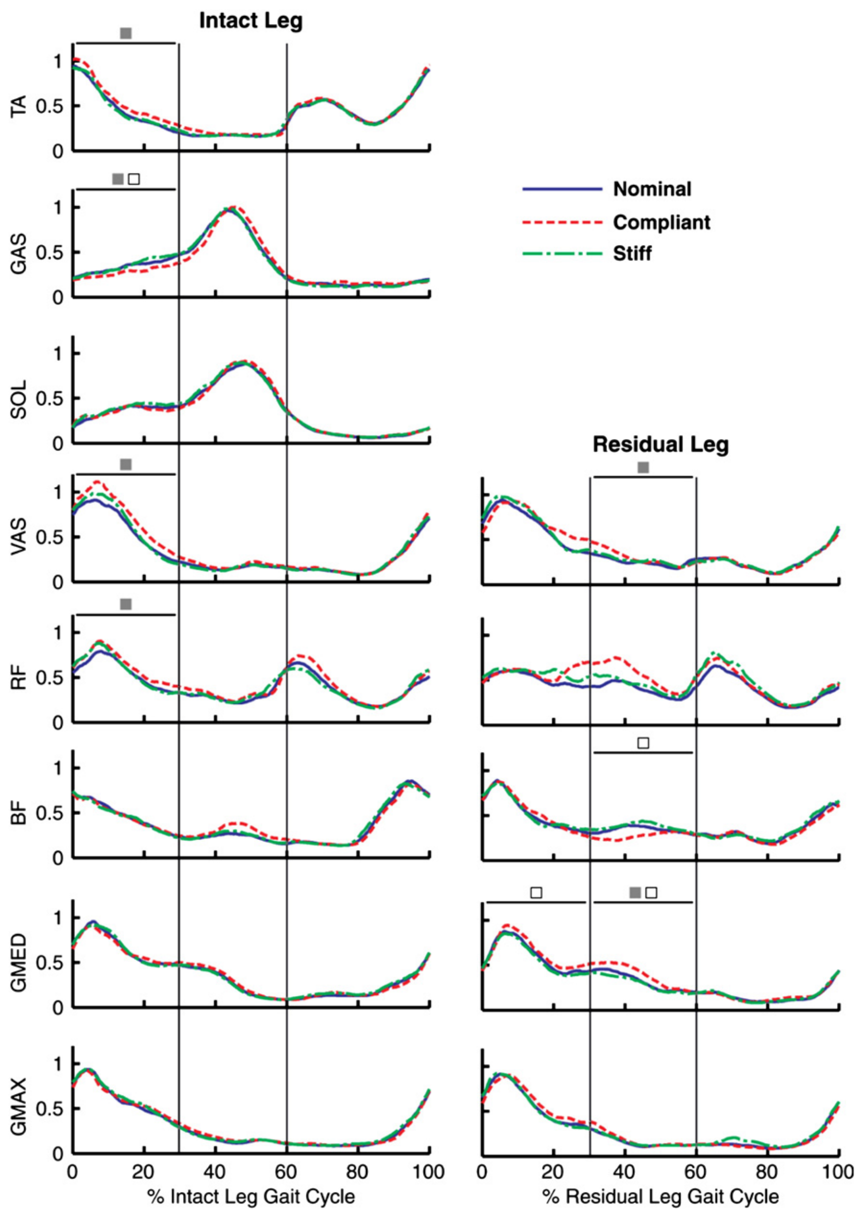

Fig. 4. Group average smoothed EMG data of the intact and residual legs. Regions of the gait cycle representing loading response and mid-stance (0–30% gait cycle), terminal stance and pre-swing (30–60% gait cycle), and swing (60–100% gait cycle) are shown by vertical lines. In each region, significant differences associated with integrated EMG magnitude values between nominal and compliant (■), compliant and stiff (□), and nominal and stiff (■) conditions are indicated.

amputee walking (Kepple et al., 1997). The increase in intact leg knee extensor moment appears to be a result of knee extensor muscles (VAS and RF) increasing their activity (Fig. 4), which is consistent with their roles to provide support in non-amputee walking (Anderson and Pandy, 2003; Liu et al., 2006; McGowan et al., 2009; Neptune et al., 2004). In addition, during early stance of the intact leg, the increased intact leg hip extensor moment is consistent with the role of hip extensor muscles (Anderson and Pandy, 2003; Kepple et al., 1997; Liu et al., 2006; Neptune et al., 2004) to provide body support in non-amputee walking.

Changes in the residual and intact leg VAS and intact leg RF muscles as stiffness decreased (Fig. 4) may have contributed to the increased braking GRFs observed in the compliant condition in both legs (Fig. 2, Table 2). VAS and RF have been shown in non-amputee walking to contribute to braking during the first half of stance (Liu et al., 2006; Neptune et al., 2004). During amputee walking using their clinically prescribed prostheses, reduced residual leg braking has been shown to increase net propulsion in the absence of the ankle muscles (Silverman et al., 2008). Therefore, an overly compliant prosthetic foot may limit the ability of amputees to make use of this mechanism due to an increased need for body support.

The energy stored in the ESAR feet is primarily returned in late stance of the residual leg (Hafner et al., 2002b). As foot stiffness decreased, the energy returned during this region increased (Fig. 5). In non-amputee walking, primary roles of the ankle plantar flexor muscles during late stance include providing forward propulsion and swing initiation by delivering energy to the trunk and leg, respectively (Neptune et al., 2001). Since the energy returned increased as stiffness decreased, the contributions of the prosthetic feet to forward propulsion and swing initiation may have increased. Since BF has been shown to contribute to propulsion throughout stance in non-amputee walking (Neptune et al., 2004), the reduced hamstring activity in the compliant condition (Fig. 4) suggests that the compliant prosthetic foot may be providing more propulsion during the second half of stance. However, the increased energy return in late stance occurred as a consequence of substantially greater energy storage

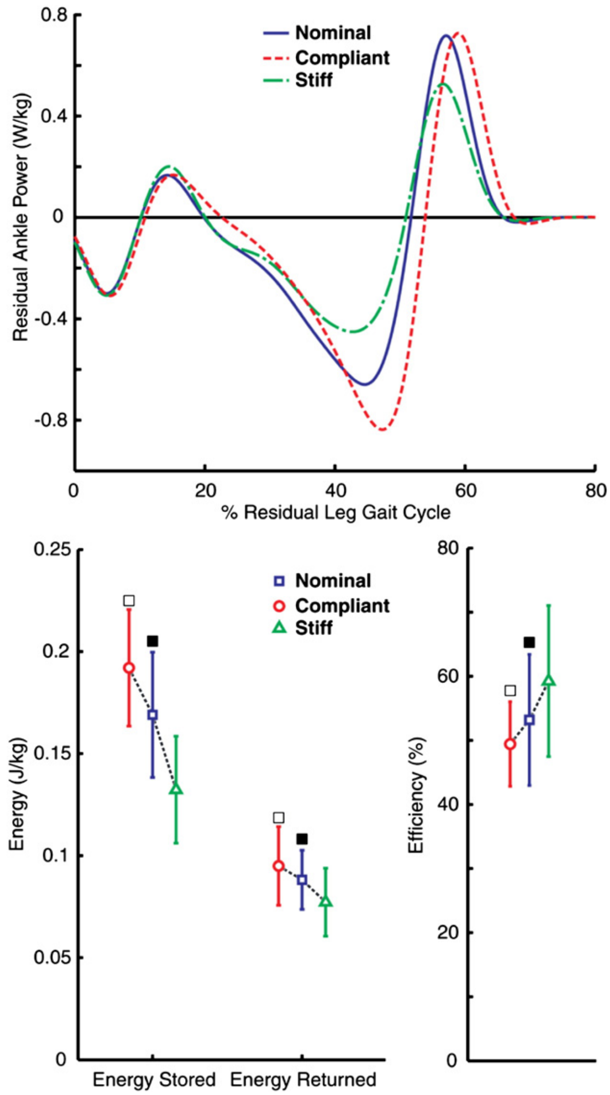

Fig. 5. Group average residual leg ankle power curves and group average (+/− standard deviation) prosthetic energy stored, energy returned and mechanical efficiency quantities. Significant differences associated with each quantity between nominal and compliant (■), compliant and stiff (□), and nominal and stiff (■) conditions are indicated.

(negative work) from mid to late stance and decreased mechanical efficiency (Fig. 5). These trends were consistent with previous studies of amputee walking and running that compared energy characteristics of high (Flex foot) and low (Seattle foot) ESAR feet to SACH prosthetic feet (Czerniecki et al., 1991; Gitter et al., 1991). These studies found that in order for the Flex and Seattle feet to return more energy in late stance compared to SACH, they also required greater than normal energy storage (negative work) in mid stance.

Like the plantar flexor muscles, the hip flexors have also been shown to be primary contributors to swing initiation in non-amputee walking during late stance (e.g., Neptune et al., 2001; Neptune et al., 2004). However, the residual hip flexion moment did not change across stiffness conditions. Therefore, the residual hip flexor contributions to swing initiation appear consistent across conditions. Despite a reduced plantarflexion moment in the compliant condition, prosthetic energy return increased as stiffness decreased. As previously noted, the contributions of the prosthetic feet to swing initiation may have increased as stiffness decreased. However, the total contribution of the hip flexors and prosthetic foot to swing initiation appear to have been insufficient since a kinematic adaptation to increase residual knee flexion angle during swing was observed (Table 2). This adaptation may have been needed to achieve toe clearance since a more flexed posture was observed as stiffness decreased from mid through late stance and at toe-off of the residual leg.

ESAR mechanical efficiency decreased as stiffness decreased (Fig. 5). However, a higher intact leg V1 GRF was observed in the stiff condition (Fig. 2, Table 2). As stiffness decreased, the decrease in intact leg V1 GRF and increase in residual leg dorsiflexion angle were consistent with a previous study that modified prosthetic foot rollover shape and found that residual leg dorsiflexion angle influences intact leg vertical GRF magnitude in the same manner (Hansen et al., 2006). Our results suggest an important tradeoff exists between ESAR prosthetic foot mechanical efficiency and intact leg loading.

Others have mechanically tested different commercially available prosthetic feet of varying designs and stiffness levels and shown that in general energy lost through loading and unloading increased as stiffness decreased (Geil, 2001). Although these feet were not clinically tested on amputees, their results were consistent with our study where we found that mechanical efficiency decreased as stiffness decreased. Studies that have clinically tested carbon fiber ESAR prosthetic feet on amputees have reported similar magnitudes of mechanical efficiency (~57%) as those calculated in this study (e.g., Barr et al., 1992). Thus, the ESAR properties of the SLS feet manufactured for this study functioned similarly.

Although the results of this study highlight the influence of prosthetic foot stiffness on gait characteristics of amputees, there are some potential limitations. First, rigid body and fixed axis of rotation assumptions were made when defining the prosthetic foot segment and residual leg ankle joint for inverse dynamics calculations. These assumptions may have influenced energy storage and return calculations. However, our goal was to observe relative differences in these estimated quantities as stiffness was altered. We expect that if a bias exists in these energy quantities across conditions, the same relative differences would be observed. Future work should consider including a deformable ESAR prosthetic foot model to account for its deflection and a possible moving axis of rotation. Another potential limitation is the same prosthetic feet were used on subjects of varying weight, which in some cases may have been substantially different than their prescribed prosthetic foot. However, we expect the same relative trends would be observed for similar percent changes in foot stiffness. Lastly, only male below-knee amputee patients were tested. Altering foot stiffness may result in different changes in walking mechanics and muscle activity for other classifications of amputees such as female or above-knee amputees.

In summary, changes in foot stiffness influenced the peak joint angles of both the residual and intact leg ankles during stance and residual leg knee flexion angle during swing. These quantities generally increased as stiffness decreased. Decreasing foot stiffness also resulted in increased braking GRFs in both the residual and intact legs and a decreased V2 GRF in the residual leg. During the same regions of residual leg stance, knee moments became more extensor in the residual (K2 and K3) and intact (K1) legs as stiffness decreased. These fluctuations in knee moments were consistent with the increased intact leg VAS and RF activity during the first half of stance; and in the residual leg, were consistent with the increased VAS activity and decreased BF activity during the second half of stance. The most apparent change in muscle activity was observed by residual leg GMED, which increased its activity throughout all of stance as stiffness decreased, consistent with its functional role to provide body support (Anderson and Pandy, 2003). All of these observed changes suggest a need for muscles to provide increased body support as prosthetic foot stiffness decreases. The results also suggest that the increased energy returned from the prosthetic feet during the second half of stance as stiffness decreased may have resulted in lower residual BF activity, which has previously been shown to contribute to body propulsion. Therefore, a tradeoff may exist between providing greater body support as stiffness increases and providing additional forward propulsion as stiffness decreases. In addition, as stiffness increased, the intact leg V1 GRF increased, which may be detrimental in amputee gait as below-knee amputees have a greater prevalence of knee joint disorders in the intact leg (Burke et al., 1978; Lemaire and Fisher, 1994).

These relationships between prosthetic foot stiffness and resulting gait mechanics and muscle activity should be considered when assessing the mobility needs of amputees and suggest that important tradeoffs exist between identifying appropriate stiffness levels to provide needed body support and forward propulsion and its potential influence on ESAR mechanical efficiency and intact leg loading.

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