Clinical Biomechanics 26 (2011) 298–303

Contents lists available at ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

Research Paper

The effects of prosthetic ankle dorsiflexion and energy return on below-knee amputee leg loading

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ARTICLE INFO

Article history:
Received 11 August 2010
Accepted 20 October 2010

Keywords:
Biomechanics
Transtibial amputee
Prosthesis
Gait
Walking
Joint kinetics

ABSTRACT

Background: Prosthetic devices are intended to return lower limb amputees to their pre-amputation functional status. However, prosthetic devices designed for unilateral below-knee amputees have yet to completely restore the biomechanical functions normally provided by the ankle muscles, leading to gait asymmetries and increased reliance on their intact leg. In an effort to improve amputee gait, energy storage and return feet have been developed that store mechanical energy in elastic structures in early to mid-stance and return it in late stance. However, little is known regarding how ankle compliance and the level of energy return influences walking mechanics. The purpose of this study was to identify the influence of prosthetic ankle dorsiflexion and energy storage and return on leg loading during steady-state walking.

Methods: Compliant ankles with different stiffness levels were attached to a Seattle Lightfoot2 in different orientations (forward- and reverse-facing).

Findings: The ankles decreased residual leg vertical ground reaction forces in late stance, increased residual leg propulsive ground reaction force impulses and increased residual leg knee joint extensor moments. The reverse-facing ankles increased residual leg vertical ground reaction forces in early stance, and the compliant forward-facing ankle increased residual leg braking impulses. In contrast to previous studies, increased energy storage and return from compliant ankles did not decrease hip joint powers or the intact leg vertical ground reaction forces.

Interpretation: These results provide insight into the relationships between ankle dorsiflexion, energy storage and return, and leg loading, which may lead to more effective prosthetic devices to improve amputee gait.

1. Introduction

Prosthetic restoration is intended to return lower limb amputees to their pre-amputation functional status. However, prosthetic feet and ankles designed for unilateral below-knee amputees have yet to completely restore the body forward propulsion, body support and leg swing initiation normally provided by the ankle muscles during walking (Liu et al., 2006; Neptune et al., 2001). This deficiency causes amputees to rely more on their intact leg than their residual leg, which results in greater intact leg stance times (Isakov et al., 2000), ground reaction forces (GRFs) (Nolan et al., 2003) and joint moments and powers (Bateni and Olney, 2002; Beyaert et al., 2008; Nolan and Lees, 2000; Nolan et al., 2003; Silverman et al., 2008) during walking. This asymmetric loading of the intact and residual legs increases the prevalence of secondary disabilities such as osteoarthritis in the intact leg, osteoporosis in the residual leg and lower back pain in amputees (Burke et al., 1978; Engsberg et al., 1991; Gailey et al., 2008; Kulkarni et al., 2005).

In an effort to improve amputee gait, energy storage and return (ESAR) feet have been developed to provide enhanced function by storing mechanical energy in elastic structures in early to mid-stance and returning it in late stance. ESAR feet are generally preferred by amputees over low energy return feet, but their effect on amputee gait has varied across studies (for review, see Hafner et al., 2002). Improvements in walking with ESAR feet have included increased self-selected walking speeds (Snyder et al., 1995), increased residual leg stride lengths and propulsive forces (Perry and Shanfield, 1993; Powers et al., 1994) and decreased peak vertical GRFs on the intact leg (Lehmann et al., 1993; Perry and Shanfield, 1993; Powers et al., 1994; Snyder et al., 1995). Most studies attribute these observed differences in gait mechanics to improved rocker motion of the prosthetic foot and lowered body center-of-mass (COM) resulting from increased dorsiflexion at the ankle joint (Powers et al., 1994; Snyder et al., 1995) or increased energy return at push-off from the ESAR feet (Czerniecki et al., 1991; Lehmann et al., 1993). However, no study has systematically altered ankle dorsiflexion and the level of energy storage and return from prosthetic feet to study their effects on amputee leg loading.

ESAR prosthetic ankles offer an effective way to systematically alter both ankle dorsiflexion and energy storage and return without changing

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0268-0033/$ – see front matter © 2010 Elsevier Ltd. All rights reserved.
doi:10.1016/j.clinbiomech.2010.10.003

the properties of the prosthetic heel and keel sections, which is a potential confounding factor in prosthetic foot studies. Previous studies analyzing prosthetic ankles have shown they can significantly influence amputee gait mechanics. A viscoelastic ankle attached to a solid-ankle foot improved comfort (McNealy and Gard, 2008) and increased residual leg propulsive GRF impulses (Zmitrewicz et al., 2006) during walking. Others have shown ESAR ankles are able to decrease either socket shear stress magnitude or duration, but not both, in below-knee amputees (Sanders et al., 2000). Studies analyzing increased power output at the ankle joint in non-amputee locomotion has also illustrated the potential influence prosthetic ankles may have on amputee gait. For example, increased ankle power output in non-amputees leads to decreased hip joint powers (Kerrigan et al., 1998; Lewis and Ferris, 2008; Mueller et al., 1994). Non-amputees have also been shown to accommodate changes in stride frequency and surface stiffness by modulating ankle and knee joint stiffness (Farley and Morgenroth, 1999; He et al., 1991; Hobara et al., 2009). Thus, changes in prosthetic ankle stiffness and the levels of energy return will likely have a significant effect on amputee gait mechanics.

The purpose of this study was to identify the influence of prosthetic ankle dorsiflexion and elastic energy storage and return on leg loading, specifically the GRFs and intersegmental joint moments and powers, during steady-state walking. Selective laser sintering, an additive manufacturing technique, was used to fabricate prosthetic ankles with different stiffness levels. In addition, the orientation of the ankles was altered to assess the influence of changes in ankle dorsiflexion and energy storage and return on gait mechanics. We tested four hypotheses: (1) increased dorsiflexion of the ESAR ankles would decrease the vertical GRF peaks; (2) decreased stiffness of the ESAR ankles would increase hip and knee joint moments; (3) increased energy storage and return of the ESAR ankles would increase braking and propulsive GRF impulses (i.e., time integral of the braking and propulsive GRFs); and (4) increased energy return of the ESAR ankles would decrease hip and knee joint powers. Understanding the relationships between ankle dorsiflexion, energy storage and return and leg loading is an important step towards designing effective prosthetic components that improve loading symmetry in amputee gait.

2. Methods

2.1. Prosthetic ankles

A C-shaped articulating ankle was designed in Solidworks (Dassault Systèmes SolidWorks Corp., Concord, MA, USA) to interface with a Seattle Lightfoot2 (Seattle Systems Inc., Poulsbo, WA, USA) low energy prosthetic foot (Fig. 1). Two variations of the ankle (stiff and compliant) were designed by modifying their cross-sectional areas. The ankles were manufactured with Rilsan™ D80 (Arkema Inc., Philadelphia, PA, USA) using selective laser sintering (Beaman et al., 1997; South et al., 2010). The stiffness levels of the ankles were tested by applying a vertical load at a rate of 20 mm/min from 50 N to 1230 N. The stiff ankle had a stiffness of 783 ± 9 N/mm and the compliant ankle had a stiffness of 388 ± 14 N/mm. The opening of the C-shape was oriented to face anterior for the forward-facing ankles (FA) and to face posterior for the reverse-facing ankles (RA). RA was designed to store and return energy during both early and late stance, whereas FA was designed to store and return energy only during late stance. The ankles were compared to a solid ankle (SA) condition, which was a Seattle Lightfoot2 with no ankle. Thus, a total of five foot conditions were analyzed: SA, stiff FA, compliant FA, stiff RA and compliant RA (Fig. 1).

2.2. Subjects

Twelve unilateral below-knee amputees were recruited for this study (12 male; 49 ± 17 years; 82 ± 13 kg; 1.78 ± 0.06 m). All subjects provided informed consent to an Institutional Review Board-approved protocol and had at least 6 months experience walking with a prosthesis (13.7 ± 14.2 years) and no additional walking impairments. For each condition, a certified prosthetist fit the subjects with the prosthesis and assured proper component alignment and pylon length. Subjects walked freely until they were comfortable with the new prosthesis. The ankle conditions were tested in random order. While subjects could visually observe orientation of the ankle, they were blinded to the ankle stiffness.

2.3. Experimental protocol

Reflective markers were placed on the T-2 vertebrae and bilaterally on the shoulder, iliac crest, posterior and anterior superior iliac spine, and greater trochanter. Markers were also placed on the intact leg at the lateral and medial femoral condyles, lateral and medial malleoli, dorsal foot, heel, and first, second, and fifth metatarsal heads, and on the prosthetic leg such that the markers were symmetric with the intact leg. Clusters of four markers each were placed on the shank and thigh of both legs and secured using Coban (3M Inc., St. Paul, MN, USA) to decrease skin movement artifacts. A 10-camera Vicon 612 system (Vicon, Centennial, CO, USA) captured kinematic data at 120 Hz. Three force plates embedded in the walkway measured GRFs at 1200 Hz. For each condition, the subject walked across a 10 m level walkway at 1.20 ± 0.06 m/s until at least two force plate hits were recorded per leg.

2.4. Data processing

Kinematic and GRF data were low-pass filtered in Visual3D (C-Motion Inc., Germantown, MD, USA) using a 4th-order Butterworth filter with cut-off frequencies of 6 and 20 Hz, respectively. The GRFs were normalized by subject body weight. Functional hip, knee and ankle joints were determined from the relative motion of the pelvis, shank, thigh and foot markers (Schwartz and Rozumalski, 2005). Intersegmental joint angles and moments were determined from marker trajectories using standard inverse kinematics and dynamics techniques. Intersegmental joint powers were calculated as the product of the joint moments and corresponding joint angular velocities. Energy stored (returned) by the prosthetic ankle was calculated during early and late stance as the time integral of the negative (positive) ankle joint power.

To test our hypotheses, the maximum prosthetic ankle dorsiflexion and energy storage and return were determined for each condition. GRFs, joint moments and joint powers were computed over stance (approximately 0–60% of the total gait cycle) to determine the effect of the ESAR ankles on leg loading. Peak vertical GRFs were determined during early and late stance (Fig. 2). Braking and propulsive impulses were calculated as the time integral of the negative and positive anterior/posterior (A/P) GRFs, respectively. Peak hip and knee joint moments and powers were identified for each condition (Figs. 3 and 4). All quantities were averaged across trials for each subject at each condition.

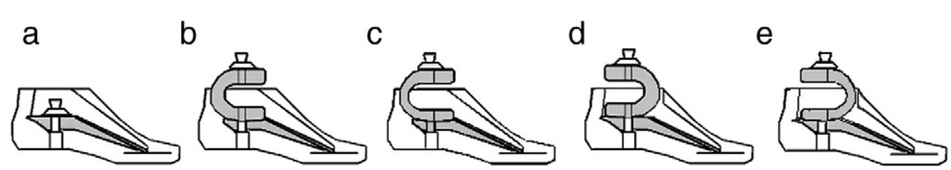

Fig. 1. Foot conditions: (a) solid ankle, SA; (b) stiff forward-facing ankle, FA; (c) compliant FA; (d) stiff reverse-facing ankle, RA; and (e) compliant RA. The pylon attached directly to the Seattle Lightfoot2 for the SA condition, whereas the ankle was attached between the pylon and the Seattle Lightfoot2 for the other four conditions.

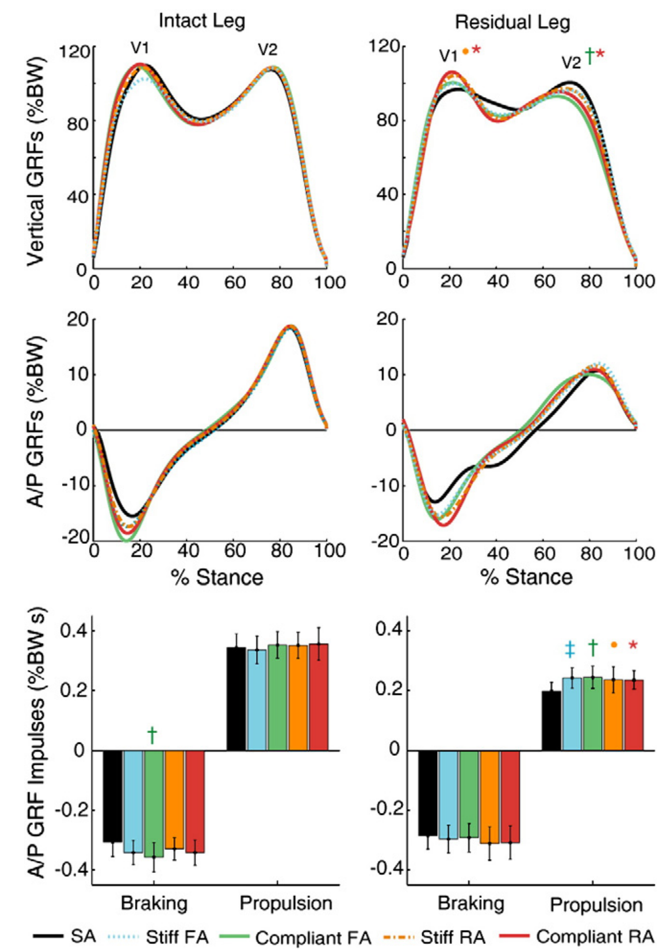

Fig. 2. Intact and residual leg vertical and anterior/posterior (A/P) ground reaction forces (GRFs) over stance and A/P GRF impulses during braking and propulsion. Noted are peak vertical GRFs in early stance (V1) and late stance (V2). Statistically significant differences from SA are noted for stiff FA (‡), compliant FA (†), stiff RA (•) and compliant RA (*).

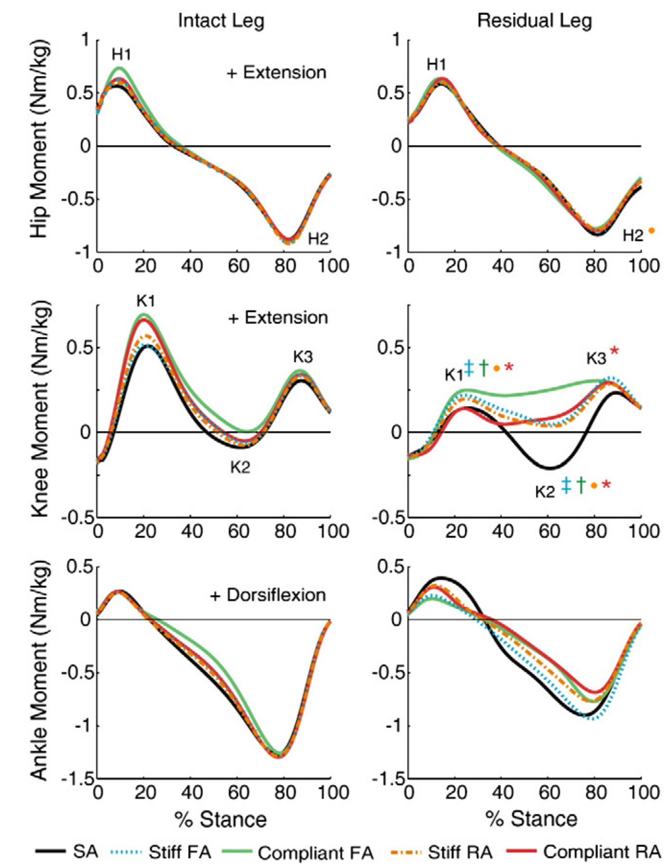

Fig. 3. Joint moments of the intact and residual leg hip, knee and ankle over stance. Noted are the peak hip joint extensor moment (H1), hip joint flexor moment (H2), knee joint extensor moment in early stance (K1), knee joint flexor moment (K2) and knee joint extensor moment in late stance (K3). Statistically significant differences from SA are noted for stiff FA (‡), compliant FA (†), stiff RA (•) and compliant RA (*).

Statistical analyses were performed using SPSS 14.0 (SPSS Inc., Chicago, IL, USA). One-factor repeated-measures ANOVAs were used to compare peak ankle dorsiflexion, energy stored and returned during early and late stance, peak vertical GRFs, braking and propulsive GRF impulses, and peak hip and knee joint moments and powers between ankle conditions. When significant differences were found ($P<0.05$), a Bonferroni adjustment was applied for multiple comparisons to determine which conditions were significantly different from each other.

3. Results

3.1. Ankle properties

All ESAR ankles had significantly greater ankle dorsiflexion ($P<0.001$ for all conditions, Table 1) and increased energy return during late stance relative to SA ($P<0.001$ for all conditions, Table 1).

3.2. Ground reaction forces

Compliant RA and stiff RA increased the first vertical GRF peak of the residual leg relative to SA ($P<0.001$ and $P=0.020$, Fig. 2). Compliant FA and compliant RA decreased the second vertical GRF peak relative to SA ($P=0.005$ and $P=0.021$). The residual leg propulsive GRF impulse was increased for all ankles: stiff FA ($P=0.030$), compliant FA ($P=0.006$), stiff RA ($P=0.026$) and compliant RA ($P=0.012$).

3.3. Joint moments and powers

The first peak knee extensor moment of the residual leg (K1) increased relative to SA and the peak knee flexor moment of the residual leg (K2) for the SA condition changed to an extensor moment for all ESAR ankle conditions ($P<0.001$ for all conditions, Fig. 3). In addition, the second peak knee extensor moment of the residual leg (K3) increased relative to SA for stiff FA ($P=0.002$). Stiff FA also increased the second peak knee joint power absorption (K3) of the residual leg ($P=0.006$).

4. Discussion

Previous studies have shown amputees rely more on their intact leg than their residual leg during walking, which is characterized by greater intact leg GRFs, joint moments and joint powers. This asymmetric loading can lead to additional musculoskeletal disorders, including osteoarthritis in the intact leg, osteoporosis in the residual leg and lower back pain. The goal of the present study was to identify the effects of prosthetic ankle dorsiflexion and energy storage and return on GRFs, joint moments and joint powers to gain insight into how prosthetic ankle design characteristics influence amputee leg loading.

The hypothesis that increased dorsiflexion of the ESAR ankles would result in decreased vertical GRF peaks was supported for the compliant ankles. However, instead of decreasing the first vertical GRF peak of the intact leg after heel strike, as previously shown to occur with ESAR feet (Hansen et al., 2004; Lehmann et al., 1993; Powers et al., 1994; Snyder et al., 1995), the compliant ESAR ankles decreased the second vertical GRF peak of the residual leg during late stance (Fig. 2). While some studies have attributed the decrease in intact leg vertical GRFs to increased ankle

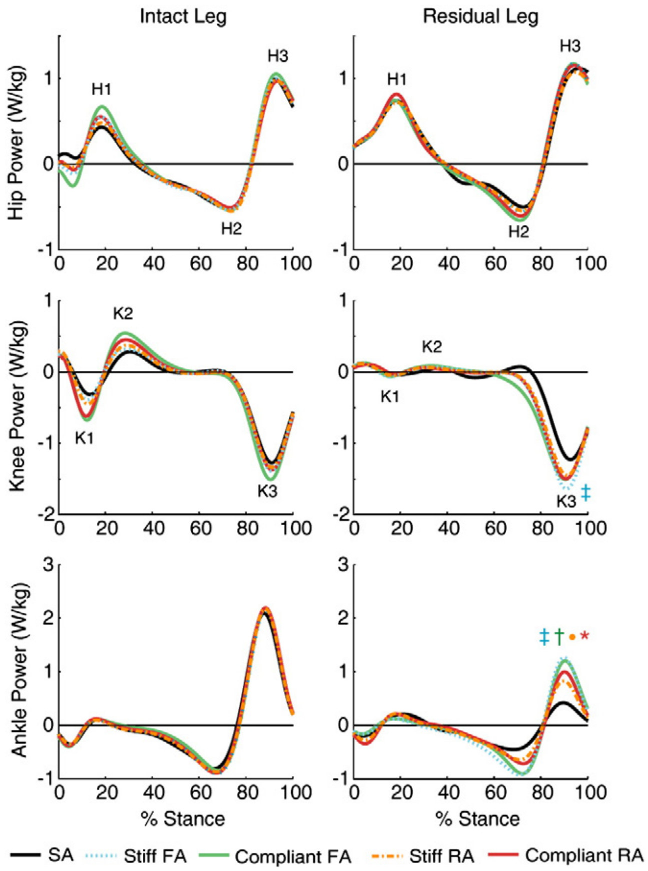

Fig. 4. Joint powers of the intact and residual leg hip, knee and ankle over stance. Noted are the peak hip joint power generation in early stance (H1), hip joint power absorption (H2), hip joint power generation in late stance (H3), knee joint power absorption in early stance (K1), knee joint power generation (K2) and knee joint power absorption in late stance (K3). Energy storage and return of the prosthetic ankle during early and late stance was defined as the negative and positive power impulses of the residual leg ankle joint. Statistically significant differences from SA are noted for stiff FA (‡), compliant FA (†), stiff RA (•) and compliant RA (*).

dorsiflexion that lowers the vertical COM at intact leg heel strike (Powers et al., 1994; Snyder et al., 1995) or improved push-off of the residual leg due to increased energy return of the ESAR feet (Lehmann et al., 1993), other studies have attributed the decrease to a greater anterior center-of-pressure (COP) excursion of the ESAR feet relative to solid-ankle feet (Hansen et al., 2004, 2006). Although energy return was increased (Table 1) and the vertical COM was lowered (Fig. 5a), the anterior COP excursion was not affected by the ESAR ankles in the present study (COP_{ANT}, Fig. 5b). Thus, the differences in vertical GRFs between the ESAR ankles in the present study and ESAR feet in the previous studies (Hansen et al., 2004; Lehmann et al., 1993; Powers et al., 1994; Snyder et al., 1995) may be related to differences in anterior COP excursion. The decreased peak vertical GRF of the residual leg during late stance is most likely related to the ESAR ankles lowering the vertical COM relative to SA.

The hypothesis that amputees would increase residual leg knee and hip joint moments to compensate for decreased stiffness of the ESAR ankles was supported at the knee (Fig. 3). The residual leg knee flexor moment (K2) changed to an extensor moment for all ESAR ankles when the ankle joint was maximally dorsiflexed. Ankle and knee joint stiffness levels are the primary mechanisms by which leg stiffness is modulated to accommodate changes in stride frequency and surface stiffness during non-amputee hopping and running (Farley and Morgenroth, 1999; He et al., 1991; Hobara et al., 2009). However, a jumping model showed that changes in ankle stiffness have a much greater effect on total leg stiffness than changes in knee stiffness (Farley and Morgenroth, 1999). These results are consistent with the findings of the present study, where an average decrease in maximum ankle plantar flexor moment of 25–50% across conditions for the ESAR ankles relative to SA resulted in an average increase in knee joint moment (K2) of 102–173%.

The hypothesis that increased energy return of the ESAR ankles would result in increased braking and propulsive GRF impulses was supported for propulsion only (Fig. 2). There was an increase in residual leg propulsion, which appears to be primarily a result of increased propulsive GRF duration for all conditions and also increased propulsive GRF magnitude for the stiff conditions (Fig. 2). The increased duration was consistent with previous work that compared braking and propulsive GRF impulses between prosthetic feet with and without a commercial viscoelastic ankle joint (Zmitrewicz et al., 2006). Compliant FA also increased the braking GRF impulses of the intact leg (Fig. 2). It is possible that subjects began braking on their intact leg to curb ankle dorsiflexion when using the compliant FA, which reached greater dorsiflexion than the intact leg ankle (Fig. 6).

The final hypothesis that increased energy return of the ESAR ankles would result in decreased hip and knee joint powers was not supported. Previous studies have shown that amputees compensate for low power generation from prosthetic feet by increasing hip power generation (Silverman et al., 2008) and, conversely, that non-amputees who increase ankle joint power generation also decrease hip joint power generation (Kerrigan et al., 1998; Lewis and Ferris, 2008; Mueller et al., 1994). The difference between the present and previous studies may be related to the increase in knee joint power absorption (K3) at the same time the ESAR ankles increased ankle joint power generation (Fig. 4), whereas knee joint power absorption decreased with increased ankle joint power generation in non-amputees (Lewis and Ferris, 2008). The increase in knee joint power absorption in the present study is likely caused by the increased ankle dorsiflexion of the ESAR ankles, which causes an increase in knee extensor moments. Therefore, it is possible that the very mechanism by which energy is stored and returned, ankle dorsiflexion and plantarflexion, is the same mechanism that decreases the effectiveness of the ESAR ankles.

One limitation with our experimental analysis is that causality can be difficult to identify. In the present study, ESAR ankles with varying stiffness and orientations were used to systematically alter ankle dorsiflexion and energy storage and return. Ankle dorsiflexion and energy storage and return were not only dependent on the ankle stiffness and orientation, but also on subject-specific walking mechanics, which varied greatly between subjects. Muscle-actuated forward dynamics simulations can be used to quantify the contribution of the prosthetic ankle joint to not only GRFs and joint moments and powers, but also joint contact forces

Table 1
Average (SD) ankle properties: peak dorsiflexion, energy storage and return in early stance, and energy storage and return in late stance. (*) denotes a significant difference from SA ($P<0.05$).

	Peak dorsiflexion (°)	Energy storage - early stance (J/kg)	Energy return - early stance (J/kg)	Energy storage - late stance (J/kg)	Energy return - late stance (J/kg)
SA	9.6 (3.0)	0.013 (0.007)	0.031 (0.024)	0.102 (0.045)	0.043 (0.013)
Stiff FA	20.0 (2.8)*	0.009 (0.006)	0.016 (0.013)	0.149 (0.039)	0.109 (0.030)*
Compliant FA	23.1 (2.4)*	0.008 (0.005)	0.016 (0.015)	0.138 (0.077)	0.109 (0.053)*
Stiff RA	17.3 (2.4)*	0.018 (0.007)	0.026 (0.018)	0.127 (0.028)	0.084 (0.021)*
Compliant RA	20.3 (2.1)*	0.021 (0.009)	0.027 (0.021)	0.119 (0.057)	0.087 (0.030)*

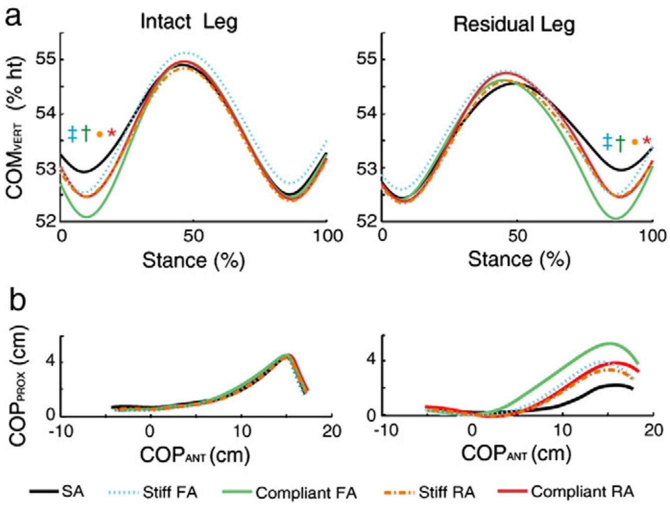

Fig. 5. (a) Vertical center-of-mass (COM_{VERT}) position during intact leg and residual leg stance and (b) center-of-pressure excursion in the proximal (COP_{PROX}) and anterior (COP_{ANT}) directions with respect to the intact and residual leg ankles. Statistically significant differences in peak COM_{VERT} from SA are noted for stiff FA (‡), compliant FA (†), stiff RA (•) and compliant RA (*).

(e.g. Sasaki and Neptune, 2010; Winby et al., 2009; Zmitrewicz et al., 2007). Future studies using subject-specific simulations will be important for determining causal relationships between ankle characteristics and leg loading quantities for individual walking mechanics, which can be used to guide prosthetic foot prescription to minimize leg loading.

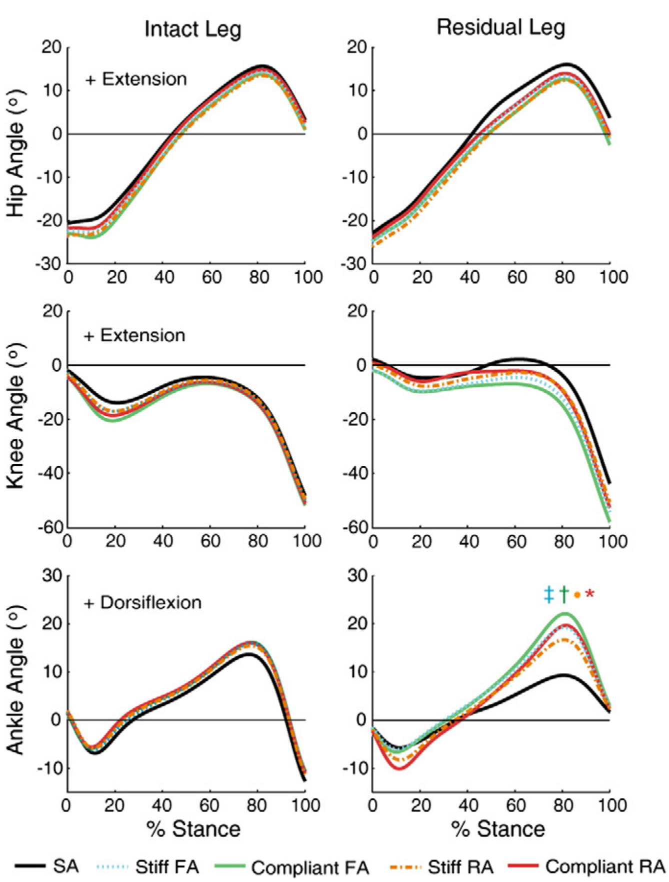

Fig. 6. Joint angles of the intact and residual leg hip, knee and ankle over stance. Statistically significant differences of the maximum residual leg ankle dorsiflexion from SA are noted for stiff FA (‡), compliant FA (†), stiff RA (•) and compliant RA (*).

The results showed that ESAR ankles, which increase ankle dorsiflexion and energy storage and return of the prosthetic foot, decrease residual leg vertical GRFs in late stance, increase residual leg propulsive GRF impulses, and increase residual leg knee joint extensor moments. The reverse-facing ankles increased both ankle energy storage and residual leg vertical GRFs in early stance. The compliant forward-facing ankle, with large increases in ankle dorsiflexion and energy return in late stance, increased both residual leg braking and intact leg knee joint powers. In contrast to previous studies, the increased energy storage and return of the ESAR ankles did not decrease hip joint powers or the intact leg vertical GRFs, likely because of increased power absorption at the knee that is associated with increased ankle dorsiflexion and differences in the anterior COP excursion, respectively. Understanding the effects of ankle dorsiflexion and energy return on amputee gait is an important step towards the design of effective prosthetic components that decrease leg loading asymmetries in amputee gait.

Acknowledgements

The authors thank Nick Fey and the members of the VA Center of Excellence for Limb Loss Prevention and Prosthetic Engineering for help with data collection. This research was supported by the Department of Veterans Affairs, Rehabilitation Research and Development Service (A4376R) and the National Science Foundation under grant 0346514.

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