The effect of prosthetic ankle energy storage and return properties on muscle activity in below-knee amputee walking

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ABSTRACT

In an effort to improve amputee gait, energy storage and return (ESAR) prosthetic feet have been developed to provide enhanced function by storing and returning mechanical energy through elastic structures. However, the effect of ESAR feet on muscle activity in amputee walking is not well understood. Previous studies have analyzed commercial prosthetic feet with a wide range of material properties and geometries, making it difficult to associate specific ESAR properties with changes in muscle activity. In contrast, prosthetic ankles offer a systematic way to manipulate ESAR properties while keeping the prosthetic heel and keel geometry intact. In the present study, ESAR ankles were added to a Seattle Lightfoot2 to carefully control the energy storage and return by altering the ankle stiffness and orientation in order to identify its effect on lower extremity muscle activity during below-knee amputee walking. A total of five foot conditions were analyzed: solid ankle (SA), stiff forward-facing ankle (FA), compliant FA, stiff reverse-facing ankle (RA) and compliant RA. The ESAR ankles decreased the activity of muscles that contribute to body forward propulsion and increased the activity of muscles that provide body support. The compliant ankles generally caused a greater change in muscle activity than the stiff ankles, but without a corresponding increase in energy return. Ankle orientation also had an effect, with RA generally causing a lower change in muscle activity than FA. These results highlight the influence of ESAR stiffness on muscle activity and the importance of prescribing appropriate prosthetic foot stiffness to improve rehabilitation outcomes.

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1. Introduction

Over one million amputees are living in the United States with major lower limb loss [1]. Lower limb amputation leads to the functional loss of the ankle plantar flexors, which are important contributors to body support, forward propulsion and leg swing initiation during walking [2,3]. To compensate for the loss of the ankle muscles, amputees have been shown to compensate through increased residual leg hamstring and rectus femoris activity during early stance and pre-swing, respectively [4–7]. These results are consistent with studies of non-amputee walking, where the hamstrings have been shown to contribute to body propulsion in early stance and the rectus femoris redistributes power from the leg to propel the trunk forward and extends the hip and knee to provide body support [8,9]. Amputees also have prolonged residual leg vasti activity compared to speed-matched non-amputees from early to mid-stance [4,6,7]. This compensatory mechanism suggests a need for increased body support in the absence of the plantar flexors, which is consistent with previous studies of non-amputee walking showing the vasti provide body support in early stance [8,9].

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 when the foot is in contact with the ground and returning it prior to the swing phase. ESAR feet with articulating ankles can return up to five times the total energy returned by solid ankle feet [10,11]. A previous electromyographic (EMG) study showed amputees have decreased residual leg hamstring activity and delayed residual leg vasti activity when wearing ESAR feet versus solid ankle feet [12]. More recent studies have found no significant differences in EMG activity between feet, although study participants were not required to walk at the same walking speed when comparing the different feet [13,14]. Differences in walking speeds may have confounded these results, since walking speed alters gait mechanics [15,16]. Another potential confounding factor is that the previous EMG studies

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Table 1
Average (±standard deviation) stride width (m), stance time ratio (residual/intact), step length ratio (residual/intact), and double support time (s) for the four ankle conditions: solid-ankle (SA), stiff force-facing ankle (FA), compliant FA, stiff reverse-facing ankle (RA) and compliant RA.

	Stride width	Stance time ratio (R/I)	Step length ratio (R/I)	Double support time
SA	0.15 ± 0.02	0.97 ± 0.03	1.21 ± 0.11	0.31 ± 0.03
Stiff FA	0.14 ± 0.03	0.95 ± 0.06	1.15 ± 0.06	0.33 ± 0.03
Compliant FA	0.13 ± 0.03	0.96 ± 0.04	1.16 ± 0.06	0.33 ± 0.03
Stiff RA	0.14 ± 0.03	0.97 ± 0.04	1.16 ± 0.07	0.32 ± 0.03
Compliant RA	0.13 ± 0.03	0.96 ± 0.06	1.20 ± 0.10	0.32 ± 0.03

analyzed feet with a wide range of geometries and material properties, making it difficult to associate specific ESAR properties to changes in muscle activity.

Prosthetic ankles offer a systematic way to manipulate ESAR properties by altering the amount of elastic energy stored and returned while keeping the prosthetic heel and keel geometry intact. Compliant prosthetic ankles attached to solid ankle feet have been shown to have a positive influence on gait mechanics over solid ankle feet alone by improving amputee comfort [17], increasing propulsive ground reaction forces [18], and redistributing socket pressure and shear stresses [19]. In addition, improvements in gait mechanics between solid ankle and ESAR feet have been attributed to increases in ankle range of motion [20]. In the present study, a novel manufacturing technique was used to generate ESAR ankles with different stiffness levels to identify how energy storage and return properties influence lower extremity muscle activity during below-knee amputee walking. The magnitude and timing of energy storage and return were controlled by altering the ankle stiffness and orientation (forward versus reverse facing ankles).

The ESAR ankles were designed to contribute more to forward propulsion than the solid ankle foot alone. Because amputees have greater activity than non-amputees in muscles that contribute to body propulsion (i.e., prolonged residual leg hamstring activity [4–7]), we expected a decrease in the activity of these muscles for ESAR ankles compared to solid ankles. In addition, we expected that the compliant ESAR ankles would provide less body support than the stiff ESAR and solid ankles, and therefore result in increased activity of muscles that provide support (i.e., residual leg vasti and rectus femoris activity [8,9]). Reverse-facing ankles were designed to provide added energy return during early stance, and therefore were expected to require less activity from muscles that provide forward propulsion than forward-facing ankles when compared to the control condition, particularly the residual leg hamstrings [8,9].

2. Methods

2.1. Prosthetic ankles

A C-shaped dynamic response ankle was designed in Solidworks (Dassault Systèmes SolidWorks Corp.) to interface with a Seattle Lightfoot2 (Seattle Systems, Inc.) prosthetic foot using a standard pyramid adapter (Fig. 1). Two variations of the ankle, stiff and compliant, were designed by modifying the cross-sectional areas of the ankles. The ankles were manufactured with Rilsan™ D80 (Arkema, Inc.) using selective laser sintering [21,22]. The ankle stiffness levels were tested without the Seattle Lightfoot2 attached by applying a vertical load to the location where the ankle interfaces with the pyramid adapter 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 ankle (FA) and posterior for the reverse-facing ankle (RA). RA was designed to increase energy storage in two regions of the gait cycle, during weight acceptance and single-limb support, and energy return during early stance and pre-swing, respectively. FA was designed to only increase energy storage during single-limb support and energy return during pre-swing. The ankles were compared to a solid ankle (SA) condition, which was the Seattle Lightfoot2 with no ankle. Thus, there were a total of five ankle conditions 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 yrs; 82 ± 13 kg; 1.78 ± 0.06 m). All subjects provided informed consent to an Institutional Review Board approved protocol, had at least 6 months experience walking with a prosthesis, and had no additional walking impairments. For each condition, a certified prosthetist fit the subjects with the ankle attached to a Seattle Lightfoot2 of appropriate length for the subject and assured proper component alignment and pylon length. Subjects walked freely until they were comfortable with the new prosthesis (Table 1). The ankle conditions were tested in random order.

2.3. Experimental protocol

To measure the body segment kinematics, reflective markers (14-mm diameter) 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.) to decrease skin movement artifacts.

Pairs of surface EMG electrodes (10-mm disk sensor contacts, 16-mm interelectrode distance, 10–500 Hz signal bandwidth) were used to measure muscle activity of the intact leg tibialis anterior (TA), medial gastrocnemius (GAS), soleus (SOL), vastus lateralis (VAS), rectus femoris (RF), biceps femoris long head (HAM), gluteus medius (GMED) and gluteus maximus (GMAX), and the residual leg VAS, RF, HAM, GMED and GMAX. Electrodes were placed along the muscle line of action between the origin and insertion points as described by Perotto and Delagi [23]. The electrodes were placed on skin that was first shaved and cleaned with alcohol to decrease impedance, and then secured using Coban (3M, Inc.). A ground electrode was placed on the sacrum.

For each condition, the subject walked across a 10-m level walkway at 1.20 ± 0.06 m/s until six force plate hits were recorded per leg. A 10-camera Vicon 612 system captured kinematic data at 120 Hz. Three force plates embedded in the walkway were used to measure ground reaction forces (GRFs) at 1200 Hz and EMG data were collected using a wireless system (Noraxon USA, Inc.) at 2160 Hz.

2.4. Data processing and analysis

Kinematic data and GRFs were low-pass filtered in Visual 3D (C-Motion, Inc.) using a 4th-order Butterworth filter with cut-off frequencies of 6 Hz and 20 Hz, respectively. Functional hip, knee and ankle joints were determined from relative motion of the pelvis, shank, thigh and foot [24]. 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 defined as the time integral of the negative (positive) power at the ankle joint. Ankle joint calculations were

— SA   ⋯ Stiff FA   — Compliant FA   ⋯ Stiff RA   — Compliant RA

Fig. 1. Ankle 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 ankle conditions.
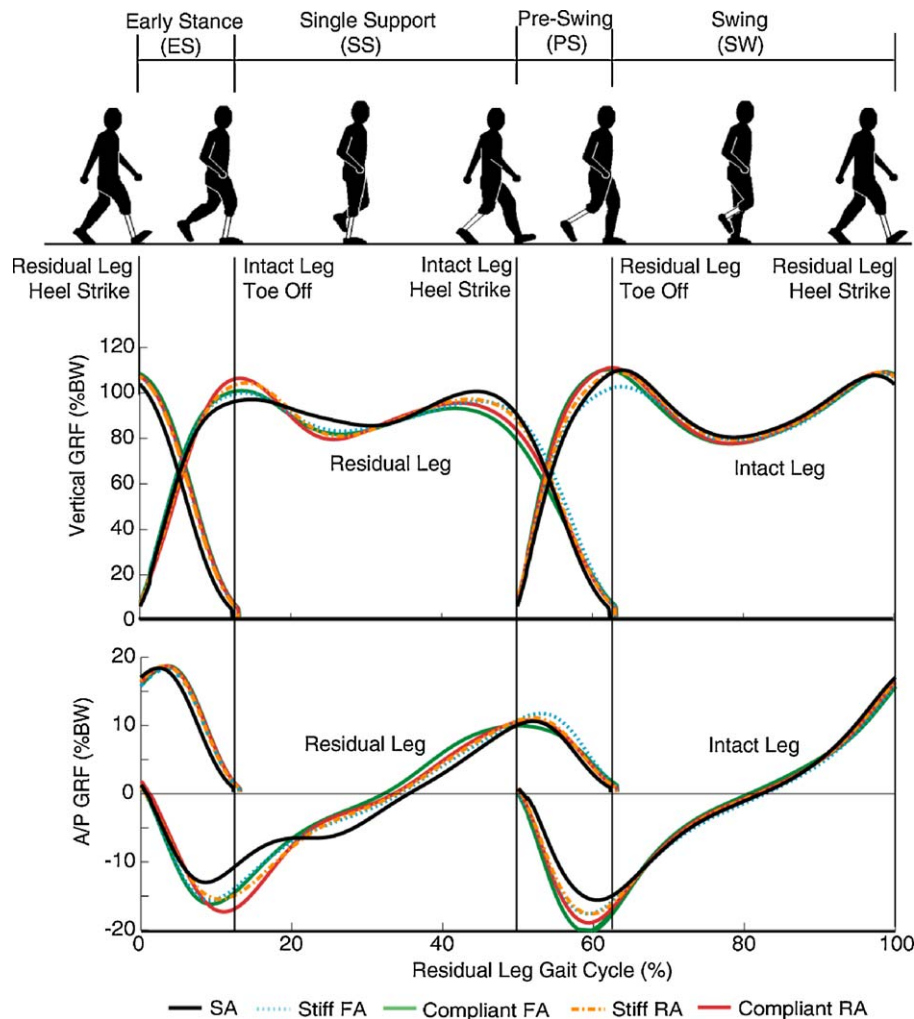
Fig. 2. Regions of the residual leg gait cycle and corresponding vertical and anterior/posterior (A/P) ground reaction forces (GRFs) of the intact and residual legs. Early stance (ES) is defined from residual leg heel strike to intact leg toe-off; single support (SS) follows until intact leg heel-strike; pre-swing (PS) follows until residual leg toe-off, and swing (SW) follows until residual leg heel-strike.

performed assuming a rigid foot segment with a single stationary rotation axis. For each condition, the prosthetic ankle angle was measured with no force applied to obtain its zero value. Energy storage and return was determined in two phases based on the power curve, with the first phase (Phase 1) occurring from approximately 0–20% and the second phase (Phase 2) from approximately 20% to 65% of the gait cycle.

The raw EMG signals were demeaned, smoothed using a moving root-mean-square (RMS) window of 80 ms, and time-normalized to the gait cycle. To quantify differences between SA and the other conditions, subject average RMS profiles for each muscle at each condition were normalized to the average peak RMS value of the same muscle during the SA condition. For statistical analysis, muscle activity was averaged within specific regions of the residual leg gait cycle: early stance (~0–15% gait cycle), single support (~15–45% gait cycle), pre-swing (~45–60% gait cycle) and swing (~60–100% gait cycle, Fig. 2). The RMS profiles and region specific average EMG magnitudes were then averaged across subjects for each condition.

Statistical analyses were performed to determine differences between SA and the four other conditions. The energy stored and returned for each of the two power phases and the average EMG magnitudes for each of the four regions of the gait cycle were compared using Student's t-tests, with a Bonferroni adjustment for multiple comparisons, to identify which values were significantly different ($p < 0.0125$).

3. Results

Stiff and compliant RA significantly increased energy storage relative to SA in Phase 1 (~0–20% gait cycle, p < 0.001 for both conditions), but without an increase in energy return (Fig. 3). Stiff and compliant FA decreased both energy storage and return relative to SA in Phase 1 ($p < 0.001$ for both conditions). All ESAR ankles increased energy return in Phase 2 (~20–65% gait cycle, $p < 0.001$ for all conditions), with the FA conditions having greater increases relative to the RA conditions.

The stiff forward-facing ankle (FA) caused greater residual leg RF activity ($p < 0.001$) and lower residual leg HAM activity ($p < 0.001$) relative to SA during single support (Fig. 4). During pre-swing, stiff FA resulted in lower residual leg HAM activity ($p < 0.001$) relative to SA.

The compliant forward-facing ankle (FA) caused greater residual leg VAS and RF activity ($p < 0.001$ and <0.001), lower intact leg VAS ($p < 0.001$) and greater intact leg GMAX activity ($p < 0.001$) relative to SA during early stance of the residual leg (Figs. 4 and 5). During single support, compliant FA resulted in greater residual leg RF, VAS and GMED activity ($p < 0.001$, <0.001, <0.001), greater intact leg GAS activity ($p = 0.002$), and lower residual leg HAM activity ($p < 0.001$). During pre-swing, compliant FA resulted in greater intact leg VAS and RF activity ($p < 0.001$ and 0.008), greater residual leg RF, GMED and GMAX activity ($p = 0.002$, <0.001 and 0.002), and lower intact leg GAS and residual leg HAM activity ($p < 0.001$ and <0.001) relative to SA.

The stiff reverse-facing ankle (RA) resulted in lower intact leg GMAX activity ($p = 0.001$) relative to SA during early stance of the residual leg (Fig. 5). During single support, stiff RA also caused greater residual leg RF and intact leg GAS activity ($p < 0.001$ and <0.001) and lower residual leg HAM activity ($p < 0.001$). During
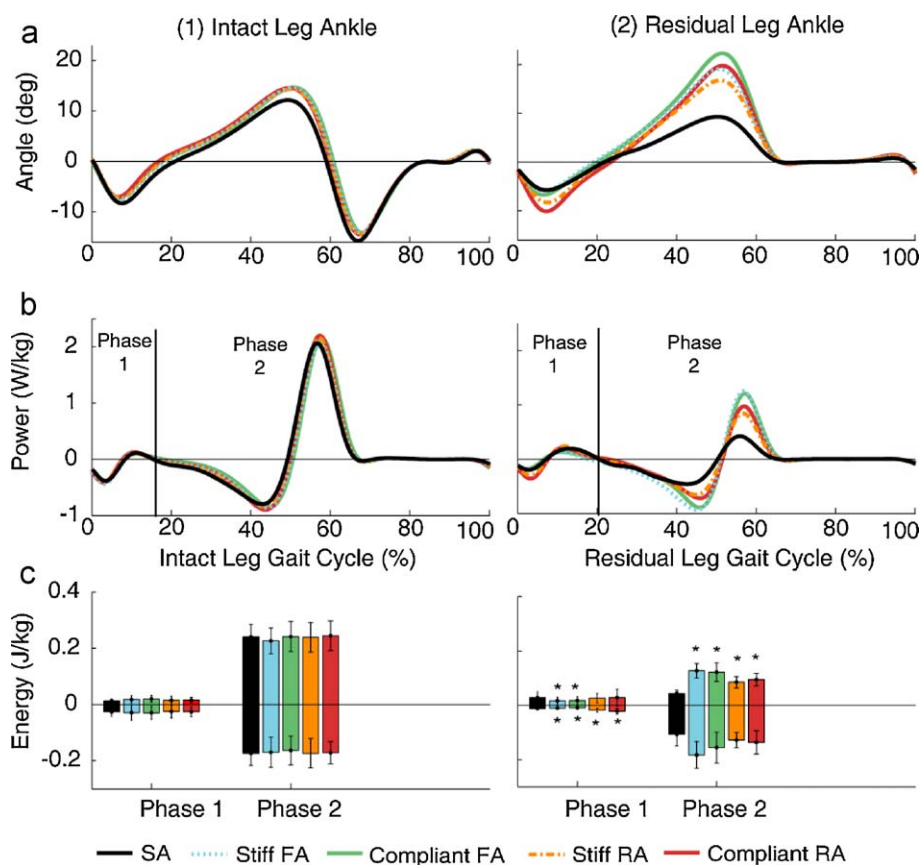
Fig. 3. Intact and Residual Leg (a) average ankle angle, (b) average ankle power, and (c) average ankle energy storage and return during power phase 1 (~0–20%) and power phase 2 (~20–60%). Negative and positive values indicate energy stored and returned, respectively. (*) denotes a significant difference from the SA condition ($p < 0.0125$).

pre-swing, stiff RA resulted in lower residual leg VAS and HAM activity ($p < 0.001$ and <0.001) and greater residual leg GMED activity ($p < 0.001$) relative to SA.

The compliant reverse-facing ankle (RA) resulted in lower intact leg VAS and GMAX activity ($p < 0.001$ and 0.007) during early stance of the residual leg (Fig. 5). During single support, compliant RA caused greater residual leg RF activity ($p < 0.001$) and lower residual leg HAM activity ($p < 0.001$) relative to SA. During pre-swing, compliant RA resulted in greater intact leg VAS and residual leg GMAX activity ($p < 0.001$ and <0.001) and lower residual leg VAS and HAM activity ($p = 0.003$ and <0.001) relative to SA (Figs. 4 and 5).

4. Discussion

The purpose of this study was to identify relationships between ESAR stiffness properties of prosthetic ankles and lower extremity muscle activity during amputee walking. The energy stored and returned by the ESAR ankles was expected to cause a decrease in activity of those muscles that normally provide forward propulsion. We also expected that the compliant ankles would result in greater changes in activity of muscles that provide body support compared to SA feet due to increased ankle dorsiflexion. Reverse-facing ankles were expected to provide added energy return during early stance and thereby require less activity from muscles that provide forward propulsion than the forward-facing ankles.

All ESAR ankles caused an increase in residual leg RF activity and a decrease in residual leg HAM activity in single support relative to the SA condition. Previous studies have shown that, although an anatomical hip flexor, RF acts to extend both the hip and knee [8,25,26]. Thus, increased RF activity suggests the ESAR

ankles require more body support from muscles than the solid ankle. Previous studies analyzing below-knee amputees wearing solid ankle feet have shown significantly more HAM activity during stance than non-amputees [5]. HAM has been shown to contribute to body propulsion in non-amputee walking [8,9]. Therefore, a decrease in residual leg HAM activity with ESAR ankles suggests that the ankles provide more body forward propulsion than the SA feet alone.

Although RA allowed energy to be stored in two regions of the gait cycle, there was no increase in energy return in Phase 1 (~0–20% gait cycle). As a result, activity of residual leg hamstring was similar for both RA and FA. We also found the average energy return was greater for FA than RA in Phase 2 (~20–65% gait cycle). Although stiff FA had greater energy return than stiff RA, muscle activity was similar between the two conditions. In contrast, the compliant ankles caused higher overall muscle activity than the stiff ankles and there were significant differences in muscle activity between compliant FA and compliant RA. During residual leg pre-swing, the compliant ankles caused a significant increase in intact leg VAS activity. VAS has been shown to contribute to body support and deceleration during non-amputee walking [8,9], the latter function being consistent with the increase in negative A/P GRFs of the intact leg observed in this region of the gait cycle (Fig. 2). An increase in VAS activity suggests that the increased dorsiflexion of the compliant ankles in pre-swing required increased body support from the intact leg. Also, compliant FA had a larger increase in residual leg VAS and RF activity compared to stiff FA but without an increase in energy return. The relative trade-offs between muscle activity and energy return between conditions highlight the importance of prescribing proper prosthetic components for a given subject. Excessive prosthetic foot-
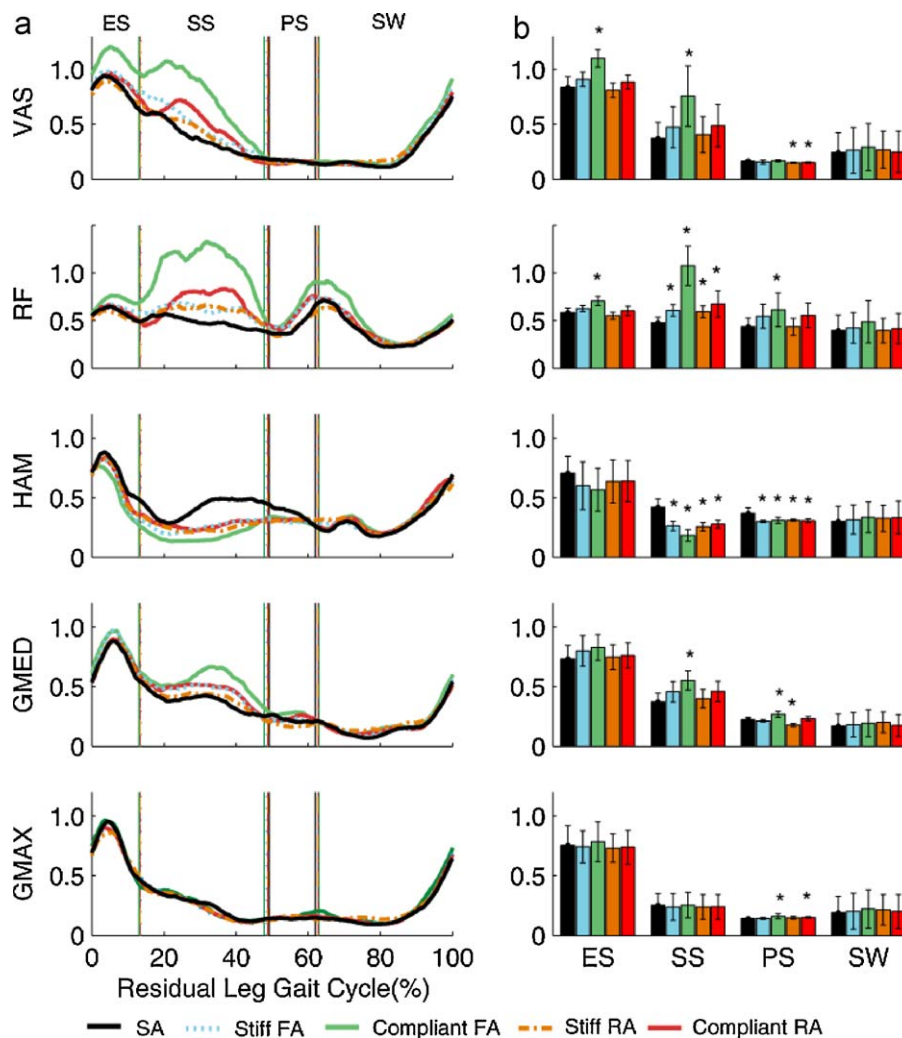
Fig. 4. Residual leg muscle activity. (a) RMS profiles of residual leg muscle activity over the gait cycle. Vertical lines denote average heel strike and toe-off events for each condition. (b) Average and standard deviations of residual leg muscle activity during early stance (ES), single support (SS), pre-swing (PS) and swing (SW). (*) denotes a significant difference from the SA condition ($p < 0.0125$).

ankle compliance could lead to increased intact knee joint loading due to higher intact leg VAS activity and GRFs.

Overall, the ESAR ankles decreased the activity of muscles that contribute to body forward propulsion (HAM) and increased dependence on muscles that provide body support (VAS and RF). These results suggest that ESAR ankles provide added propulsion, but at the cost of reduced body support. A simulation analysis of symmetric amputee walking found that when an ESAR ankle is stiff enough to provide the body support, there is a deficit in providing the forward propulsion normally provided by the plantar flexors [27]. Thus, there appears to be an optimal ankle stiffness that balances the need for both body support and forward propulsion. The optimal ankle stiffness has the potential of decreasing both joint loading and metabolic cost of amputee walking by reducing required muscle compensation. Further work is needed to understand the relationships between an amputee's walking mechanics and the design characteristics of the ankle-foot unit (e.g., rollover shape [28] and foot stiffness [22]) to develop prosthetic feet that provide optimal support and energy return to improve amputee mobility.

One limitation of the current study is that uniarticular hip-flexor muscle activity was not measured. Others have shown increased intact leg hip flexor moments for amputees compared to non-amputees [29,30], which is a compensatory mechanism also

seen in simulation analyses [27]. Another limitation is that EMG only measures muscle activity and does not quantify muscle force, which is important in determining muscle contributions to providing body support and forward propulsion. Future work using muscle-actuated forward dynamics simulations are needed to analyze muscles that are difficult to measure with EMG (e.g., hip flexors) and also to quantify each muscle's contribution to body support and forward propulsion [8,27].

Another potential limitation was the high variability in amputee response to the different ankle conditions, as each subject responded differently to the ESAR ankles depending on their walking mechanics. For example, some subjects were able to achieve a higher ratio of energy return to energy storage from compliant RA than from compliant FA because their center of mass was more forward than other subjects. The varying weight of the participants could also influence their response to the ankles. Future studies should analyze ankles with stiffness values proportional to each participant's body weight. In addition, a larger subject pool would allow amputees to be divided into subgroups based on walking mechanics to begin relating ESAR properties to specific amputee gait characteristics. Another source of the variability could be due to different acclimation times needed by each subject. Each subject was allowed as much time as needed to acclimate to the new ankle, but further adaptations may
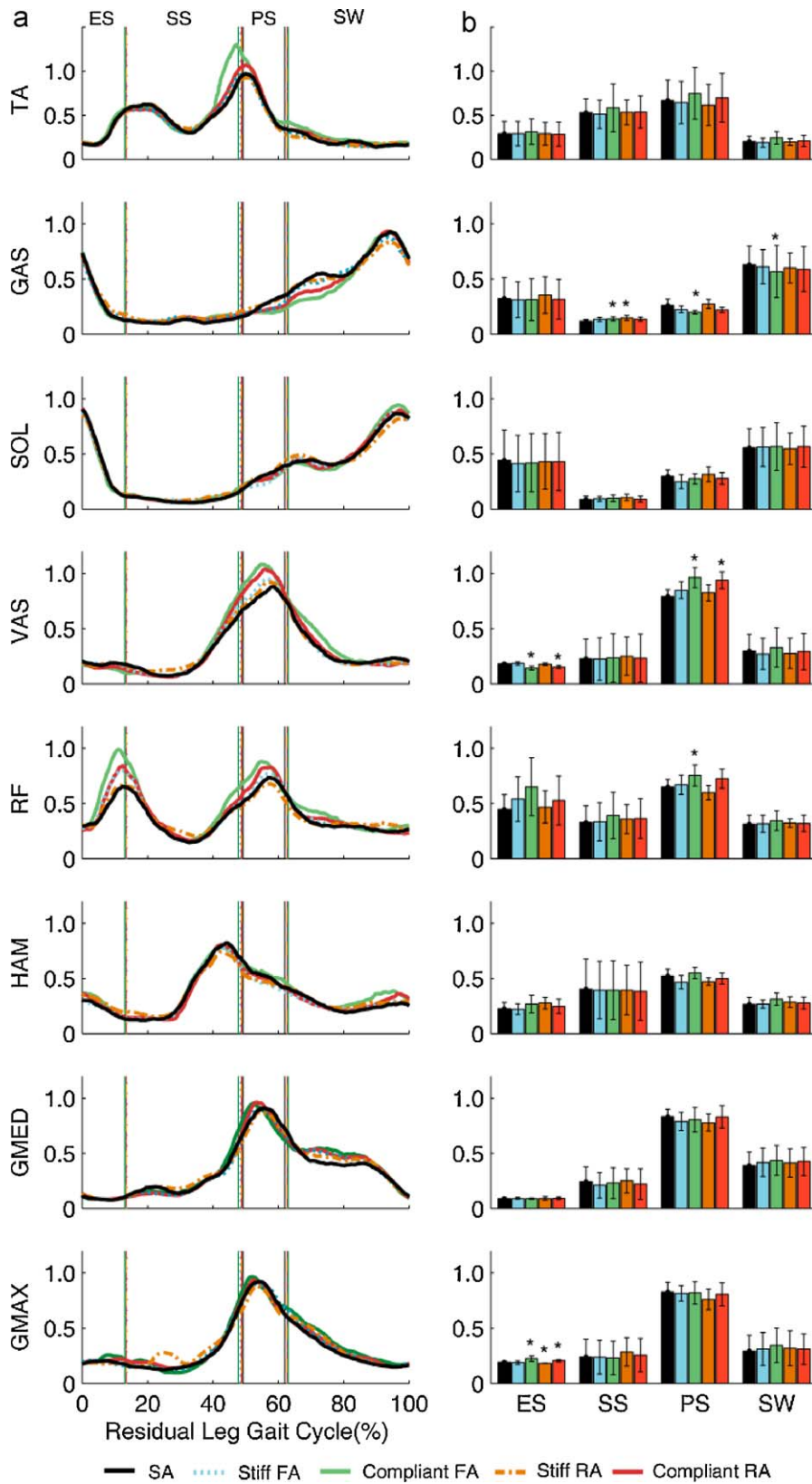
Fig. 5. Intact leg muscle activity. (a) RMS profiles of intact leg muscle activity over the gait cycle. Vertical lines denote average heel strike and toe-off events for each condition. (b) Average and standard deviations of intact leg muscle activity during early stance (ES), single support (SS), pre-swing (PS) and swing (SW). (*) denotes a significant difference from the SA condition ($p < 0.0125$).

occur over time. Developing formal methods to assess acclimation remains an important area for future work.

Energy storage and return of the prosthetic ankles was calculated from the ankle power input and output assuming a rigid foot segment with a single stationary rotation axis, which is consistent with previous prosthetic foot studies [11,31]. However, the axis of rotation of a flexible prosthetic foot-ankle unit may be better modeled by a moving axis along the foot or by a finite element model [32]. Thus, future studies may consider integrating such alternatives to improve the fidelity of the energy calculations.

In summary, the ESAR ankles decreased the activity of muscles that contribute to body forward propulsion (HAM) and increased activity of those muscles that provide body support (VAS and RF). The compliant ankles generally caused greater changes in muscle activity than the stiff ankles, but without a corresponding increase in energy return. Ankle orientation did affect muscle activity, with RA generally causing a lower change in muscle activity than FA. The results from this study highlight the importance of proper prosthetic foot stiffness prescription for amputees to assure effective rehabilitation outcomes. Future studies should be directed towards identifying an optimal ankle stiffness for a given subject that helps minimize necessary muscle compensations, which would improve amputee mobility by potentially decreasing joint loading and metabolic cost.

Acknowledgements

Conflict of interest statement

There is no conflict of interest.

References

[1] Ziegler-Graham K, MacKenzie EJ, Ephraim PL, Travison TG, Brookmeyer R. Estimating the prevalence of limb loss in the United States: 2005 to 2050. Arch Phys Med Rehabil 2008;89:422–9.

[2] Neptune RR, Kautz SA, Zajac FE. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. J Biomech 2001;34:1387–98.

[3] Liu MQ, Anderson FC, Pandy MG, Delp SL. Muscles that support the body also modulate forward progression during walking. J Biomech 2006;39:2623–30.

[4] Fey NP, Silverman AK, Neptune RR. The influence of increasing steady-state walking speed on muscle activity in below-knee amputees. J Electromyogr Kinesiol 2010;20:155–61.

[5] Isakov E, Burger H, Krajnik J, Gregoric M, Marincek C. Knee muscle activity during ambulation of trans-tibial amputees. J Rehabil Med 2001;33:196–9.

[6] Powers CM, Rao S, Perry J. Knee kinetics in trans-tibial amputee gait. Gait Posture 1998;8:1–7.

[7] Winter DA, Sienko SE. Biomechanics of below-knee amputee gait. J Biomech 1988;21:361–7.

[8] Neptune RR, Zajac FE, Kautz SA. Muscle force redistributes segmental power for body progression during walking. Gait Posture 2004;19:194–205.

[9] Liu MQ, Anderson FC, Schwartz MH, Delp SL. Muscle contributions to support and progression over a range of walking speeds. J Biomech 2008;41:3243–52.

[10] Geil MD. Energy loss and stiffness properties of dynamic elastic response prosthetic feet. Prosthet Orthot Sci 2001;13:70–3.

[11] Ehara Y, Beppu M, Nomura S, Kunimi Y, Takahashi S. Energy storing property of so-called energy-storing prosthetic feet. Arch Phys Med Rehabil 1993;74:68–72.

[12] Culham EG, Peat M, Newell E. Below-knee amputation: a comparison of the effect of the SACH foot and single axis foot on electromyographic patterns during locomotion. Prosthet Orthot Int 1986;10:15–22.

[13] Torburn L, Perry J, Ayyappa E, Shanfield SL. Below-knee amputee gait with dynamic elastic response prosthetic feet: a pilot study. J Rehabil Res Dev 1990;27:369–84.

[14] Perry J, Shanfield S. Efficiency of dynamic elastic response prosthetic feet. J Rehabil Res Dev 1993;30:137–43.

[15] Neptune RR, Sasaki K, Kautz SA. The effect of walking speed on muscle function and mechanical energetics. Gait Posture 2008;28:135–43.

[16] Chiu MC, Wang MJ. The effect of gait speed and gender on perceived exertion, muscle activity, joint motion of lower extremity, ground reaction force and heart rate during normal walking. Gait Posture 2007;25:385–92.

[17] McNealy LL, Gard SA. Effect of prosthetic ankle units on the gait of persons with bilateral trans-femoral amputations. Prosthet Orthot Int 2008;32:111–26.

[18] Zmitrewicz RJ, Neptune RR, Walden JG, Rogers WE, Bosker GW. The effect of foot and ankle prosthetic components on braking and propulsive impulses during transtibial amputee gait. Arch Phys Med Rehabil 2006;87:1334–9.

[19] Sanders JE, Greve JM, Clinton C, Hafner BJ. Changes in interface pressure and stump shape over time: preliminary results from a trans-tibial amputee subject. Prosthet Orthot Int 2000;24:163–8.

[20] Isakov E, Keren O, Benjuya N. Trans-tibial amputee gait: time-distance parameters and EMG activity. Prosthet Orthot Int 2000;24:216–20.

[21] Beaman JJ, Barlow JW, Bourell DL, Crawford RH, Marcus HL, McAlea KP. Solid Freeform Fabrication: A New Direction in Manufacturing. Boston: Kluwer Academic Press; 1997.

[22] South BJ, Fey NP, Bosker G, Neptune RR. Manufacture of energy storage and return prosthetic feet using selective laser sintering. J Biomech Eng 2010;132: 015001 (1–6).

[23] Perotto A, Delagi EF. Anatomical guide for the electromyographer: the limbs and trunk. Springfield, IL, USA: Charles C. Thomas; 1994.

[24] Schwartz MH, Rozumalski A. A new method for estimating joint parameters from motion data. J Biomech 2005;38:107–16.

[25] Hernandez A, Dhaher Y, Thelen DG. In vivo measurement of dynamic rectus femoris function at postures representative of early swing phase. J Biomech 2008;41:137–44.

[26] Hernandez A, Lenz AL, Thelen DG. Electrical stimulation of the rectus femoris during pre-swing diminishes hip and knee flexion during the swing phase of normal gait. IEEE Trans Neural Syst Rehabil Eng 2010;18:523–30.

[27] Zmitrewicz RJ, Neptune RR, Sasaki K. Mechanical energetic contributions from individual muscles and elastic prosthetic feet during symmetric unilateral transtibial amputee walking: a theoretical study. J Biomech 2007;40:1824–31.

[28] Hansen AH, Meier MR, Sessoms PH, Childress DS. The effects of prosthetic foot roll-over shape arc length on the gait of trans-tibial prosthesis users. Prosthet Orthot Int 2006;30:286–99.

[29] Nolan L, Lees A. The functional demands on the intact limb during walking for active trans-femoral and trans-tibial amputees. Prosthet Orthot Int 2000;24:117–25.

[30] Silverman AK, Fey NP, Portillo A, Walden JG, Bosker G, Neptune RR. Compensatory mechanisms in below-knee amputee gait in response to increasing steady-state walking speeds. Gait Posture 2008;28:602–9.

[31] Hafner BJ, Sanders JE, Czerniecki J, Fergason J. Energy storage and return prostheses: does patient perception correlate with biomechanical analysis? Clin Biomech (Bristol Avon) 2002;17:325–44.

[32] Saunders MM, Schwentker EP, Kay DB, Bennett G, Jacobs CR, Verstraete MC, et al. Finite element analysis as a tool for parametric prosthetic foot design and evaluation. Technique development in the solid ankle cushioned heel (SACH) foot. Comput Methods Biomech Biomed Eng 2003;6:75–87.