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Differentiation between solid-ankle cushioned heel and energy storage and return prosthetic foot based on step-to-step transition cost

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Abstract—Decreased push-off power by the prosthetic foot and inadequate roll-over shape of the foot have been shown to increase the energy dissipated during the step-to-step transition in human walking. The aim of this study was to determine whether energy storage and return (ESAR) feet are able to reduce the mechanical energy dissipated during the step-to-step transition. Fifteen males with a unilateral lower-limb amputation walked with their prescribed ESAR foot (Vari-Flex, Ossur; Reykjavík, Iceland) and with a solid-ankle cushioned heel foot (SACH) (1D10, Ottobock; Duderstadt, Germany), while ground reaction forces and kinematics were recorded. The positive mechanical work on the center of mass performed by the trailing prosthetic limb was larger (33%, p = 0.01) and the negative work performed by the leading intact limb was lower (13%, p = 0.04) when walking with the ESAR foot compared with the SACH foot. The reduced step-to-step transition cost coincided with a higher mechanical push-off power generated by the ESAR foot and an extended forward progression of the center of pressure under the prosthetic ESAR foot. Results can explain the proposed improvement in walking economy with this kind of energy storing and return prosthetic foot.

Key words: amputation, ankle power, center of mass mechanics, ESAR prosthetic foot, gait, lower-limb prosthesis, mechanical energy, roll-over shape, SACH prosthetic foot, walking.

INTRODUCTION

Walking with a lower-limb prosthesis results in a higher metabolic energy cost than walking with two intact limbs [1]. With the introduction of the energy storage and return (ESAR) foot in the early 1980s, a passive-elastic prosthetic foot was marketed that was able to more closely mimic the human ankle by storing energy during stance and releasing this energy at push-off. It was assumed that this would reduce the metabolic energy cost while walking [2–3]. Whereas several studies have shown that prosthetic users subjectively choose the ESAR foot over the solid-ankle cushioned heel (SACH) foot [2], conflicting evidence is found with regard to its (clinically relevant) effect on metabolic energy cost [4–6]. Remarkably, however, the underlying effects of ESAR feet on the mechanics of walking have not yet been thoroughly investigated. Hence, it can be questioned whether

Abbreviations: COM = center of mass, COP = center of pressure, ESAR = energy storage and return, SACH = solid-ankle cushioned heel.

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the proposed mechanical effects of the ESAR foot are actually achieved.

In human walking, the amount of metabolic energy needed to walk has been shown to be related to the mechanical work associated with the step-to-step transition [7–10]. The double inverted pendulum model shows that during the step-to-step transition, negative mechanical work needs to be performed under the leading limb in order to redirect the body center of mass (COM) velocity from one circular arc to the next. In order to preserve walking speed, a similar amount of positive mechanical work needs to be performed during the gait cycle. The most efficient way to produce this positive work is by generating push-off work at the ankle at, or prior to, heel contact of the contralateral limb. This strategy minimizes the mechanical energy lost during collision and, therefore, the amount of mechanical work and metabolic energy required when walking [7–10]. Because of the absence of ankle musculature, people with amputation need to revert to other less efficient strategies to compensate for the reduced push-off power at the ankle [11–12]. These compensational strategies, primarily using hip muscle work during midstance, have been shown to result in a higher mechanical and metabolic energy cost when walking with a prosthesis [13].

In addition to differences in push-off work, mechanical work on the COM at collision has been shown to depend on the shape of the feet [9–10,14]. When instead of point feet the double inverted pendulum model is fitted with arc-shape feet, mimicking the human plantigrade foot, the center of pressure (COP) is able to move forward along the curved foot. This reduces the necessary directional change of the COM velocity, and concomitantly, the step-to-step transition cost [14–15]. In terms of metabolic energy cost, an optimal radius of the arc-shape of 0.3 times the limb length has been found [14,16]. Interestingly, recent research has indicated that the value most influencing the step-to-step transition might not be the radius but rather the length of the arc-shape foot [15]. Whereas a number of studies have used quasistatic mechanical loading to characterize the roll-over shape of different prosthetic feet [16–17], information about the roll-over shape in different prosthetic feet while walking is lacking.

The ESAR feet are specifically designed to improve push-off work by storing and releasing elastic energy; moreover, these feet can potentially restore the biological roll-over shape of the foot. The exact mechanism through which this would eventually lead to a potential lower metabolic energy cost has not been clearly investigated in the past. Recent insight into the mechanism of step-to-step transition cost and the mediating effect of foot shape indicates that ESAR feet are expected to reduce the mechanical energy lost during the step-to-step transition. Consequently, a reduction in mechanical energy lost could result in a lower metabolic energy cost when walking with the ESAR compared with a SACH foot. However, these hypotheses have not yet been confirmed. This study set out to investigate the mechanical effect of the ESAR foot and specifically to determine whether walking with a widely prescribed ESAR foot (Vari-Flex, Össur; Reykjavík, Iceland), indeed reduces the mechanical step-to-step transition cost compared with the SACH foot. Moreover, this study investigated the underlying mechanisms by looking at differences in prosthetic push-off work and roll-over characteristics while walking with the two prosthetic feet. We hypothesized that the ESAR foot can provide more push-off work and, together with more favorable roll-over shape characteristics, reduces the mechanical step-to-step transition cost during walking.

METHODS

Ethics Statement

After verbal and written clarification of the research procedure, written informed consent was obtained. This study was approved by the INAIL research board (Commissione Tecnico Scientifica; Budrio, Italy).

Subjects

Subjects who had walked with an ESAR prosthesis for at least the previous 2 yr and were able to ambulate without walking aids were included. In total, 15 male subjects with a unilateral transtibial amputation participated (age 55.8 ± 11.1 yr, weight 86.0 ± 12.6 kg, and height 1.74 ± 0.04 m). All subjects had undergone amputation because of trauma and were free of any musculoskeletal disorder or neurological disease that could affect the obtained results.

Data Acquisition

Subjects visited the prosthetic center twice. During the first visit, data were collected while wearing their prescribed ESAR foot and walking at a fixed walking speed of 1.2 m s^{-1}. To determine how much this fixed speed differed

from subjects' preferred walking speed, subjects' preferred walking speeds were obtained while walking overground. At the end of the first visit, subjects were fitted with the SACH foot (1D10, Ottobock; Duderstadt, Germany), which was aligned by an experienced prosthetist. Subjects returned to the laboratory after on average 25.7 h (range 21.0–28.4 h).

During both visits, subjects walked on a 10 m walkway while marker trajectories were tracked using a 10-camera VICON system at 100 Hz (VICON; Oxford, United Kingdom). Markers were placed on the anterior and posterior superior iliac spine, both the lateral and medial epicondyli and malleoli, and the calcaneus. For the prosthetic side, the malleoli markers were placed at a distal point at the rigid part of the prosthesis that approximated the sound limb malleoli position. Ground reaction forces were recorded at 1,000 Hz using two force plates (Kistler; Winterthur, Switzerland) embedded in the middle of the walkway. For each foot, a minimum of five trials were collected in which walking velocity (measured using photocells [MICROGATE RaceTime 2; Bolzano, Italy]) was within 0.05 m $s^{-1}$of the target speed, i.e., 1.2 m s^{-1}, and clean hits of the individual feet were recorded on the consecutive force plates.

Data Analysis

Ground reaction force data were low-pass filtered at 100 Hz using a fourth-order zero lag Butterworth digital filter. The three trials during which subjects showed the smallest change in walking speed at the start and end of the measurement were selected (i.e., those trials that subjects did not speed up or slow down). This was done by selecting the three trials that had the lowest total net mechanical work because net mechanical work ought to approximate zero when walking at constant speed. Basic spatiotemporal step parameters (i.e., stride and step lengths) were calculated using the ground reaction forces and the location of the calcaneus marker. The difference between the prosthetic and the intact step length is used as the measure of step length asymmetry.

Mechanical Work Performed on Center of Mass

The external mechanical power generated on the COM during the step-to-step transition was calculated as the dot product of the ground reaction force vector and the COM velocity vector for each limb independently [7]. The acceleration vector of the COM was calculated using the resultant of the ground reaction forces under both feet. COM velocity was then calculated by integrating the acceleration vector of the COM over time while assuming periodic strides [7]. The mechanical positive work performed under the trailing limb during push-off (W_{DS_trail}, J kg^{-1}) was calculated by integrating the trailing limb external mechanical power during the period from heel contact of the leading limb until toe-off of the trailing limb. The negative mechanical work performed under the leading limb (W_{DS_lead}, J kg^{-1}), was calculated as the integral of leading limb power between heel contact up until the instant the leading limb power became positive [9]. The leading limb power during the remaining time period was integrated to gain information about the net work performed during single stance (W_{SS}, J kg^{-1}). Calculated power profiles and work per walked trial were subsequently averaged for each subject and foot type.

Prosthetic Push-Off Work

Because the prosthetic foot-ankle segment cannot be modeled as a rigid body with a hinge joint, using inverse dynamics to calculate the mechanical power acting at the prosthetic ankle joint might introduce errors [18]. Therefore, a different approach was adopted in which the power at the ankle during stance was calculated by summing both the translational power and rotational power transferred from the foot (that is the deformable part of the prosthesis) to the shank (**Equation**) [19–20]:

$$P_{dist} = P_{translation} + P_{rotational} = \vec{F}_{dist} \bullet \vec{v}_{dist} + \vec{M}_{dist} \bullet \vec{\omega}_{shank},$$

where the subscript *dist* represents a distal point at the rigid part of the prosthetic leg at approximately the level of the malleoli at the contralateral side. $\vec{F}_{dist} \bullet \vec{v}_{dist}$ is the dot product of the ground reaction forces and linear velocities of the distal point. $\vec{M}_{dist}$ and $\vec{\omega}_{shank}$ represent the moment at the ankle and the angular velocity of the shank at the distal point, respectively. Push-off work (W_{push}, J kg^{-1}) was determined as the time integral of the positive power prior to toe-off. By integrating the remaining power profile during stance (e.g., excluding the positive work performed for push-off), the net amount of work that is either stored or dissipated prior to push-off was determined (W_{neg}, J kg^{-1}). Again, outcome parameters were separately analyzed for each of the three trials, after which trials were averaged to obtain a mean score for each subject and foot type.

Roll-Over Characteristics

For each subject and foot type, roll-over shapes were determined by transforming the COP data from a laboratory-based reference frame to the three-dimensional shank-based coordinate system [21]. Because the COP data progress forward from heel to toe during stance, while the shank is rotated over the foot, the roll-over shapes can be modeled as the lower half of a circle in the shank's plane of progression. The characteristics of this arc were obtained for each subject by fitting the equation that represents the lower half of a circle on the averaged data over the three trials performed for each foot. The data were fitted using a fitting algorithm with a nonholonomic constraint, ensuring that the obtained radius was larger or equal to the maximal vertical displacements of the roll-over shape. The roll-over shape represents the transformed COP data during the time period from heel contact to contralateral heel contact. However, close examination revealed that during the first rocker movement, roll-over shapes deviated strongly from circular. This problem was noted previously by Hansen et al. [16] and is also evident in the roll-over shapes presented by Miff et al. [22]. Therefore, in this study the first data points were excluded from the roll-over shape calculation using the second derivative of the Savitzky-Golay filtered data. This resulted in disregarding on average 20 and 22 percent of the data points and reduced the error of the circular fit by 49.1 and 50.1 percent for the ESAR and SACH foot, respectively. Adamczyk and Kuo showed in a modeling and experimental study in nondisabled persons that the length of the curved foot could have more of an effect on the amount of COM work than the shape of the curvature (i.e., the radius) [15]. Therefore, in addition to the radius and the total arc length of the roll-over shape, the forward travel of the COP on the ground (s) as a function of the angle between the shank and the vertical axis (α) was calculated [23].

Statistics

Data were tested for normality using a Kolmogorov-Smirnov test, and all parameters were normally distributed. Differences between fixed and preferred walking speed and between prosthetic feet were tested using a paired sample t-test. Significance was set a priori at $p = 0.05$.

RESULTS

All subjects were able to walk at the requested 1.2 m s^{-1} with both the ESAR and SACH foot. The averaged stride length and stride time did not differ between the two prosthetic feet (**Table 1**). Interestingly, the two prosthetic feet differed in step length asymmetry ($p < 0.001$). Subjects took smaller steps with the intact limb than with the prosthetic limb when walking with the SACH foot as compared with walking with the ESAR foot (**Table 1**).

Mechanical Work Performed on Center of Mass

Figure 1(a) shows that the external mechanical power generated by the trailing prosthetic limb on the COM during the push-off is larger and the negative power under the leading intact limb during collision is lower when push-off is performed by the ESAR prosthesis compared with the SACH prosthesis. This is confirmed by a 33 percent larger W_{DS_trail} of the trailing prosthetic limb ($p = 0.01$) and a 13 percent lower W_{DS_lead} under the leading intact limb ($p = 0.04$) while walking with the ESAR compared with the SACH foot (**Table 2**). Additionally, net work performed on the COM over the subsequent single stance period (W_{ss}) was lower with the ESAR foot. Interestingly, while no difference in push-off work (W_{DS_trail}) was found in the intact limb in either foot condition, more negative work (W_{DS_lead}) under the leading prosthetic limb was found when walking with the SACH foot than with the ESAR foot (16.7%, $p = 0.003$, **Figure 1(b)**).

Table 1.
Gait parameter (mean ± standard deviation).

Parameter	ESAR	SACH
Stride		
Speed (m s^{-1})	1.22 ± 0.02	1.22 ± 0.02
Stride Length (m)	1.38 ± 0.06	1.37 ± 0.07
Step		
Prosthetic Side Step Length (m)	0.70 ± 0.04	0.72 ± 0.04*
Intact Side Step Length (m)	0.68 ± 0.03	0.67 ± 0.04*
Asymmetry Step Length (m)†	0.01 ± 0.04	0.05 ± 0.04*

*Statistical difference between prosthetic feet ($p < 0.05$).
†Difference in step length calculated as prosthetic minus intact side step length.
ESAR = energy storage and return, SACH = solid-ankle cushioned heel.

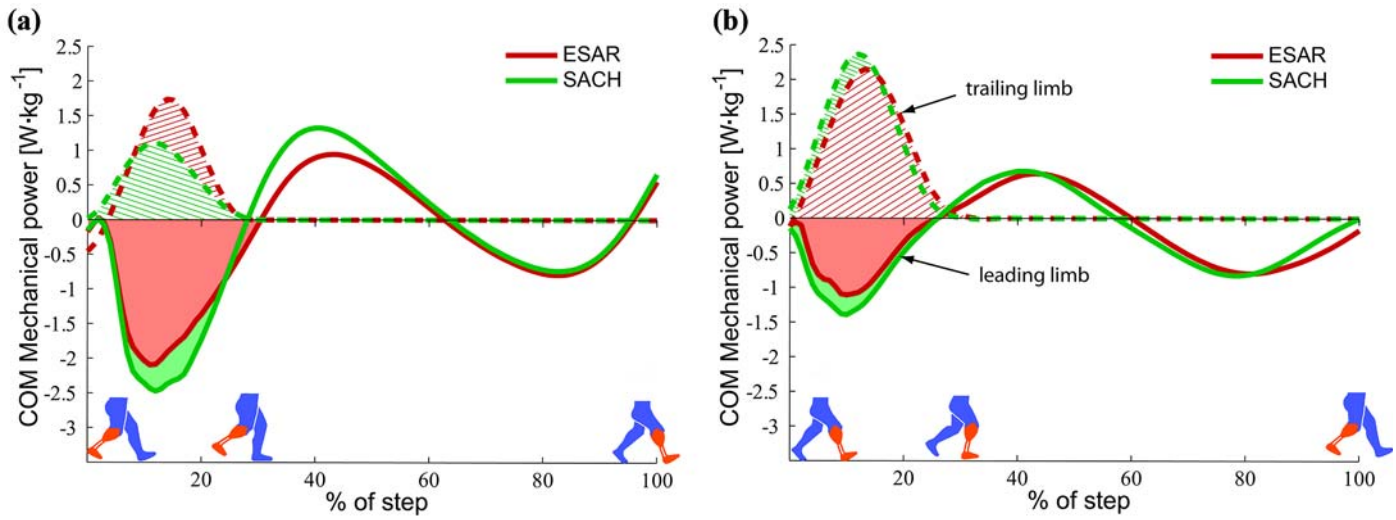

Figure 1.
Center of mass (COM) mechanical power profiles. COM mechanical power profiles of step during which prosthetic limb is **(a)** trailing limb and **(b)** leading limb. Dashed lines represent COM mechanical power under trailing limb, while solid lines represent mechanical power under leading limb. Hatched areas represent part over which push-off work of trailing limb was calculated, while solid areas represent amount of negative work during collision under leading limb. ESAR = energy storage and return foot, SACH = solid-ankle cushioned heel.

Prosthetic Push-Off Work

In **Figure 2**, the push-off power for the prosthetic and nonprosthetic limb is shown from heel contact until toe-off of that limb (i.e., stance period). When comparing **Figures 2(a)** and **2(b)**, it can be seen that substantially less positive work is generated when push-off was performed by the prosthetic side (**Figure 2(a)**) than on the intact side (**Figure 2(b)**). When comparing both prosthetic feet, larger positive work is generated by the ESAR foot (0.11 ± 0.03 J kg^{-1}) than by the SACH foot (0.05 ± 0.02 J kg^{-1}, $p < 0.001$, **Table 2**). As expected, the W_{neg} was larger in the ESAR foot (-0.29 ± 0.09 J kg^{-1}) than in the SACH foot (-0.19 ± 0.06 J kg^{-1}, $p < 0.001$, **Table 2**). No differences were found in push-off power of the intact limb (**Figure 2(b)**).

Roll-Over Characteristics

The COP data expressed in the shank-based coordinate system (thin line) and the fitted roll-over shape (thick line) are depicted in **Figure 3(a)**. The characteristics of this roll-over shape are summarized in **Table 2**. No difference in roll-over shape radius was found between both prosthetic feet ($p = 0.99$), while the total arc length of the ESAR foot was larger than that of the SACH foot ($p < 0.001$). **Figure 3(b)** clearly shows that the forward travel of the COP as a function of shank angle is not circular. As opposed to a steady increasing line reflecting a constant curvature, the line shows the typical S-shape pattern previously reported by Curtze et al. [17]. Large similarities in shape are seen between both prosthetic feet during the heel and ankle rocker. However, when walking with the SACH foot the line flattened at an earlier shank angle. Moreover, the SACH foot has a lower total COP forward displacement than the ESAR foot (**Figure 3(b)**).

DISCUSSION

This study showed that walking with the ESAR foot resulted in a reduced step-to-step mechanical cost compared with the SACH foot, as indicated by the reduced negative mechanical work under the leading limb during collision during push-off. The amount of mechanical work performed on the COM found in the current study is in close agreement with results from previous studies using similar prosthetic feet [13,24–27]. Even though more positive COM work can be generated with the ESAR foot than with the SACH foot, values are still substantially lower compared with positive COM work performed under the intact limb (36.8% lower) or values

Table 2.
Center of mass and ankle mechanical work and roll-over shape characteristics (mean ± standard deviation).

Parameter	ESAR	SACH
Center of Mass Mechanical Work		
Prosthetic Trailing		
W_{DS_lead} (J kg^{-1})	–0.20 ± 0.10	–0.23 ± 0.08*
W_{DS_trail} (J kg^{-1})	0.12 ± 0.06	0.09 ± 0.04*
W_{SS} (net, J kg^{-1})	0.03 ± 0.15	0.09 ± 0.08*
Intact Trailing		
W_{DS_lead} (J kg^{-1})	–0.10 ± 0.07	–0.12 ± 0.05*
W_{DS_trail} (J kg^{-1})	0.19 ± 0.06	0.20 ± 0.05
W_{SS} (net, J kg^{-1})	–0.04 ± 0.11	–0.03 ± 0.07
Prosthetic and Intact Limb Mechanical Ankle Joint Work		
Prosthetic Limb		
W_{neg} (J kg^{-1})	–0.29 ± 0.09	–0.19 ± 0.06*
W_{push} (J kg^{-1})	0.11 ± 0.03	0.05 ± 0.02*
Intact Limb		
W_{neg} (J kg^{-1})	–0.24 ± 0.07	–0.23 ± 0.07
W_{push} (J kg^{-1})	0.22 ± 0.06	0.24 ± 0.06
Roll-Over Characteristics		
Radius†	0.24 ± 0.04	0.24 ± 0.05
Total Arc Length†	0.20 ± 0.03	0.16 ± 0.02*

*Statistical difference between prosthetic feet ($p < 0.05$).
†Parameters are normalized to center of mass height.
ESAR = energy storage and return, SACH = solid-ankle cushioned heel, W_{DS_lead} = center of mass mechanical work performed under leading limb during double support, W_{DS_trail} = center of mass mechanical work performed under trailing limb during double support, W_{neg} = negative mechanical ankle work, W_{push} = ankle push-off mechanical work, W_{SS} = center of mass work during single stance.

found in nondisabled controls walking at a similar speed (53.9% lower) [28].

Prosthetic Push-Off Work

In normal walking, the work generated at the ankle is the major contributor to the total COM push-off work under the trailing limb during step-to-step transition. This factor can indeed explain part of the differences found in step-to-step transition cost between prosthetic feet. Prosthetic push-off work was 120 percent higher with the ESAR than with the SACH prosthesis; however, it was still 50 percent less than that generated by the contralateral intact ankle. These values are in line with previous studies [29]. Because the ESAR prosthesis is a passive device, the amount of push-off work that can be generated is related to the amount of elastic-strain energy that can be stored during the preceding stance period. Congruous with literature [30], negative work performed by the prosthetic limb was 55.5 percent higher during stance in the ESAR foot than in the SACH foot (**Table 2**). The period during which this energy was stored was the same in both prostheses and predominantly occurred in mid to late stance (**Figure 2**). To summarize, differences in step-to-step transition cost can be partly explained by the ability of the ESAR feet to store power during stance and return this power during push-off, thereby reducing the collision loss.

Roll-Over Characteristics

The ability of the human ankle and foot to move in a controlled fashion while the COP progresses forward under the foot greatly influences the directional change of the COM, which are necessary during step-to-step transition and thereby directly influences the transition cost [14–15]. Contrary to our hypothesis, roll-over shape radii found in the current study did not differ between the two prosthetic feet. These results seem to contradict previous findings by Curtze et al., who stated that when feet are tested using quasistatic mechanical loading, markedly different radii are found between both prosthetic feet [17]. However, inherent mechanical properties of the prosthetic feet might be subdued because of alignment alterations made by the prosthetist when fitting the prosthesis [31]. Moreover, the strikingly large similarities found in the roll-over shape curve between both feet (**Figure 3(a)**) might indicate that differences in prosthetic roll-over shape characteristics, inherent to the prosthetic feet, are attenuated when walking with the prosthesis. More importantly, the estimation of the roll-over shape characteristics is based on the assumption that the shank-based COP trajectory can be represented by a lower half of an arc. However, the radius of both the biological and also a prosthetic foot-ankle complex does not represent a perfect circular arc [23,32].

Nevertheless, as mentioned previously, the total amount of forward progression of the COP under the foot (i.e., the functional foot length) in relation to the shank angle might be a more important parameter than the foot radius [15]. The resulting typical S-pattern that is then seen (**Figure 3(b)**) may be understood using the three rocker phases as described by Perry [33].The initial relative flat period can be attributed to the heel rocker, during which the tibia progresses forward while pivoting on the heel. The second period is steep and represents the ankle rocker, during which the COP is moved along the foot while the tibia progresses further forward. The final flat period represents the forefoot rocker, during which the

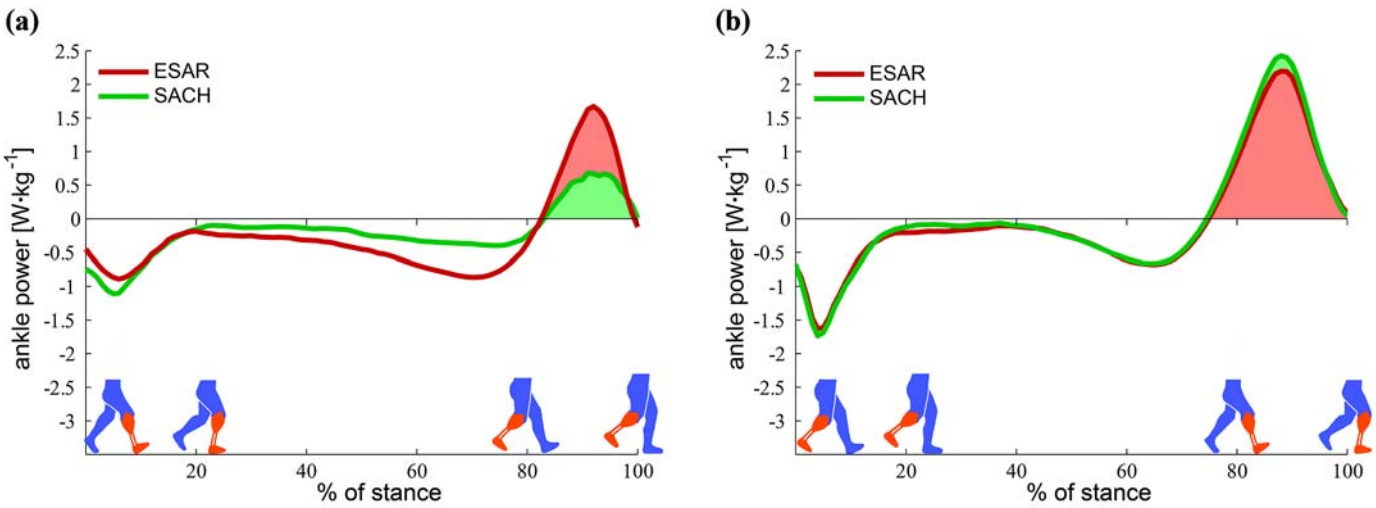

Figure 2.
Push-off power over stance period. Push-off power transferred from foot to shank of **(a)** prosthetic limb and **(b)** intact limb. Filled areas represent part over which push-off work was calculated. ESAR = energy storage and return, SACH = solid-ankle cushioned heel.

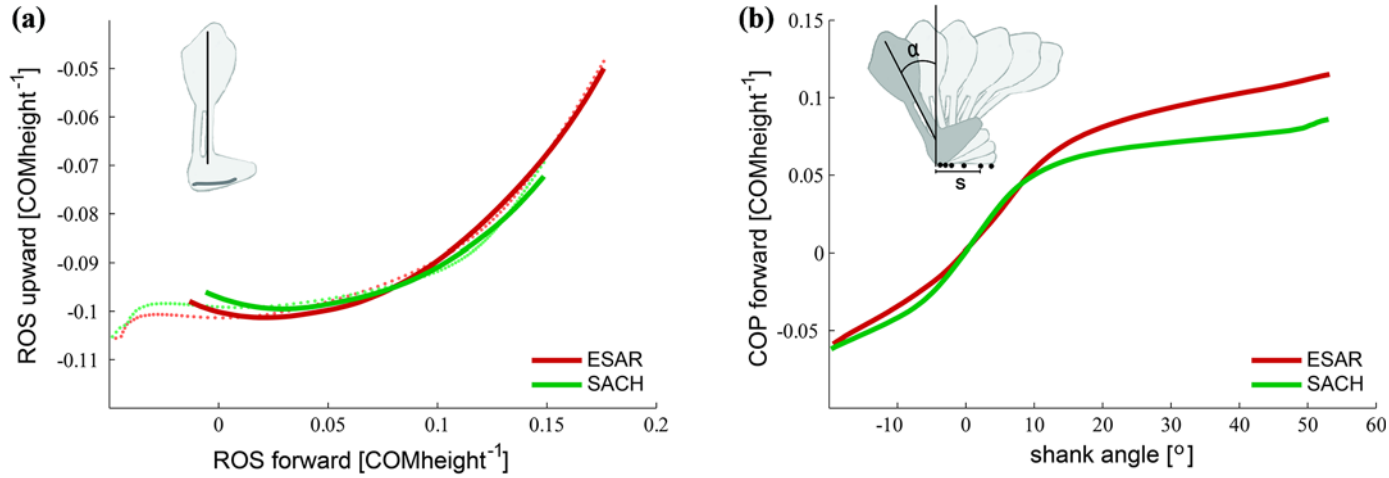

Figure 3.
Roll-over shape (ROS) and forward progression of center of pressure under foot. **(a)** ROS in shank-based coordinates (see inset). Thin (lighter) lines represent center of pressure data from heel contact till opposite heel contact in shank-based coordinates. Thick (darker) lines are fitted circular arcs through data points, excluding initial part during which shank-based center of pressure data deviated from circular. **(b)** Forward travel of center of pressure (COP) (*s*) over ground depicted as function of shank angle (α) (see inset). All parameters (excluding shank angle) are normalized to height of averaged pelvic markers to ground during static stance trial. COM = center of mass, ESAR = energy storage and return, SACH = solid-ankle cushioned heel.

heel is lifted from the ground while pivoting on the forefoot. This last period is initiated at an earlier shank angle when walking with the SACH foot, indicating an earlier onset of forefoot rocker when walking with the SACH foot. Moreover (or possibly as a consequence), the SACH foot has a lower total COP forward displacement than the ESAR foot (**Figure 3(b)**). The shorter COP forward progression (the smaller functional foot length) found

when walking with the SACH foot and the earlier onset of the forefoot rocker could have contributed to the larger collision loss found in this foot as it increases the necessary directional change of the COM at a given step length [15,34].

Step Length

A strategy that subjects can adopt to reduce the step-to-step transition cost is reducing the length of the step taken. Previous experimental results showed that step-to-step transition cost and metabolic energy cost will increase in proportion to the fourth power of step length [35]. Hence, possible changes in step length between conditions ought to be checked in order to allow for proper interpretation of the results. In the current study, no difference in stride length was found between prosthetic feet, and as such stride length did not influence the difference in step-to-step transition cost between feet. Interestingly, however, subjects had a larger step length asymmetry with the SACH foot than with the ESAR foot. More specifically, subjects took shorter steps when push-off was performed with the SACH prosthesis (i.e., shorter intact limb step length) than when performed by their intact limb, thereby potentially mitigating the increased step-to-step transition cost with the SACH foot. Conversely, subjects took a relatively larger consecutive prosthetic step (intact push-off). This increase in prosthetic step length with the SACH prosthesis could explain the higher collision loss (16.7%) found in that step (**Figure 1(b)**). The cause for the relatively smaller intact step when push-off is performed with the SACH prosthesis could be related to the reduced push-off power that is generated with the SACH foot. Additional explanation can be sought in the inherent mechanical properties of the SACH prosthesis limiting long steps. The rigid ankle of the SACH foot restrains dorsal flexion during late stance [36], resulting in an early heel rise and concomitantly shorter steps [2]. Moreover, the shorter keel found in the SACH foot compared with the ESAR foot results in a highly flexible forefoot section and contributes to an earlier onset of forefoot rocker. Close examination of **Figure 3(b)** affirms the notion that with the SACH foot the forefoot rocker is initiated earlier (earlier flattening of the line).

Optimizing Prosthetic Feet

Current results show that the ESAR foot is able to reduce the step-to-step transition cost. The verification of this mechanical effect of the ESAR foot implies that these feet could potentially reduce metabolic energy cost during walking. However, in order to maximize such a reduction in metabolic cost, the energy storage and release of the prosthetic foot ought to be both of the right magnitude and at the right instant during gait. For example, previous studies have shown that excessive push-off work can lead to compensational joint work requiring metabolic energy to ensure stability [24,27]. Additionally, the period in gait during which energy is stored [24] and the timing of energy release [8,10] are factors that, if not optimal, can attenuate any positive effects of an increased push-off work on metabolic energy cost. Apart from optimizing energy storage and release, functional keel length of the foot ought to be optimized in order to ensure that the redirection of the COM is reduced while simultaneously controlling for the metabolic penalty imposed when feet are too long [15]. In addition to these factors, forward dynamic modeling studies have demonstrated that alterations in prosthetic foot stiffness can greatly affect gait mechanics and result in compensatory muscle activations [37].

The absence in the literature of a clear metabolic benefit when walking with the ESAR feet stresses the fact that the efficacy of a prosthesis to lower the overall metabolic energy cost is not merely related to the ability to reduce the step-to-step transition cost. While the current results clearly indicate that ESAR feet can indeed reduce mechanical work for the step-to-step transition, and therefore has the potential to reduce metabolic cost, it is important to further elucidate which factors affect the metabolic energy cost and possibly attenuate the found positive effect of the ESAR feet on metabolic cost of walking.

Limitations

One of the limitations of this study was the relatively short accommodation period of 1 d. This could have amplified differences in the gait pattern between the prescribed ESAR and the SACH prosthesis. Unfortunately, longitudinal studies into adaptation time with a novel prosthesis are scarce. With an adaptation time of 1 d, we did find the anticipated differences, though it is unclear to what extent these differences might change after more adaptation time. It may be noted that the subjects in our study were all in some degree familiar with the SACH prosthesis as either their bath prosthesis or as their first prosthesis after amputation. Another limitation of this study was the fact that we used one specific ESAR (Vari-Flex)

prosthetic foot. Previous research has shown that differences in prosthetic design can influence mechanics and muscle activity [37]; consequently, results with this particular prosthetic foot cannot be generalized to all ESAR feet that are commercially available. Because outcome variables are greatly influenced by walking speed, we had subjects walk at a set walking speed of 1.20 m s^{-1}. Although subjects' preferred speed (1.27 m s^{-1}) differed significantly ($p = 0.03$) from the 1.2 m s^{-1} enforced during the measurement, differences are small and it is unknown whether these differences are clinically irrelevant.

CONCLUSIONS

This study examined the underlying mechanical advantages of a contemporary ESAR foot and showed that this ESAR foot resulted in a lower mechanical step-to-step transition cost compared with walking with the SACH prosthesis. Close examination of possible explaining factors showed that this difference was explained by the higher amount of positive work performed by the ESAR foot during push-off and the larger forward travel of the COP under the prosthetic ESAR foot compared with the SACH foot. Results confirm the mechanical advantage, and potential metabolic advantage, of ESAR feet. Moreover, these results provide insight in how underlying properties such as the push-off power generating capacity and the roll-over characteristics of the foot might influence possible mechanical advantages. The lack of convincing evidence in the literature supporting a clinically significant reduction in metabolic energy cost while walking with an ESAR foot suggests that other factors outside those related to step-to-step transition cost might attenuate the potential benefits of the ESAR prosthetic foot. It remains a formidable challenge to disentangle and optimize these potential influencing factors while at the same time maintaining the observed positive mechanical characteristics of ESAR feet.

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