Optimizing Prosthetic Gait

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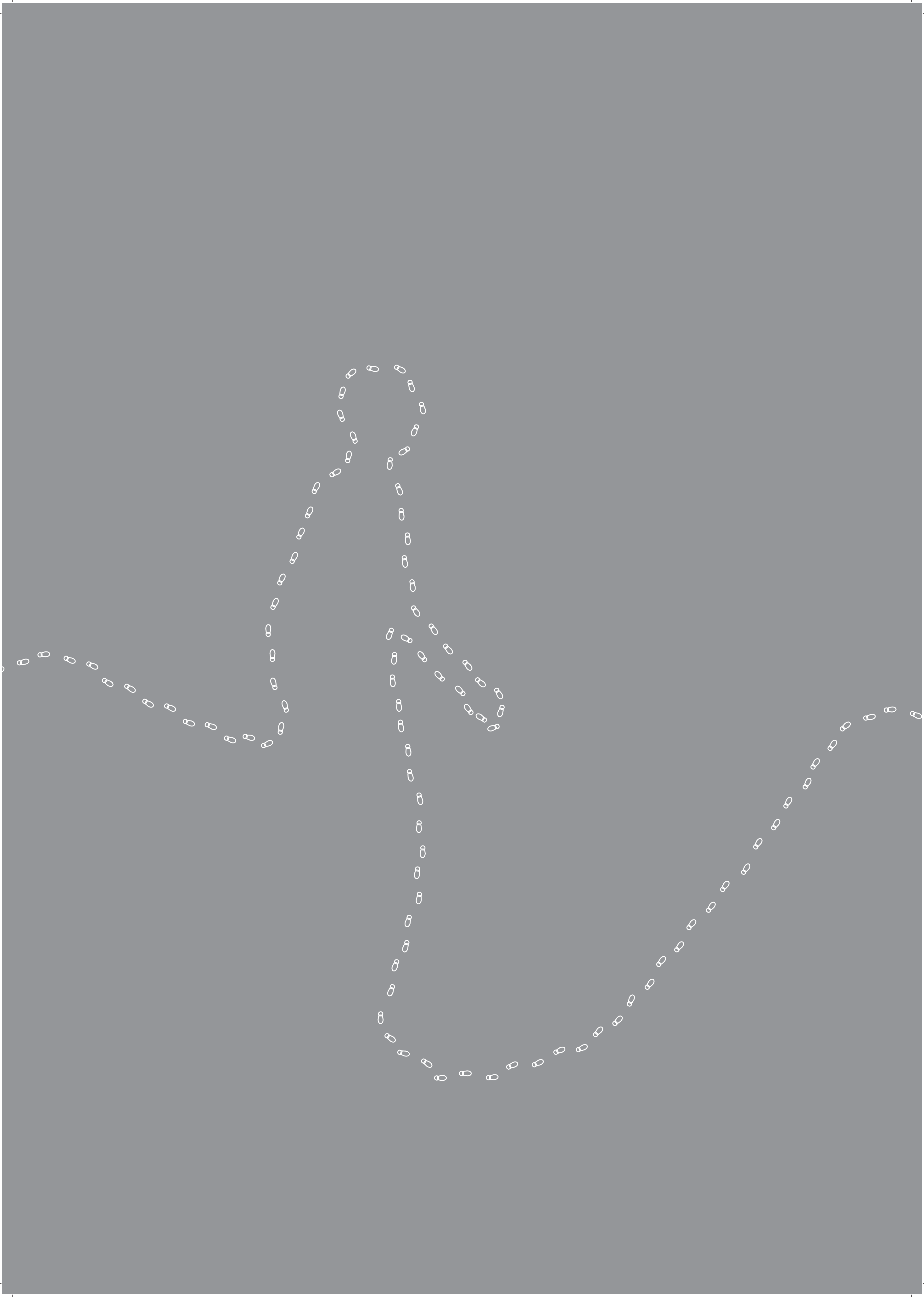
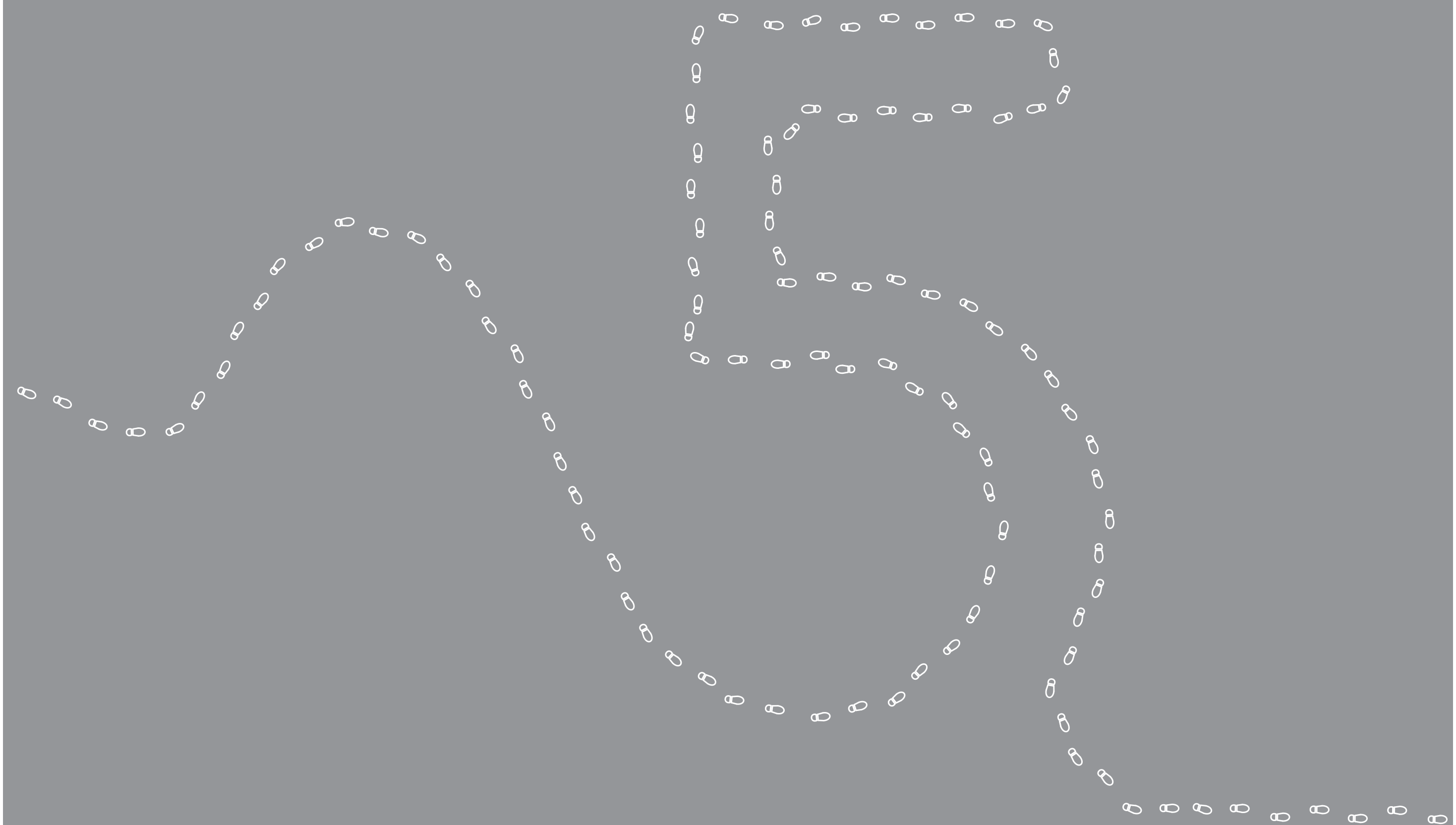


Differentiation between prosthetic feet based on in vivo center of mass mechanics



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Abstract

A decreased push-off power at the ankle, as seen in people walking with a prosthesis, will result in an increase in the energy dissipated during the step-to-step transition. Energy storage and return prosthetic feet have been developed to improve this push-off power, and therefore, are thought to reduce the step-to-step transition cost and concomitantly the metabolic energy required to walk. Interestingly however, no convincing evidence is found that support the notion that the metabolic energy cost is reduced compared to more conventional feet. This entails that either these feet are unable to reduce the step-to-step transition cost, or other disadvantageous factors attenuate the positive metabolic effect of the reduced step-to-step transition cost. The aim of this study was to determine whether walking with the energy storage and return foot reduce the step-to-step transition cost. Fifteen persons with a unilateral lower limb amputation walked with their prescribed energy storage and return foot and with a conventional solid ankle cushioned heel foot, while ground reaction forces and kinematics were recorded. The mechanical push-off power under the trailing limb was larger (33%) and the negative power under the leading limb was lower (13%) when push-off was performed by the energy storage foot compared to the conventional foot. The improved push-off power under the trailing prosthetic limb can be explained by the increased push-off power that was generated by the energy storage and return foot. Moreover, the longer forward progression of the center of pressure under the prosthetic foot augments the reduced step-to-step transition cost when walking with the ESAR prosthesis. Walking with the energy storage and return foot reduces the step-to-step transition cost. Hence, the limited effects on metabolic energy cost when walking with the ESAR feet should be attributed to metabolic costly side effects of these prosthetic feet.

Introduction

Walking with a lower limb prosthesis results in a higher metabolic energy cost than walking with two unimpaired legs [226]. 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, which was assumed to reduce the metabolic energy cost while walking [88-89, 244]. Whereas several studies have shown that prosthetic users, subjectively choose the ESAR foot over the conventional soft ankle cushioned heel (SACH) foot [88], conflicting evidence is found with regard to its effect on metabolic energy cost [104, 212, 219]. Hence, it can be questioned whether the proposed mechanical effect of the ESAR foot is actually achieved.

In human walking the amount of metabolic energy needed to walk is to a large extent related to the mechanical work associated with the step-to-step transition [56, 129, 131, 181]. The double inverted pendulum model shows that 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. 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 contra-lateral limb. This will reduce the mechanical energy lost during collision, and therefore, the amount of mechanical work and metabolic energy required [56, 129, 131, 181]. Due to the absence of ankle musculature, people who walk with a lower limb prosthesis have a reduced push-off work at the prosthetic side and need to revert to other, less efficient, strategies to generate the required positive mechanical work during the step-to-step transition, resulting in an increase in metabolic energy cost [6, 110, 152].

The ESAR feet are specifically designed to improve push-off work, and thus are expected to reduce the mechanical energy lost during collision, and as such are expected to require less metabolic energy cost. Hence, the lack of evidence supporting a reduction in metabolic energy cost when walking with the ESAR foot is remarkable. Two possible causes for the limited effectiveness of the ESAR foot to reduce the metabolic energy cost can be postulated: 1) despite the fact that the ESAR foot is able to store and return energy it does not adequately reduce step-to-step transition cost, or 2) other disadvantageous factors (e.g. a compromised walking stability) are attenuating the positive effect of the reduced step-to-step transition cost.

In addition to push-off work, mechanical work at collision has been shown to depend on the roll-over shape of the feet [1, 131, 181]. When, instead of point feet, the double inverted pendulum model is mounted with arc-shape feet, mimicking the human plantigrade foot, the center of pressure is able to move forward along the curved foot reducing the necessary directional change of the COM velocity, and concomitantly, the step-to-step transition cost [1]. Theoretically, step-to-step transition cost decreases quadratically with increasing roll-over radius, although, in terms of metabolic energy cost an optimal radius of 0.3 times the leg length is found [1, 95]. This corresponds to the values found in able-bodied subjects [94, 144] and people walking with a lower limb prosthesis [33]. Possible differences in shape can either diminish or augment the potential positive effect of an increased push-off work. Therefore, it is imperative to gain information about the roll-over shape characteristics when walking with the prosthesis in order to interpret the mechanical step-to-step transition cost. Whereas a number of studies have used quasi-static mechanical loading to characterize the roll-over shape of different prosthetic feet [32, 91, 93], information from in vivo studies in people walking with different prostheses is lacking.

To date, no studies have been published that combine in vivo measurements of center of mass mechanics with both prosthetic ankle power and roll-over characteristics of the prosthetic foot. To add to this limited body of knowledge, this study set out to firstly, determine whether walking with the widely prescribed ESAR foot reduces the step-to-step transition cost compared to the conventional SACH foot, and secondly, to relate the possible differences in transition cost to differences in the push-off work generated by the prosthetic foot and ankle and the roll-over characteristics of the two prosthetic ankles. We hypothesize that the ESAR foot can provide more push-off work, and together with a more favorable roll-over shape radius, can reduce the step-to-step transition cost during walking. If true, this would corroborate the notion that although the ESAR foot is able to reduce step-to-step transition cost, other disadvantageous factors are responsible for the limited effect on metabolic energy cost found in the literature.

Methods

Subjects

Patients were included who had walked with an ESAR prosthesis (Vari-Flex, Össur, Iceland) for at least the previous two years and were able to ambulate without walking aids. In total, 15 male subjects with a unilateral amputation at the level of the shank participated (age 55.8 years [SD = 11.1], weight 86.0 kg [SD = 12.6] and height 1.74 m [SD = 0.04]). All subjects had undergone amputation due to trauma and were free of any musculoskeletal disorder or neurological disease that could affect obtained results. After verbal and written clarification of the research procedure, informed consent was obtained.

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 $m \cdot s^{-1}$. At the end of the first visit subjects were mounted with the SACH foot (1D10, Ottobock, Germany) which was aligned by an experienced prosthetist. Subjects returned to the laboratory after on average 25.7 hours (range 21-28.4).

During both visits subjects walked on a 10 m walkway while segment trajectories were tracked using a 10-camera VICON System at 100Hz (VICON, Oxford, United Kingdom). Markers were placed on the anterior and posterior superior iliac spine, both the lateral and medial epicondyli and malleoli, and on the calcaneus. For the prosthetic side the malleoli markers were placed at a distal point at the rigid part of the prosthesis which approximated the sound limb malleoli position. Reapplication of the markers for the second measurement session was supported by images of the marker location during the first measurement session. Ground reaction forces were recorded at a 1000 Hz using two force plates (Kistler, Winterthur, Switserland) 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 $m \cdot s^{-1}$ of the target speed, i.e. 1.2 $m \cdot s^{-1}$, and clean hits of the individual feet were recorded on the consecutive force plates.

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Data analysis

Ground reaction force data were low-pass filtered at 100 Hz using a fourth-order zero lag Butterworth digital filter. If walking speed is kept equal over a step, total mechanical work ought to approximate zero. To ensure that walking speed remained constant over the step only the three trials with the lowest total net mechanical work were used. Basic spatio-temporal step parameters were calculated using a combination of the ground reaction forces and the location of calcaneus marker. The difference between the prosthetic minus the intact step length is used as the measure of step length asymmetry.

Mechanical work performed on the COM

The anterior and posterior superior iliac spine markers were used to estimate the position of the COM. The external mechanical power generated on the center of mass 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 [56]. 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 [56]. The mechanical positive work performed under the trailing limb during push-off (W_{DS_trail}, $J \cdot kg^{-1}$) was calculated by integrating the trailing limb power during dual support. The negative mechanical work performed under the leading leg (W_{DS_lead}, $J \cdot kg^{-1}$), was calculated as the integral of leading limb power between heel contact up until the instant the leading limb power became positive [131]. The leading limb power during the remaining time period was integrated to gain information about the net work performed during single stance (W_{ss}, $J \cdot kg^{-1}$). Calculated power profiles and work per walked trial were subsequently averaged for each subject and foot type.

Ankle work

Because the prosthetic foot-ankle segment cannot be modeled as a rigid body with a hinge joint, using conventional inverse dynamics to calculate the mechanical power acting at the prosthetic ankle joint might introduce errors [186]. Therefore a different approach was adopted in which the power at the ankle was calculated by summing both the translational power and rotational power transferred from the foot to the shank [175]:

$$P_{ankle} = P_{translation} + P_{rotational} = \vec{F}_{ankle} \times \vec{v}_{ankle} + \overline{M}_{shank} \times \overline{\omega}_{shank} \qquad (1)$$

where $\vec{F}_{ankle} \times \vec{v}_{ankle}$ is the dot product of the ground reaction forces and linear velocities of the ankle. $\overline{M}_{shank} \times \overline{\omega}_{shank}$ and represent the moment at the ankle and the angular velocity of the shank at the distal point, respectively. Push-off work (W_{push}, $J \cdot kg^{-1}$) was determined as the time integral of the positive power found prior to toe-off. By integrating the remaining power profile (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}, $J \cdot kg^{-1}$). Again, outcome parameters were separately analyzed for each of the three trials, after which the trails were subsequently 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 center of pressure (COP) data from a laboratory-based reference frame to the three dimensional shank-based coordinate system [67, 223]. Because the COP data progresses forward from heel to toe during stance while the shank is rotated over the foot, the roll-over shapes can be modeled as a lower half of a circle in the shanks 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 non-holonomic constraint, ensuring that obtained radius was larger or equal to the maximal vertical displacements of the roll-over shape. As suggested by Hansen and colleagues (2004) [94], roll-over shape represents the COP during the time period from heel contact to contra-lateral heel contact. However, close examination revealed that during the first rocker movement roll-over shapes deviated strongly from circular. This problem has been noted previously by Hansen and colleagues (2004) [91] and is also evident in the roll-over shapes presented by Miff and colleagues (2008) [149]. This downward shift (as can be observed in Figure 3) is presumably caused by the compression of the heel at initial contact. Moreover forces are still low during this phase as the trailing limb is still in contact with the ground resulting in noisy forces and incorrect COP localization [91]. Therefore, in this study the first rocker movement was 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 improved the error of the circular fit (i.e. reduced it) with 49.1% and 50.1% for the ESAR and SACH foot, respectively. Logically, excluding part of the shank-based COP data results in a reduced total arc length of the roll-over shapes, but because in both the ESAR and SACH trials the same amount of data were excluded this did not alter conclusions. Reliable estimation of roll-over shape characteristics is based on the assumption

Table 1. Gait parameters (mean ± SD).

	ESAR	SACH
Stride parameters		
Speed	1.22(0.02)	1.22(0.02)
Stride length	1.38(0.06)	1.37(0.07)
Stride time	1.13(0.06)	1.12(0.05)
Step parameters		
Prosthetic side		
Step length	0.70(0.04)	0.72(0.04)*
Step time	0.57(0.03)	0.56(0.02)
Intact side		
Step length	0.68(0.03)	0.67(0.04)*
Step time	0.56(0.03)	0.56(0.03)
Asymmetry		
Step length†	0.01(0.04)	0.05(0.04)*

Abbreviations: ESAR: energy storage and return foot (Vari-Flex, Össur, Iceland); SACH: soft ankle cushioned heel foot (1D10, Ottobock, Germany).
* denotes statistical difference between prosthetic feet ($p < .05$).
† difference in step length calculated as the prosthetic minus the intact side step length.

that the shank-based COP can be represented by a lower half of an arc. However, the radius of both the biological, but also a prosthetic foot-ankle complex might not represent a perfect circular arc [33, 92]. Consequently, important insight into the roll-over characteristic of the feet as the shank progresses forward might be lost. Therefore, we also calculated the forward travel of the COP on the ground (s) as a function of the angle between the shank and the vertical (α) (Figure 3B) [33].

Statistics

Data were tested for normality using a Kolmogorov-Smirnov test and all parameters were normally distributed. Differences between prosthetic feet were tested using a paired sample *t* test. Significance was set a priori at the 5% level.

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Results

All subjects were able to walk at the requested 1.2 $m \cdot s^{-1}$ 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

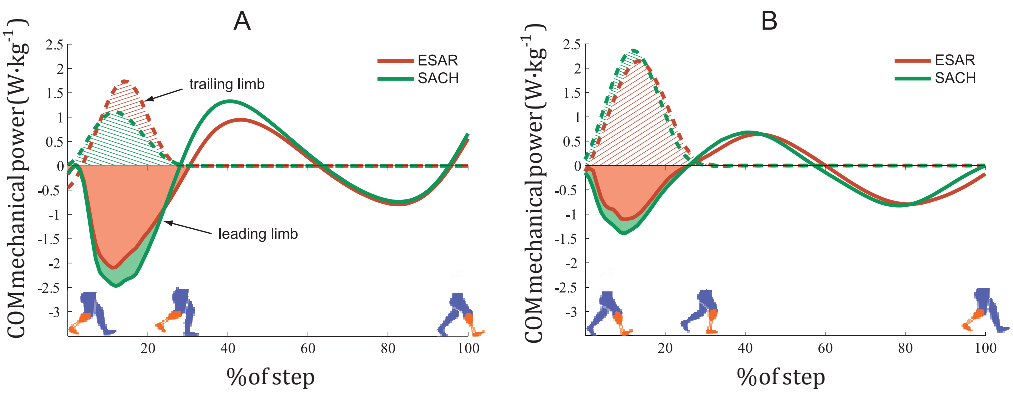

Figure 1. Center of mass mechanical power profiles.

The center of mass mechanical power profiles of the step during which the prosthetic limb is the trailing limb **(A)** and the leading limb **(B)**. Dashed lines represent center of mass mechanical power under the trailing limb, while solid lines represent the mechanical power under the leading limb. Hatched dashed and solid areas represent the part over which push-off work and collision loss was calculated, respectively.
Abbreviations: ESAR: energy storage and return foot (Vari-Flex, Össur, Iceland); SACH: soft ankle cushioned heel foot (1D10, Ottobock, Germany).

length asymmetry ($p < .001$). Subjects took smaller steps with the intact limb compared to the prosthetic limb when walking with the SACH foot as compared to walking the ESAR foot (Table 1).

Mechanical work performed on the COM

Figure 1A shows that the external mechanical power performed by the trailing leg on the COM is larger, and the negative power under the leading limb during collision is lower when push-off is performed by the ESAR prosthesis compared to the SACH prosthesis. This is confirmed by a 33% larger W_{DS_trail} of the trailing prosthetic foot ($p = .01$) and a 13% lower W_{DS_lead} under the leading intact limb ($p = .04$) while walking with the ESAR compared to 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 was found in the intact limb in either foot condition, more negative work under the leading prosthetic limb was found when walking with the SACH foot compared to the ESAR foot (16.7%, $p = .003$; Figure 1B).

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Ankle work

In Figure 2 the ankle power for the prosthetic and non-prosthetic limb is shown from heel contact until toe-off of that limb (i.e. stance period). It can be seen that when push-off was performed with the prosthetic limb (Figure 2A), substantial less positive work is generated at the ankle compared to the intact limb. The largest difference between intact and prosthetic push-off ankle power is seen when walking with the SACH foot. As expected, the W_{neg} was larger in the ESAR foot ($-0.29 \pm 0.09\ J \cdot kg^{-1}$) compared to the SACH foot ($-0.19 \pm 0.06\ J \cdot kg^{-1}$, $p < .001$; Table 2). No differences are found in push-off power of the intact limb (Figure 2B).

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 3A. The characteristics of this roll-over shape are summed in Table 2. No difference in roll-over shape radius was found between both prosthetic feet ($p = 0.992$), while the total arc length of the ESAR foot was larger compared to that of the SACH foot ($p < .001$). Figure 3B 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 and colleagues (2009) [32]. The different phases can be understood using the three rocker phases as described by Perry and colleagues (1992) [168]. The

Table 2. Center of mass, ankle work, and roll-over shape characteristics (mean ± SD).

	ESAR	SACH
Center of mass mechanical work		
Prosthetic trailing 		
W_{DS_lead} (J·kg^{-1})	-.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)
Ankle mechanical 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)*

Abbreviations: ESAR: energy storage and return foot (Vari-Flex, Össur, Iceland); SACH: soft ankle cushioned heel foot (1D10, Ottobock,Germany); W_{DS_lead}: center of mass mechanical work performed under the leading limb during dual support; W_{DS_trail}: center of mass mechanical work performed under the trailing limb during dual support; W_{SS}: center of mass work during single stance; W_{neg}: negative mechanical ankle work; W_{push}: ankle push-off mechanical work.
* denotes statistical difference between prosthetic feet ($p < .05$).
† parameters are normalized to center of mass height.

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 heel is lifted from the ground while pivoting on the forefoot. 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, indicating an earlier onset of forefoot rocker when walking with the SACH foot. As a result, a lower total COP forward displacement in the SACH foot is seen compared to the ESAR foot (Figure 3B).

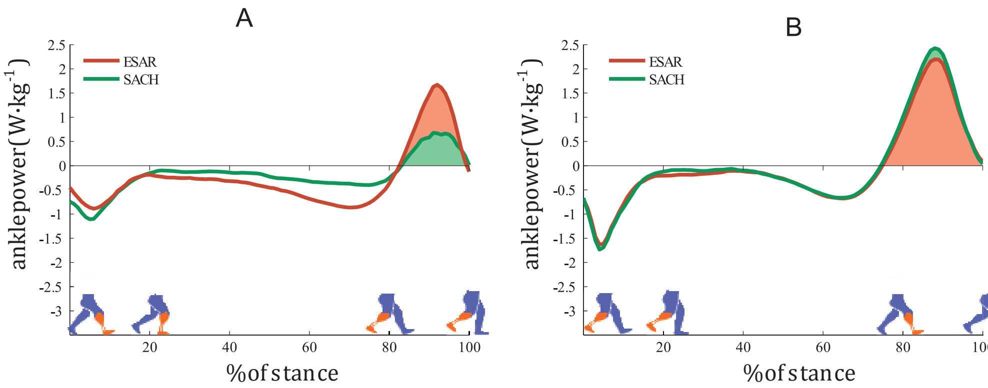

Figure 2. Push-off power over the stance period.

Push-off power transferred from the foot to the shank of the prosthetic limb **(A)** and the intact limb **(B)**. The filled areas represent the part over which ankle push-off work was calculated.
Abbreviations: ESAR: energy storage and return foot (Vari-Flex, Össur, Iceland); SACH: soft ankle cushioned heel foot (1D10, Ottobock, Germany).

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Discussion

This study showed that walking with the energy storage and return (ESAR) foot resulted in a reduced step-to-step mechanical cost compared to a conventional solid ankle cushioned heel (SACH) foot. 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 [98, 110, 152, 192, 241]. Although more positive COM work can be generated with the ESAR foot compared to the SACH foot, values are still substantially lower compared to positive COM work performed under the intact limb (36.8% lower) or values found in able-bodied controls walking at a similar speed (53.9% lower) [52].

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 transition cost between prosthetic feet. Ankle push-off work was 120% higher with the ESAR compared to the SACH prosthesis, however, it was still 50.0% less than that generated by the contra-lateral intact ankle. These values are in line with previous studies [183]. 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

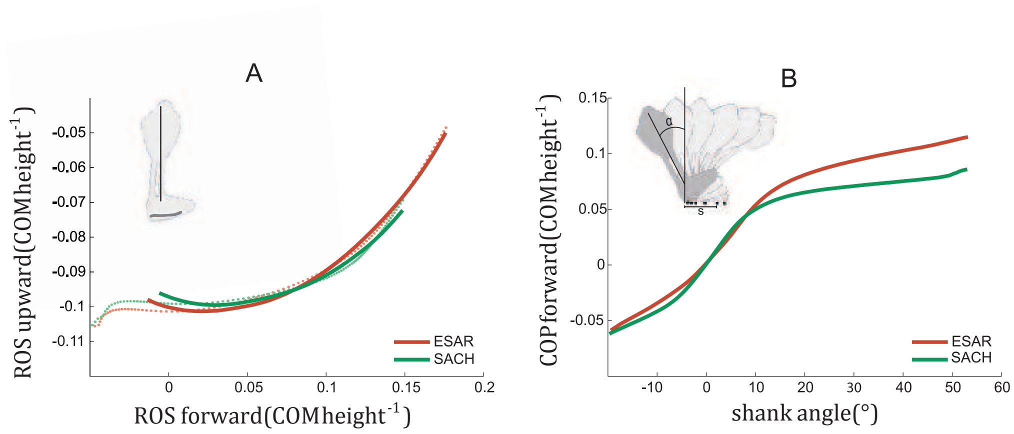

Figure 3. Roll-over shape and forward progression of the center of pressure under the foot.

(A) Roll-over shape in shank based coordinates (see inset). The thin (lighter) lines represent the center of pressure data from heel contact till opposite heel contact in shank based coordinates. The thick (darker) lines are the fitted circular arcs through the data points, excluding the initial part during which shank-based center of pressure data deviated from circular. **(B)** The forward travel of the center of pressure (s) over the ground depicted as a function of shank angle (α). All parameters (excluding shank angle) are normalized to center of mass height.
Abbreviations: ESAR: energy storage and return foot (Vari-Flex, Össur, Iceland); SACH: soft ankle cushioned heel foot (1D10, Ottobock, Germany); COM: center of mass.

that can be stored during the preceding stance period. Congruous with literature [84], negative work performed by the prosthetic foot and ankle was 55.5% higher during stance in the ESAR foot compared to the SACH foot (Table 2). Part of this higher negative work is stored to allow for the increased push-off work. 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 sum, differences in step-to-step transition cost can be partly explained by the ability of the ESAR feet to store during stance and return this power during push-off, thereby, reducing the collision loss.

Step length

In addition to improved ankle push-off work, step-to-step transition cost is dependent on the step length; taking shorter steps will require less directional change of the COM, and as such, reduce the collision loss. 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 [55]. In the current study, no difference in stride length was found between prosthetic feet. However, subjects demonstrated a larger step length asymmetry with the SACH foot compared to the ESAR foot. More specifically, subjects took relatively shorter steps when push-off was performed with the SACH prosthesis (i.e. shorter intact limb step length), 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 1B). 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 midstance [174, 222] resulting in an early heel rise and concomitantly shorter steps [88]. Moreover, the shorter keel found in the SACH foot compared to the ESAR foot results in a highly flexible forefoot section and contributes to an earlier onset of forefoot rocker. Close examination of Figure 3B affirms the notion that with the SACH foot the forefoot rocker is initiated earlier (earlier flattening of the line).

Roll-over characteristics

The ability of the human ankle and foot to move in a controlled fashion while the center of pressure progresses forward under the foot does not only result in larger intact step length, it also greatly influences the directional change of the COM necessary during step-to-step transition, and thereby directly influences the transition cost [1]. When taking into account the fact that radii of the roll-over shape were normalized using the measured COM height as opposed to leg length, roll-over shape radii found in this study approximate the aforementioned optimal radius of 0.3 times the leg length. Contrary to our hypothesis, roll-over shape radii found in the current study did not differ between the two prosthetic feet. These results seems to contradict previous findings by Curtze and colleagues (2009) [32] who stated that when feet are tested using quasi-static mechanical loading markedly different radii are found between both prosthetic feet. However, inherent mechanical properties of the prosthetic feet might be subdued due to alignment alterations made by the prosthetist when fitting the prosthesis [90]. Moreover, subjects might alter their kinematics in order to ensure a foot-ankle curvature with a radius of .3 times the leg length [32, 90]. As roll-over shape radii did not differ between feet during walking this factor does not affect the difference in step-to-step transition cost found between both prosthetic feet. However, in accordance with literature [91], the total arc length was shorter in the SACH foot as opposed to the ESAR foot. The shorter keel length of the SACH foot directly influences the total arc length as it limits COP forward progression during late stance. The shorter arc length, and concomitantly shorter COP forward progression could possibly contribute to the larger collision loss found in the SACH foot as it increases the necessary directional change of the COM at a given step length [124]. Besides the shorter forward COP progression and the earlier onset of forefoot rocker in the SACH prosthesis, strikingly large similarities are found in the shape of the progression curve between both feet (Figure 3B), indicating that inherent differences in prosthetic roll-over shape characteristics are attenuated when walking with the prosthesis.

The optimal prosthesis

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In order to achieve a reduction in metabolic energy cost the energy storage and release of the prosthetic foot needs to be of the right size 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 [147, 192, 241]. Additionally, the period in gait during which energy is stored [192] and the timing of energy release [129, 181] are factors that, if not optimal, can attenuate any positive effects of an increased push-off work on metabolic energy cost [88]. Recently an ESAR foot has been developed that is able to actively control the timing of the energy release [26]. This prosthesis has been shown to enable push-off work similar to that of a biological ankle. However, whereas positive results were found on metabolic energy cost in able-bodied subjects [26] it led to an unexpected increase in metabolic cost in people with a lower limb amputation [192]. Contrary to the passive devices, the new generation 'bionic' prostheses are able to generate net positive work during the stance phase of walking [6, 219]. A recent study showed that metabolic energy cost can be reduced towards that of able-bodied walkers [98]. Although promising, it is important to note that these prostheses use batteries to generate ankle work, which would, theoretically, result in a metabolic cost below that of able-bodied walkers.

Evidently, the efficacy of a prosthesis to reduce the metabolic energy cost is not merely related to the ability to reduce the step-to-step transition cost, other factors like for example the prosthetic-stump interface, an altered balance control [110, 218, 228], alterations in swing-leg kinematics [53], compensational proximal joint moments [68, 192], and the inability to use energy transfer through bi-articular muscles [158] could influence overall metabolic energy requirement. Current results underscore the importance to further elucidate how these factors affect metabolic energy cost and attenuate the positive effect of a decrease in step-to-step transition cost, in order to further optimize prosthetic design.

Limitations

One of the limitations of this study was the relative short accommodation period of one day. 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. In a study by Grabowski and colleagues (2010) [87] no differences on metabolic energy cost were found after three days while differences after 21 days were significant. This might indicate that a minimum of at least more than three days is required to detect differences in metabolic energy cost. Whether the suggested adaptation time also applies to mechanical outcome measures remains undetermined. With an adaptation time of one day we did find the anticipated differences, though it is unclear whether these would in-, or decrease after more adaptation time. Moreover, contrary to Grabowski and colleagues (2010) [87] 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.

Because outcome variables are greatly influenced by walking speed, we had subjects walk at a set walking speed of 1.20 $m \cdot s^{-1}$. To compare this speed to subjects' preferred walking speed we had subjects walk at their own preferred walking speed for 40 meters through a corridor while walking speed was determined based on the time it took to walk the middle 20 meters. Although the preferred speed (1.27 $m \cdot s^{-1}$) differed significantly ($p = .032$) from the 1.2 $m \cdot s^{-1}$ enforced during the measurement, differences are small and maybe clinically irrelevant. However, we cannot exclude that deviation from subjects' preferred walking speed might have influenced the results.

Conclusion

This study showed that walking with the widely prescribed ESAR foot resulted in a lower step-to-step transition cost compared to walking with the conventional SACH prosthesis. This difference is explained by the higher amount of positive work performed by the ESAR foot during push-off and the larger arc length of the roll-over shape compared to the SACH foot. The lack of evidence supporting a reduction in metabolic energy cost while walking with an ESAR foot suggests that other factors outside those related to step-to-step transition cost contribute to the overall metabolic energy cost. In order to enhance prosthetic development, it remains a formidable challenge to disentangle and optimize these factors in order to reduce the metabolic energy cost while walking with a prosthesis.