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Greater external negative mechanical work is accompanied by a greater metabolic cost of walking for socket-suspended versus bone-anchored prosthesis users with transfemoral limb loss

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ABSTRACT

Background: Bone-anchored prostheses may address the high metabolic cost of walking in transfemoral (i.e., above-knee) amputees if they mitigate the mechanical energy dissipated due to socket use, reflecting a lower metabolic cost of walking than socket-suspended prosthesis users. Therefore, we compared external mechanical work between socket-suspended and bone-anchored transfemoral prosthesis users during walking and identified if these differences are accompanied by those in their metabolic costs of walking. We hypothesized that socket users would perform more net and negative external mechanical work in their prosthetic limb over the gait cycle and exhibit a higher metabolic cost of walking than bone-anchored prosthesis users.

Methods: High-functioning (Medicare K-level ≥ 3, no dysvascular limb loss) transfemoral amputees with a socket-suspended or bone-anchored prosthesis were recruited ($N = 12$ per group). Participants walked at 1.0 m/s on a treadmill as pulmonary gases were measured for metabolic cost estimates and overground as ground reaction forces were measured to calculate net and negative external mechanical work across the gait cycle. Metabolic cost and prosthetic-limb external mechanical work outcomes were compared between groups using independent samples *t*-tests ($\alpha < 0.05$) corrected for multiple comparisons.

Findings: Net mechanical work done by the prosthetic limb was not significantly different between groups ($p = 0.93$, $g = 0.64$). However, socket users showed greater external negative mechanical work ($p = 0.005$, $g = 1.17$), which was accompanied by a 10.13 % greater metabolic cost of walking than bone-anchored prosthesis users ($p = 0.03$, $g = 0.54$).

Interpretation: Bone-anchored prostheses may address the high metabolic costs of transfemoral amputee walking by mitigating socket-related energy loss.

1. Introduction

Transitioning from one period of single-support to the next during a walking gait cycle involves considerable external mechanical work to redirect the body's center of mass during periods of double-support (Adamczyk and Kuo, 2009; Donelan et al., 2002a). In able-bodied gait, the leg leading the step during double-support performs negative mechanical work, dissipating mechanical energy upon foot-ground collision, while the trailing leg concomitantly performs positive mechanical work to redirect the body's center of mass upward and forward into the next single-support period (Donelan et al., 2002a). Such external mechanical work performed by the ground reaction forces on the body's center of mass during step-to-step transitions exacts a proportional metabolic cost among able-bodied persons (Donelan et al., 2002b). Similarly, the mechanical work done on the body's center of mass during the complete gait cycle strongly correlates with metabolic energy consumption during able-bodied walking (Burdett et al., 1983). However, unilateral above-knee (i.e., transfemoral) limb loss is accompanied by deviations from able-bodied gait mechanics. Notably, individuals with a transfemoral prosthetic limb walk with smaller propulsive and vertical ground reaction forces on the prosthetic side during stance compared to able-bodied persons (Carse et al., 2020). Consequently, external

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mechanical work also differs between able-bodied individuals and those with lower limb loss (Bonnet et al., 2014; Houdijk et al., 2009). Individuals with transfemoral limb loss who are high-functioning, i.e., very mobile, strong, and fit, still, on average, have a high metabolic cost of walking compared to able-bodied individuals (Jarvis et al., 2017; Miller et al., 2024), motivating determination of whether other modifiable characteristics like prosthesis characteristics can substantially affect the work and energy of walking. Bone-anchored prostheses are increasingly prescribed as a mobility solution to transfemoral limb loss (Frölke et al., 2017) and appear to have some benefits for mobility and physical function versus traditional socket-suspended prostheses (Kooiman et al., 2023). However, it is unclear if bone-anchored prostheses affect the work and energy of walking, questioning if they can be recommended with confidence as a solution to address one of the main biomechanical issues of transfemoral limb loss (i.e., high metabolic cost).

Bone-anchored prostheses differ from socket-suspended prostheses in that their abutment is anchored directly to the residual bone using osseointegration, bypassing a socket and its challenges (Frölke et al., 2017). The residual limb rotates and translates inside the socket which is related to poor patient-reported outcomes (Gale et al., 2020). Cross-sectionally, socket-suspended prosthesis users experience a larger degree of prosthesis pistoning, i.e., vertical motion of the residual limb relative to the prosthesis, compared to bone-anchored prosthesis users (Kooiman et al., 2023) and may consequently perform more negative external mechanical work in the prosthetic limb than bone-anchored prosthesis users. The energy lost due to rotations and translations at the socket-residual limb interface must be restored by positive mechanical work such that the net work of the limb (i.e., sum of positive and negative work) nears zero, incurring a lower metabolic cost of walking than if the net work were negative. Presuming the interface between the residual limb and the socket is compliant and heavily damped, it seems reasonable that socket-suspended users will perform net negative mechanical work, which may then be accompanied by a higher metabolic cost of walking than bone-anchored prosthesis users. This socket-related energy loss expectation is additionally motivated by findings from Miller, Bell and Esposito (Miller et al., 2024) who showed a 1 % increase in the metabolic cost of optimal control simulations of walking when pistoning (i.e., an additional degree of freedom) was added to a rigid osseointegration-type residual limb-prosthesis interface model. Relatedly, Kooiman et al. (Kooiman et al., 2023) reported oxygen consumption in treadmill walking was ~28 % greater among socket-suspended prosthesis users who also exhibited more pistoning than bone-anchored prosthesis users. However, it is unclear from the aforementioned studies if a larger degree of motion at the residual limb-prosthesis interface translates to socket compared to bone-anchored prosthesis users performing (i) more negative external mechanical work reflecting energy loss, and (ii) more positive external mechanical work to replace the energy lost, which in turn is energetically costly. Clarifying whether differences in limb-level work of socket-suspended prosthesis users and bone-anchored prosthesis users are accompanied by differences in the metabolic cost of walking can help establish cause (s) of changes in metabolic cost when revising a socket into a bone-anchored interface, and the benefits expected from bone-anchoring as a mobility solution after limb loss.

Therefore, the aims of this study were (i) to examine the differences in external mechanical work between socket-suspended prosthesis users and bone-anchored prosthesis users with unilateral transfemoral limb loss during walking, and (ii) identify if these differences are reflected in their metabolic costs of walking. Given the potential for substantial energy loss due to deformations and damping at the socket-residual limb interface (Kooiman et al., 2023), we hypothesized that socket-suspended prosthesis users will (i) perform more net and negative external mechanical work in their prosthetic limb than bone-anchored prosthesis users over the gait cycle, and (ii) exhibit a higher metabolic cost of walking than bone-anchored prosthesis users. In addition to the mechanical work performed over the entire gait cycle, mechanical work performed during step-to-step transitions exacts a proportional metabolic cost among able-bodied adults (Donelan et al., 2002b) and differs between able-bodied persons and individuals with lower limb loss (Bonnet et al., 2014; Houdijk et al., 2009). Thus, our exploratory aim was to compare the mechanical work performed by individual limbs during step-to-step transitions between socket-suspended and bone-anchored prosthesis users during walking. We expected that the net and negative external mechanical work performed by the prosthetic limb when it is leading a step will be greater among socket-suspended users compared to bone-anchored prosthesis users.

2. Methods

2.1. Subjects

We performed an a priori power analysis (G*Power version 3.1.9.6) for sample size estimation based on Houdijk, Pollmann, Groenewold, Wiggerts and Polomski (Houdijk et al., 2009), who compared the double-support external mechanical work performed by the intact and prosthetic limbs of individuals with below-knee limb loss to that performed by able-bodied controls. Using their effect size of 1.57, with an error probability of 0.05 and power of 0.80 revealed that $N = 16$ subjects were required for an independent samples t-test. Thus, we recruited $N = 24$ individuals with transfemoral limb loss for this cross-sectional study. Twelve individuals were traditional socket-suspended prosthesis (SO) users, and twelve were bone-anchored prosthesis (BAP) users. For both groups, participants were included and considered "high-functioning" if they (i) had a unilateral transfemoral amputation and walked with a prosthesis, (ii) were 18–65 years old, (ii) were at least 18 months past their most recent limb loss-related surgery and rehabilitation, (iv) were not diagnosed with metabolic syndrome or hypertension, (v) were cleared by their medical professional to walk without assistance other than their prosthesis, (vi) presented no chronic or functional limitations that hindered their ability to walk unaided, (vii) had been using their current prosthesis for at least six months, and (viii) had a Medicare K-level of 3–4 indicating a high level of mobility. Participants were excluded if they had undergone limb loss due to dysvascular complications from diseases like diabetes. All participants self-reported frequent participation in at least one additional form of recreational sport or exercise including cycling, running, hiking, swimming, tennis, weight-lifting, CrossFit, wrestling, football, and/or basketball. Given that aspects of physical fitness like muscle strength can influence the metabolic cost of walking among individuals with limb loss (Jarvis et al., 2017; Russell Esposito and Miller, 2018), we chose to recruit high-functioning individuals in both groups to control for these effects and increase confidence that effects could be attributed to the dependent variable (prosthesis abutment type) vs. health- and fitness-related confounders. All participants provided written informed consent to participate in study procedures. This study was approved by the University of Maryland College Park's Institutional Review Board.

2.1.1. Experimental setup

Treadmill walking was performed on a research-grade treadmill (Trackmaster, FullVision Inc., Kansas, USA). Pulmonary gases were measured breath-by-breath using a cart-based indirect calorimeter (TrueOne 2400, Parvo Medics, Salt Lake City, UT, USA). During overground walking, ground reaction forces were measured at 1000 Hz using eight piezoresistive force platforms with a sensitivity of 0.1 N (model 9260AA; Kistler, Winterthur, Switzerland), and positions of retroreflective markers were measured at 200 Hz using a 13-camera optical motion capture system (Vicon, Oxford, UK).

2.1.2. Experimental protocol

Participants were instructed to refrain from eating a substantial meal at least 3 h before visiting the laboratory, and to refrain from vigorous

exercise the day before and day of their participation. Upon arrival, a certified prosthetist ensured each participant's prosthesis was fit comfortably. Metabolic cost measurements began with a two-minute familiarization period of walking on a treadmill at the participants' preferred speed. Subsequently, the treadmill belt speed was changed to 2.2 miles/h, which is ~1 m/s. All participants walked at this speed for ten minutes as their oxygen uptake and carbon dioxide output were measured using a breath-by-breath analyzer. Participants wore comfortable light-weight clothing, their own footwear, and their customary prosthesis for walking. All treadmill tests were performed at a level grade.

Participants were allowed to rest following treadmill walking for as long as they required before beginning the gait analysis procedures, to minimize fatigue effects. Participants wore form-fitting clothes, and the same footwear and prosthesis used in the treadmill test. We affixed retroreflective markers bilaterally to participants' anterior-superior iliac spine, posterior-superior iliac spine, iliac crest, greater trochanter, and one marker on the sacrum at the L5-S1 joint location to calculate anterior-posterior pelvis velocity as a surrogate for walking speed during subsequent data processing. Participants performed ten overground walking trials at a set speed of ~1 m/s across a 15-m level walkway. Walking speed was standardized to control for its known effects on ground reaction forces in persons with transfemoral limb loss (Kobayashi et al., 2022). The speed of 1 m/s was selected based on previous studies of high-functioning individuals with above-knee limb loss that we could be confident most eligible participants would be able to perform, while still being fast enough to be relevant for daily living in this population (Jarvis et al., 2017; Russell Esposito et al., 2018). The walkway was embedded with eight force plates measuring three-dimensional forces and thirteen motion capture cameras recording three-dimensional marker positions. Walking speed during data collection was monitored using a hand-held stopwatch to provide participants with feedback on modifying their speed to walk near 1 m/s. The actual speed of walking was later verified using marker data during data processing and trials that were not between an average speed of 0.95–1.05 m/s were excluded.

2.1.3. Data processing

Metabolic rate was calculated from pulmonary gas rates averaged over the final three minutes of steady-state treadmill walking using the equation from Brockway (Brockway, 1987). Metabolic cost was then calculated by dividing the metabolic rate by body mass (including the prosthesis mass) and by average walking speed.

Three-dimensional ground reaction forces were measured such that the anterior-posterior force ($F_x(t)$) was measured along the x-axis, the mediolateral ($F_y(t)$) along the y-axis, and the vertical force ($F_z(t)$) along the z-axis (Fig. 1). Forces were filtered using a zero-lag, fourth-order, low-pass Butterworth filter with a cutoff frequency of 45 Hz in Visual 3D software (HAS-Motion, Ontario, Canada). For our primary research question, we calculated external mechanical work (i.e., mechanical work performed on the body's center of mass) using the individual (lower) limbs method outlined by Donelan et al. (2002) (Donelan et al., 2002a) and detailed in Supplementary Material 1. We applied this method to calculate the mechanical work done by each limb – intact and prosthetic (i.e., amputated) – over five full strides. A stride was defined as beginning at the foot strike of one limb and ending when the same limb subsequently contacted the ground. With a force platform sensitivity of 0.1 N, the instants of foot-strike and toe-off were determined when the ground reaction force magnitude exceeded and fell below one newton, respectively.

External mechanical work for each limb was calculated in MATLAB (The MathWorks Inc., Natick, MA, USA) as the cumulative time-integral of external mechanical power, where the power for each limb was defined as the dot product of the external ground reaction force acting on the limb and center of mass velocity (Donelan et al., 2002a). Mechanical work was separated into positive and negative, representing energy generation and dissipation, respectively. While net and negative mechanical work performed by the prosthetic limb was our primary focus, we calculated intact-limb negative mechanical work and positive mechanical work for additional information regarding external mechanical work patterns in our population of interest. Positive mechanical work was calculated by restricting the integral to positive portions of the power-time curve while negative mechanical work was calculated by restricting the integral to negative portions of the power-time curve for each limb. The total positive and negative work performed during a

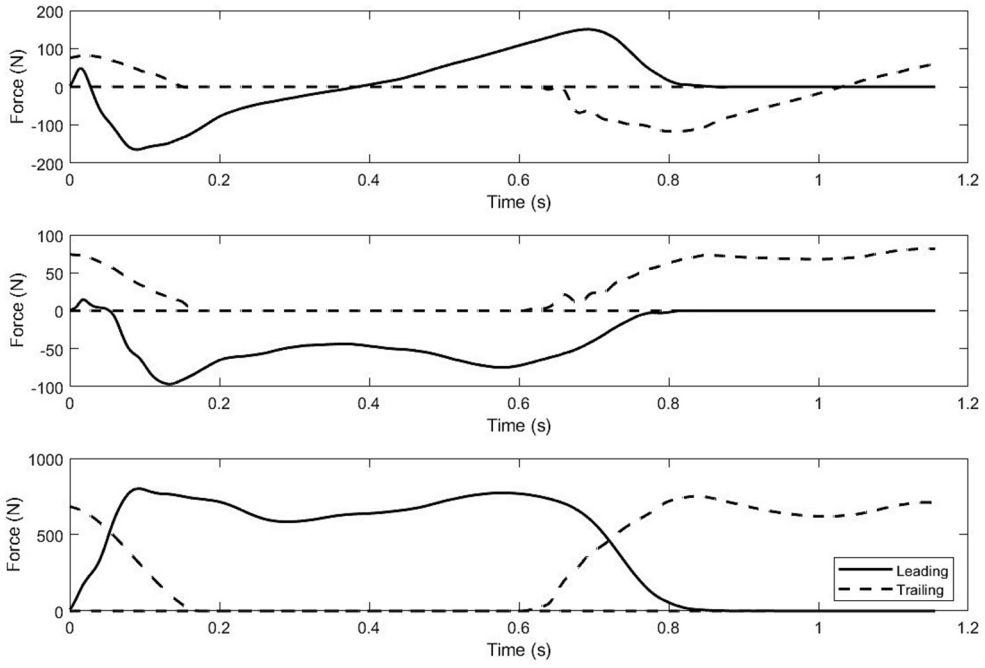


Fig. 1. Example anterior-posterior (top), mediolateral (middle), and vertical (bottom) ground-reaction forces generated by the leading and trailing limbs during a stride of walking.

given stride was calculated as the sum of the positive and negative work done by both limbs, respectively. The net work performed individually by the prosthetic and intact limbs, and by both limbs combined, during a given stride was calculated as the sum of their positive and negative work. The equations used to calculate external mechanical power and work are detailed in Supplementary Material 1.

Marker data was used to define the pelvis segment, and its center of mass anterior-posterior velocity was used as a proxy for average walking speed. The pelvis segment was defined as a Visual3D pelvis, using the right and left anterior-superior and posterior-superior iliac spine and iliac crest markers affixed to the hips, and its motion was tracked by recording their X, Y, Z positions in the global coordinate system (X = anterior-posterior, Y = mediolateral, Z = vertical). Each marker trajectory was filtered using a zero-lag, fourth-order, low-pass Butterworth filter with a cutoff frequency of 6 Hz in Visual 3D software.

2.2. Statistical analysis

All calculations of primary outcome variables (i.e., external net and negative mechanical work by the prosthetic limb) were performed on five strides and subsequently averaged across strides to obtain work values for each subject used in statistical analysis. We used one-tailed independent samples *t*-tests ($\alpha = 0.05$) to compare external net and negative mechanical work performed by the prosthetic limb over the stride and metabolic cost between groups. Given that three hypothesis-related *t*-tests were performed, the threshold for significance was adjusted using the Benjamini-Hochberg method (Benjamini and Hochberg, 2018). As a secondary analysis, positive mechanical work performed by the prosthetic and intact limbs, net and negative mechanical work performed by the intact limb, and total positive and negative mechanical work across the gait cycle were compared between groups using two-tailed independent samples t-tests to provide additional information regarding external mechanical work patterns in this population. Wilcoxon rank sum tests were used for non-normally distributed data, as determined using the Shapiro-Wilk tests for normality ($\alpha = 0.05$). We calculated Hedges' g effect sizes for each comparison (Borenstein et al., 2009).

2.2.1. Exploratory analysis

As an exploratory analysis, we calculated external mechanical work over a step, using eqs. 2–15 (Supplementary Material 1), whereby a step was defined as beginning at the foot-strike of the prosthetic limb (i.e., leading limb) and ending at the foot-strike of the intact limb (i.e., trailing limb). The integration constants used in eqs. 4–6 (Supplementary Material 1) were determined by replicating the methods in Donelan, Kram and Kuo (Donelan et al., 2002a). External mechanical work for a step was calculated during double-support, single-support, and for the complete step. The statistical methods described above were used here, comparing positive, negative, and net work performed by the prosthetic and intact limbs during double-support, positive, negative, and net work performed by the prosthetic limb during single support, and total positive and negative work performed over a step.

3. Results

Table 1
Demographic and anthropometric characteristics for the socket-suspended (SO) and bone-anchored prosthesis (BAP) user groups.

	SO	BAP	p-value	Hedges' g
Sex distribution	F = 4, M = 8	F = 4, M = 8	–	–
Age (years)	40.25 ± 13.40	47.66 ± 10.11	0.14	0.62
Height (meters)	1.77 ± 0.07	1.73 ± 0.07	0.38	0.47
Body mass (kg)	81.59 ± 14.44	77.35 ± 20.20	0.56	0.24
Prosthesis mass (kg)	4.56 ± 0.59	4.58 ± 0.79	0.93	0.03
Time since limb loss (years)	9.2 ± 10.3	12.4 ± 7.7	0.39	0.33
Time since bone-anchor surgery (years)	–	3.3 ± 1.3	–	–
Source of limb loss: trauma/cancer/congenital/infection	7/1/3/1	8/3/0/1	–	–
Preferred overground walking speed (m/s)	1.21 ± 0.16	1.09 ± 0.14	0.07	0.79
Preferred treadmill walking speed (m/s)	0.87 ± 0.16	0.94 ± 0.19	0.37	0.39

* $p < 0.05$, indicates a statistically significant difference between groups. Note: prosthesis mass was calculated from force data obtained by placing the prosthetic leg on the force plate. Data collected at preferred walking speed were not analyzed, but preferred walking speed was measured and is reported here to provide participant background.

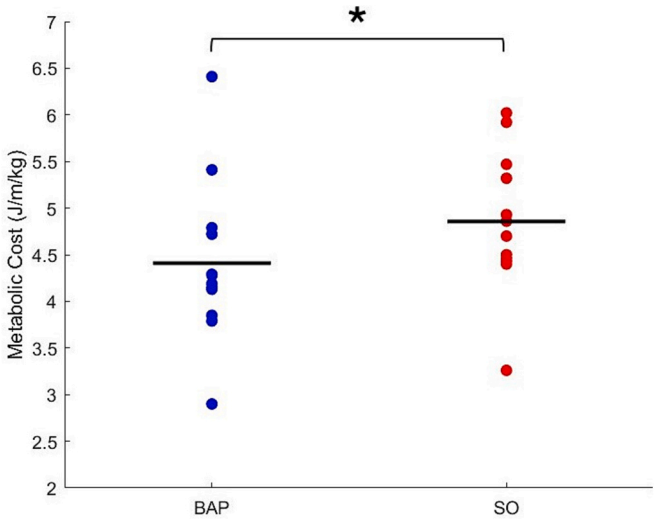


Fig. 2. The metabolic cost of walking (y-axis) for the socket-suspended prosthesis (SO) users group and the bone-anchored prosthesis (BAP) user group walking at 2.2 miles/h on a treadmill. The horizontal black bars represent the group means. *$p < 0.05$, indicating the difference was statistically significant between groups; BAP = bone-anchored prosthesis group, SO = socket-suspended prosthesis group, J/m/kg = joules/m/kg.

3.1.1. Primary analysis

Demographics and anthropometrics for both groups are presented in Table 1. There were no significant differences between groups in sex distribution. All but one participant (belonging to the SO group) used microprocessor prosthetic knees.

The metabolic cost of walking was 10.13 % greater in the SO (4.85 ± 0.75 J/m/kg) compared to BAP groups (4.40 ± 0.87 J/m/kg; $p = 0.03$, g = 0.54) (Fig. 2), determined using a one-tailed independent samples *t*-test.

Using a one-tailed independent samples t-test, net mechanical work performed by the prosthetic limb over the gait cycle was not significantly different between groups ($p = 0.93$, g = 0.64) (Fig. 3). The SO group performed significantly more external negative mechanical work with their prosthetic limb than the BAP group over the gait cycle ($p = 0.005$, g = 1.17) (Fig. 4), determined using a Wilcoxon ran sum test.

In the secondary analysis, external positive mechanical work by the intact and prosthetic limbs, negative work by the intact limb, net work by the intact limb and both limbs combined (Fig. 4), and total positive and negative work were not different between groups (all $p > 0.06$)

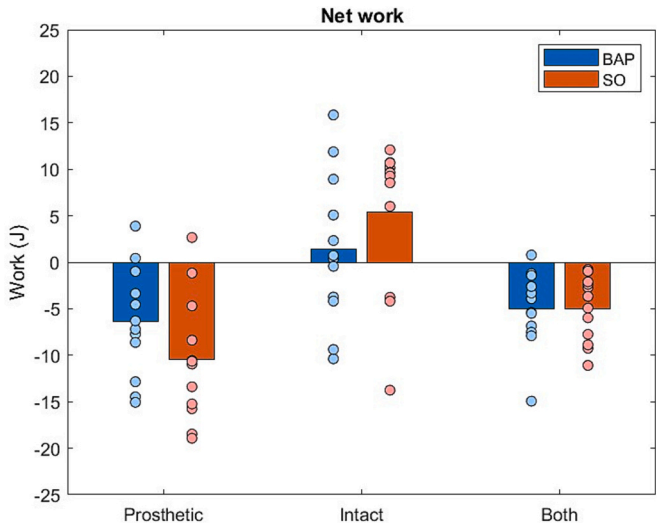


Fig. 3. Net external mechanical work performed over the gait cycle by the prosthetic and intact limbs individually, and both combined to yield overall net work during the gait cycle; BAP = bone-anchored prosthesis group, SO = socket-suspended prosthesis group, J = joules.

(Fig. 4), determined using a two-tailed independent samples *t*-test.

3.1.2. Exploratory analysis

During double-support, the SO group performed significantly more external negative mechanical work ($p = 0.001$, g = 1.35) and net work ($p \leq 0.001$, g = 1.43) with their prosthetic limb (i.e., leading limb), and total negative work ($p = 0.008$, g = 1.18) for the entire step than the BAP group (Fig. 5).

4. Discussion

We aimed to determine if the external net and negative mechanical work performed by the prosthetic limb during the gait cycle is greater for high-functioning transfemoral amputees who use a socket-suspended versus bone-anchored prosthesis, and if this work is accompanied by a higher metabolic cost of walking. The inclusion/exclusion criteria for being deemed "high-functioning" were noted in the Introduction. We expected that the energetic consequences of performing significantly more net and negative mechanical work in the prosthetic limb would be reflected in a substantially greater metabolic cost of walking for the socket-suspended prosthesis users than the bone-anchored prosthesis users. Our hypothesis was motivated by past findings among able-bodied persons that the work done to redirect and accelerate the body's center of mass exacts a substantial metabolic cost of walking (Donelan et al., 2002b; Kuo et al., 2005) and comprises nearly 45 % of the net metabolic cost of walking (Grabowski et al., 2005). While net mechanical work did not differ between groups, SO users performed 30 % more external negative mechanical work in their prosthetic limb across the gait cycle than BAP users, partially confirming our first hypothesis. Substantially more net and negative mechanical work was also performed by their prosthetic limb during the double-support phase of step-to-step transitions than BAP users. This finding was accompanied by a 10 % greater metabolic cost in the SO than BAP group which aligns with past cross-sectional evidence that high-functioning SO users consume more oxygen than BAP users, as well as a reduction in oxygen consumption observed longitudinally among SO users who transition to an osseointegrated implant due to socket-related complaints (Kooiman et al., 2023; Van de Meent et al., 2013). Thus, our findings suggest that the mechanical energy dissipated due to socket use may cost metabolic energy, as reflected in a greater metabolic cost of walking for SO than BAP users. However, we emphasize that we cannot causally relate these mechanical and metabolic energies with the present study design (see limitations).

In the present study, SO users dissipated 8.2 joules and 6.0 joules more mechanical energy in their prosthetic limb compared to BAP users during the gait cycle and the double-support phase of step-to-step transitions, respectively. Given that the net external mechanical work (i.e., positive work + negative work) did not differ significantly between groups across the gait cycle, SO users would need to eventually perform additional positive work to compensate for the substantially greater negative work their prosthetic limb performed relative to BAP users such that the inter-group net external mechanical work is comparable. The

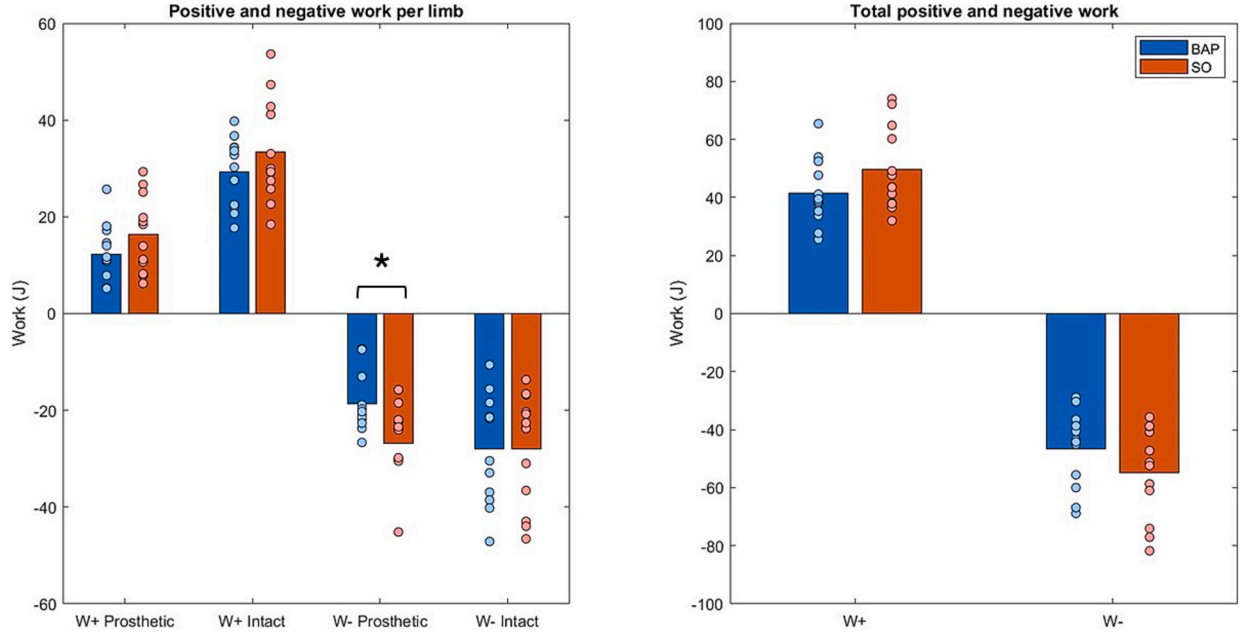


Fig. 4. Positive and negative external mechanical work performed by the intact and prosthetic limbs (left-side graph), and total positive and negative work performed by both limbs combined (right-side graph) over the gait cycle. *p < 0.05, indicating the difference was statistically significant between groups; BAP = bone-anchored prosthesis group, SO = socket-suspended prosthesis group, J = joules.

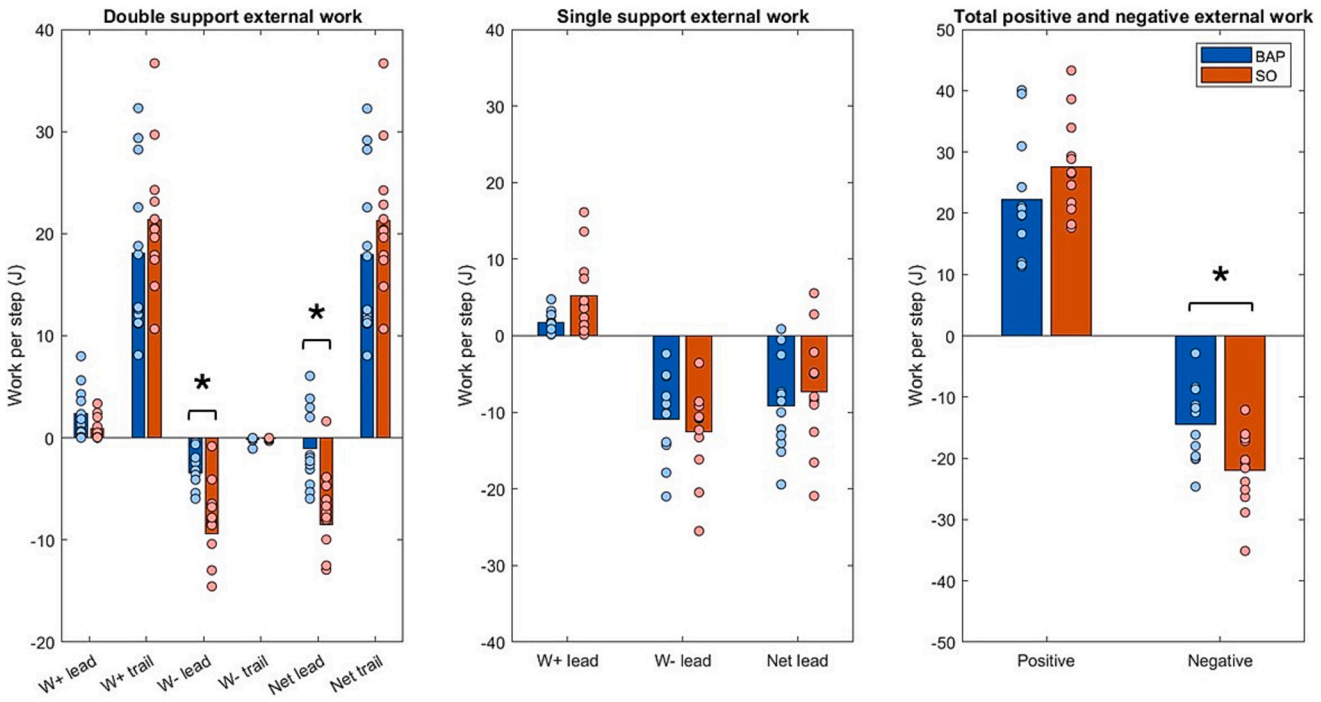


Fig. 5. External mechanical work (y-axes) performed by the prosthetic limb (i.e., leading) and intact limb (i.e., trailing) during the double-support phase (left), single-support phase (middle), and over the entire step (right) of step-to-step transitions. *indicates a statistically significant difference between groups ($p < 0.05$).

need to replace the lost energy through concentric muscle actions likely requires a higher metabolic energy expenditure for SO users than BAP users. Over a typical daily step count of 6100 steps (or 3050 strides) (Hagberg et al., 2023), replacing an additional 6–8 joules of energy per gait cycle with positive mechanical work could result in a greater energetic cost for SO users than BAP users during the day, limiting their engagement in activities of daily living that contribute to physical function.

Compliance at the residual limb-prosthesis interface during the gait cycle may explain why SO users performed more negative mechanical work with their prosthetic limb than BAP users, reflecting a higher metabolic cost of walking. Unlike a BAP's rigid attachment, the residual limb rotates between 7 and 11 degrees and translates up to 3 cm inside the socket (Gale et al., 2020). The materials deformed by these motions are typically compliant and heavily damped, e.g. muscle, fat, skin, socket liners, potentially dissipating mechanical energy as the limb supports body weight and redirects the body's center of mass. This hypothesis finds some support in past optimal control simulations of walking using a three-dimensional model of transfemoral limb loss, which found a higher metabolic cost of walking for a musculoskeletal model with pistoning (i.e., an added degree of freedom) at the residual limb-prosthesis interface than one that simulated a rigid osseointegration-type attachment (Kooiman et al., 2023). However, we would need to progressively add degrees of freedom at the residual limb-prosthesis interface, progressing to three rotations and three translations, and examine its effects on metabolic cost to adequately assess if socket-related compliance contributes to the negative mechanical work done by the prosthetic limb and to the whole-body metabolic cost of walking. Energy loss in the socket could also potentially be quantified from instrumented gait analysis data using deformable models similar to those used to quantify mechanical energy of prosthetic limbs (Takahashi et al., 2012).

In addition to the compliance at the residual limb-prosthesis interface, variables like center of mass or trunk excursions could play a role in the magnitude of mechanical energy dissipated at the residual limb-prosthesis interface and perhaps, thereby, the energy expended to walk. For instance, mediolateral trunk excursion is positively associated with pistoning in transfemoral socket users who consume 28 % more oxygen during walking than bone-anchored prosthesis users (Kooiman et al., 2023). From the present data on external work, however, we cannot infer whether trunk excursion, or any other kinematic variable for that matter, was a mechanistic pathway between external mechanical work and the metabolic cost of walking for this population.

Currently, bone-anchoring is primarily used as a surgical revision for individuals with transfemoral limb loss who have poor outcomes with a socket. In the present BAP group, only two (of 12) participants had never had a socket. The other 10 participants all first used socket-based prostheses for on average 10 years before electing to have a BAP revision due to poor mobility, in contrast to the present group of socket-based prosthesis-users who were highly mobile with their sockets and were not seeking a BAP revision. Yet, BAP users were more economical walkers than the high-functioning SO users in our study, conducted post-surgery. Thus, we hypothesize that BAP use enables socket users with socket-related challenges to regain mobility and the ability to walk more economically than they did pre-surgery, as previously seen (Kooiman et al., 2023; Van de Meent et al., 2013). Without pre- and post-surgery physical function and energy expenditure data, however, this hypothesis cannot be tested and thereby remains speculative.

The effect of prosthesis attachment type on mechanical energy dissipation and metabolic cost is likely confounded by walking speed and step length. Among able-bodied persons, the redirection of the body's center of mass velocity during walking is mainly determined by walking speed and step length (Adamczyk and Kuo, 2009). Walking speed affects oxygen consumption in able-bodied persons and those with transfemoral limb loss (Russell Esposito et al., 2018). Consequently, these variables may contribute to the differences in mechanical work and metabolic cost seen between the SO and BAP groups in this study in addition to the effects of prosthesis attachment type (i.e., socket-suspended versus bone-anchored). Walking speed, calculated as the anterior-posterior velocity of the pelvis, was not significantly different between groups (SO = 1.04 ± 0.02 m/s, BAP = 1.01 ± 0.08 m/s, *p*-value = 0.32, hedges g = 0.41). Similarly, neither step (SO = 0.65 ± 0.05, BAP = 0.64 ± 0.03, p-value = 0.48, g = 0.24) nor stride length (SO = 1.29 ± 0.08, BAP = 1.24 ± 0.05, p-value = 0.17, g = 0.74)

significantly differed between the SO and BAP groups. These findings suggest that prosthesis attachment type rather than walking speed and step or stride length likely influenced the metabolic cost of walking and the body's center of mass motion and, consequently, the external negative mechanical work performed by the prosthetic limb during the gait cycle in our study. While examining the asymmetry of spatiotemporal variables was beyond our scope, it is tied to metabolic cost in able-bodied persons (Ellis et al., 2013; Roemmich et al., 2019) and could, if significantly different between SO and BAP groups, play a role in energy utilization by each group. However, the only study to date on the relationship between spatiotemporal measures and energy expenditure during walking in transfemoral amputees specifically suggests no significant correlation between energy expenditure and spatiotemporal measures including step length (Mahon et al., 2019).

Some potentially limiting considerations should be made. External mechanical work represents the work performed by the limbs to accelerate the body's center of mass. It is not independent of the work done to move body segments relative to the body's center of mass and does not precisely reflect the work generated by individual muscles (Kautz and Neptune, 2002). Consequently, musculotendon work performed during the gait cycle is underestimated (Sasaki et al., 2009). Thus, while a socket-suspended prosthesis results in greater external mechanical energy dissipation than a bone-anchored prosthesis, internal and musculotendon work should be quantified to understand the energetic consequences of prosthesis-related differences in mechanical energy dissipation among individuals with transfemoral limb loss.

Further, we note that the net work during the complete gait cycle was non-zero for both groups in our study, likely due to aperiodic strides. Small differences in the height or velocity of the center of mass between the beginning and end of a stride can alter potential and kinetic energy estimates, yielding a non-zero change in mechanical energy over the complete stride. Thus, non-zero net external work values are possible, as also previously seen (Sasaki et al., 2009).

Socket fit could affect the amount of pistoning and other motions of the residual limb relative to the prosthesis, in turn potentially affecting limb work and metabolic cost. Here we prioritized subjective comfort in the fit of the SO group. Other protocols for socket fit could potentially reduce variables like pistoning and energy loss by the prosthetic, though perhaps at the expense of comfort.

Finally, our metabolic cost results could have been influenced by the distribution of the cause of amputation (Waters et al., 1976), which was not exactly the same between groups (Table 1). Waters, Perry, Antonelli and Hislop (Waters et al., 1976) found metabolic cost to be different between individuals with limb loss due to trauma vs. vascular etiology. However, age and walking speed differences between their groups likely confound their ability to attribute differences in metabolic cost to the cause of amputation specifically. Trauma accounted for most of the limb loss in both our groups, and individuals with limb loss due to dysvascular disease, who tend to have a higher metabolic cost of walking than individuals with traumatic limb loss (Waters et al., 1976), were excluded from the present study. Unlike the groups in Waters, Perry, Antonelli and Hislop (Waters et al., 1976), our groups did not differ significantly in age or walking speed, both of which affect metabolic cost (Bobbert, 1960; Das Gupta et al., 2019; Russell Esposito et al., 2018). There were three individuals in the SO group with congenital limb loss and no such individuals in the BAP group, but these three individuals had relatively low metabolic costs within their group (4.51 J/m/kg vs. 4.97 J/m/kg for the rest of the group), and there is no prior evidence that individuals with congenital limb loss have unique mechanics and energetics of gait. We therefore think it is unlikely the present results were confounded by differences in the causes of limb loss.

Similarly, participants were instructed to use their customary prosthesis for walking, and one individual in the SO group used a prosthesis with a mechanically passive knee. All other participants used prostheses with microprocessor-controlled knees. Some studies report a reduction in metabolic cost when using microprocessor vs. passive knees (Seymour et al., 2007), so this difference in prosthesis type is a potential confounder. However, findings on changes in metabolic cost with microprocessor knees are inconsistent between studies, particularly when testing moderate or preferred walking speeds (Wong et al., 2012). This individual had a below-average metabolic cost in their group (4.70 vs. 4.87 J/m/kg for the rest of their group), and the statistical outcomes were robust to their exclusion, so we elected to include them.

We were limited by the lack of a force-instrumented treadmill, hence the metabolic data was captured during treadmill walking while the ground reaction force data was captured during overground walking. The lack of a common walking surface and setting could have contributed to the disproportionate inter-group differences in external negative mechanical work and the metabolic cost of walking. More generally, the metabolic and mechanical energy data were not measured simultaneously, so we hesitate to make any strong causal suggestions relating these data. However, metabolic cost is not significantly different during treadmill versus overground walking at comparable speeds among transfemoral amputees (Starholm et al., 2016), and only the peak braking force is greater during overground than treadmill walking with comparable vertical and mediolateral ground reaction forces between surfaces among able-bodied individuals (Lee and Hidler, 2008). Thus, it is unlikely that walking on different surfaces for the metabolic and biomechanics testing contributed to the inter-group differences in external negative mechanical work being substantially larger than the inter-group differences in metabolic cost.

5. Conclusion

High-functioning individuals with transfemoral limb loss who utilize a socket-suspended prosthesis perform significantly more external negative mechanical work with their prosthetic limb during overground walking than those with a bone-anchored prosthesis. This may be potentially due to residual limb motion within the socket, though other explanatory variables not studied here, like mediolateral trunk excursions, merit future examination based on past research (Kooiman et al., 2023). Consequently, a bone-anchored prosthesis may mitigate some of the mechanical energy dissipated due to socket fit-related movement during walking. Such mechanical energy dissipation was accompanied by a greater metabolic cost during treadmill walking for socket-suspended compared to bone-anchored prosthesis users, which suggests that despite pre-surgery socket-related challenges, bone-anchored prosthesis users may be able to walk more economically upon their transition than high-functioning socket-suspended prosthesis users with no socket-related complaints.

CRediT authorship contribution statement

Gauri A. Desai: Writing – review & editing, Writing – original draft, Visualization, Software, Methodology, Investigation, Formal analysis, Conceptualization. **John R. Pope:** Writing – review & editing, Investigation. **Chioma Ezeajughi:** Writing – review & editing, Investigation. **Jae Kun Shim:** Writing – review & editing, Resources, Funding acquisition. **Ross H. Miller:** Writing – review & editing, Supervision, Resources, Project administration, Methodology, Investigation, Funding acquisition, Conceptualization.

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Declaration of competing interest

None.

Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.clinbiomech.2025.106598.

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