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Lower-limb joint work symmetry responses to load carriage and prosthetic foot type during transtibial amputee walking

Stephanie L. Molitor [a,*], Krista M. Cyr [b], Glenn K. Klute [b,c], Richard R. Neptune [a]

[a] Walker Department of Mechanical Engineering, The University of Texas at Austin, Austin, TX, USA
[b] Department of Veterans Affairs, Center for Limb Loss and MoBility, Seattle, WA, USA
[c] Department of Mechanical Engineering, University of Washington, Seattle, WA, USA

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ABSTRACT

Individuals with lower-limb amputations are typically prescribed passive prosthetic feet for everyday walking. However, when carrying additional loads, users demonstrate increased reliance on their intact limb which can lead to overuse injuries. Powered prostheses have been shown to reduce compensations during unloaded walking, but their efficacy during load carriage remains unknown. Furthermore, how the position of an added load affects intact limb reliance is not well understood. This study determined (1) how the presence and placement of an added load affects between-limb joint work symmetry for individuals walking with transtibial protheses, and (2) how prosthetic foot type (passive or powered) affects joint work symmetry during load carriage. Kinematic and kinetic data were collected from unilateral transtibial prosthesis users ($n = 12$) wearing a passive or powered prosthesis while walking with no added load, a 13.6 kg load worn on the front of their torso, or the same load worn on their back. Work symmetry, calculated as the difference between intact-side and prosthesis-side net joint work, was determined for the ankle, knee and hip to quantify intact limb reliance. Ankle and hip work symmetry improved for the Front Load compared to the No Load condition, due to a tradeoff between intact ankle and intact hip demand. Furthermore, ankle work symmetry improved for all load conditions while wearing the powered prosthesis. These results highlight changes in compensatory strategies during different load carriage conditions and encourage the use of powered prostheses to reduce intact limb reliance during activities involving load carriage.

1. Introduction

Individuals with unilateral transtibial amputations rely on the use of prostheses to ambulate and perform activities of daily living. Clinicians typically prescribe passive energy storage and return (ESAR) prostheses with a fixed value for the ankle–foot stiffness tuned to the user's anticipated activity level and body weight. When carrying added loads (e.g., wearing a backpack or carrying children), able-bodied individuals are able to modulate their dynamic ankle joint stiffness to adapt to the changing task demands (e.g., Bayram and Bayram, 2018; Frigo et al., 1996; Kern et al., 2019). However, passive ESAR prosthesis stiffness values cannot be modulated to accommodate changing loads. Therefore, the prosthetic ankle–foot stiffness for individuals using ESAR prostheses may no longer be optimal for their total combined weight when carrying added loads. As a result, previous studies have observed changes in gait kinematics in the presence of an added load, such as increased residual (prosthesis-side) hip and knee flexion (Doyle et al., 2014; Schnall et al., 2014). The intact limb also provides increased spatiotemporal, kinetic and muscle compensations with added loads, as evidenced by increased intact-limb stance time (Schnall et al., 2014; Sinitski et al., 2016), joint power and work (Brandt et al., 2016; Doyle et al., 2014) and knee joint contact forces (Lefranc et al., 2025; Templin et al., 2021).

Previous studies have primarily used weighted backpacks when studying the effects of added loads (e.g., Brandt et al., 2016; Doyle et al., 2014; Sinitski et al., 2016). However, in activities of daily living that involve load carriage (e.g., carrying children or groceries), the load is not always worn on the back. As such, it remains unclear how the placement of an added load (e.g., an anteriorly or posteriorly placed load) affects between-limb asymmetries for prosthesis users. Prolonged asymmetry and reliance on the intact limb are linked to an increased prevalence of overuse injuries, such as intact-knee osteoarthritis (e.g., Gailey et al., 2008). Therefore, a better understanding of how load

* Corresponding author.
E-mail address: stephanie.molitor@utexas.edu (S.L. Molitor).

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placement impacts between-limb symmetry and intact limb reliance for prosthesis users is necessary to help reduce the development of these injuries. The first objective of this study was to determine how the presence and placement of an added load affects between-limb joint work symmetry for individuals wearing unilateral transtibial prostheses in order to better understand intact limb reliance during load carriage conditions. We hypothesized that the presence of an added load would result in increased joint work asymmetry with more reliance on the intact limb. Furthermore, we hypothesized that the placement of the added load (e.g., anterior or posterior load) would influence joint work symmetry.

In order to more closely replicate the function of the transected plantarflexor muscles and reduce intact limb compensations, powered transtibial prostheses have been developed to aid in push-off by generating net positive ankle–foot work (Au and Herr, 2008; Herr and Grabowski, 2011). In contrast, passive ESAR prostheses can only recover a portion of the energy stored earlier in the gait cycle and therefore cannot produce net positive work (Zmitrewicz et al., 2007). Previous studies have determined that reduced ESAR prosthetic ankle–foot push-off work is associated with increased intact knee adduction moment, which is an indirect measure of knee loading and osteoarthritis risk (Grabowski and D'Andrea, 2013; Morgenroth et al., 2011). Furthermore, reduced power output from ESAR prostheses compared to the biological foot–ankle results in increased demand from the more proximal residual joints (Silverman et al., 2008) and increased overall metabolic cost of walking (Herr and Grabowski, 2011). Compared to ESAR prostheses, powered devices have shown improvements in metabolic cost and between-limb joint work symmetry for walking without added loads (e.g., Au and Herr, 2008; Grabowski and D'Andrea, 2013; Herr and Grabowski, 2011). Although previous studies have demonstrated that powered prostheses can also improve metabolic cost during load carriage (Brandt et al., 2016; Hedrick et al., 2019), it remains unclear whether powered prostheses can similarly improve between-limb joint work symmetry during load carriage conditions. Therefore, the second objective of this study was to determine how prosthetic foot type (i.e., passive versus powered) affects joint work symmetry in load carriage conditions. We hypothesized that the powered foot would provide improved joint work symmetry during load carriage due to its ability to generate net positive work.

2. Methods

2.1. Data collection

Kinematic and kinetic data were previously collected at the VA Center for Limb Loss and MoBility from 12 unilateral transtibial prosthesis users (see Ardianuari et al., 2024 for details). Data from one subject used a different marker set from the rest and was not included in the analysis due to significant marker occlusion and compromised model segment definitions (2 female; age: 44 ± 15 years; height: 1.76 ± 0.10 m; mass: 97.0 ± 16.6 kg). Each participant provided informed consent approved by the governing Institutional Review Board. Three-dimensional full-body kinematic data were collected at 120 Hz using a 16-camera motion capture system (Vicon, Oxford, UK) and a set of 66 retroreflective markers following a marker set adapted from the Plugin-Gait and Istituto Ortopedico Rizzoli IORGait (Leardini et al., 2007) models. Three-dimensional ground reaction force (GRF) data were collected at 1200 Hz using five overground force plates (AMTI, Watertown, MA, USA).

Participants completed overground walking trials using both their clinically prescribed passive prosthetic foot and a commercial powered foot (Empower; Ottobock, Duderstadt, GER). Four participants were unable to use the powered foot due to residual limb length constraints. For each prosthetic foot condition, participants completed three sets of walking trials: with a 13.61 kg (30 lb) load worn on the back of their torso, the same load worn on the front, and with no additional load

Table 1
Combinations of prosthetic foot type and load conditions that were tested with their associated abbreviations.

		Load Condition		
		No Load	**Back Load**	**Front Load**
Foot Type	**Prescribed Passive**	PR_NL	PR_BL	PR_FL
	Powered	PW_NL	PW_BL	PW_FL

(Table 1). For trials that included the added load, subjects wore a weighted pack (Ergobaby, Los Angeles, CA, USA) secured to their torso. An additional set of three markers was placed on the load pack to track its movement.

During their initial visit, self-selected walking speed was determined for each participant by the average of four 14 m walk tests while wearing their clinically prescribed passive prosthesis (group mean ± standard deviation: 1.27 ± 0.18 m/s). Participants then familiarized themselves with each prosthetic foot and load combination by practicing walking for at least 15 min or until they felt comfortable in each condition. Prosthesis prescription, alignment and training were supervised by a certified prosthetist. Although there is currently no consensus for how much training time should be allotted before testing a new prosthesis (Wanamaker et al., 2017), previous work evaluating gait changes with a new prosthesis suggests that gait symmetry typically stabilizes within the first twenty minutes of walking (Fiedler and Zhang, 2016). Rest breaks were provided as needed.

After at least one overnight rest period, each participant returned for a second visit, where they completed the walking trials (Table 1). Participants walked across the 10 m force plate-embedded walkway in a straight line with each foot landing on its own force plate. Trials were repeated until the participant's walking speed was within 10% of their previously determined self-selected walking speed, and only strides with a single, complete foot contact on each force plate were included in the analysis.

2.2. Data processing and analysis

A 15-segment (plus one additional segment for the added load) inverse dynamics model was created for each participant in Visual3D (HAS-Motion, Kingston, Ontario). Segment masses, center-of-mass (COM) positions, and inertial properties were estimated based on geometric approximations (de Leva, 1996). On the prosthesis side, the residual shank mass was reduced to 35% of the intact shank and the residual shank COM position was adjusted more proximally to the knee joint by 35% (Smith et al., 2014). Kinematic and GRF data were low-pass filtered at 8 Hz and 15 Hz, respectively, using a fourth-order double-pass Butterworth filter. Individual strides (defined as ipsilateral heel-contact to subsequent ipsilateral heel-contact) were determined based on GRF and kinematic event detection using a custom MATLAB (MathWorks, Natick, MA, USA) script. Sagittal-plane joint angles, moments and powers were computed for the intact ankle, knee and hip, as well as the residual (prosthesis side) knee and hip. Prosthetic ankle angle and moment were not computed due to the lack of an anatomical ankle joint. For both the intact and prosthesis sides, ankle power was calculated as distal shank power to uniformly capture power from the biological ankle–foot complex on the intact side and the distal residual limb and prosthesis for the prosthesis side (Takahashi et al., 2012). Further use of the term *ankle power* refers to distal shank power.

Joint moments and powers were normalized by participant height and weight (including weight of the additional load for relevant trials) following ISB recommendations (Derrick et al., 2020). Refer to the Supplementary Data for joint moment and power curves. Joint powers were integrated over the duration of the gait cycle to calculate net joint work. The difference in net joint work between the intact and prosthesis side (ΔW) was calculated (Eq. (1)) for each joint to quantify the reliance on the intact limb, where $\Delta W > 0$ indicates higher reliance on the intact limb, $\Delta W < 0$ indicates higher reliance on the prosthesis-side limb, and $\Delta W = 0$ indicates perfect symmetry.

$$\Delta W = net\ joint\ work_{intactside} - net\ joint\ work_{prosthesis\ side} \quad (1)$$

2.3. Statistical analyses

A set of linear mixed-effects models determined how the between-limb difference in net joint work (ΔW) changed with prosthetic foot and load conditions. The dependent variable was ΔW and the fixed factors included the prosthetic foot type, load condition, and their interaction. The random factors included participant, the interaction between participant and prosthetic foot type since only a subset of the participants were able to use the powered foot, and a random intercept. A linear mixed-effects model was chosen for its flexibility in accounting for unequal group sizes, such as the passive ($n = 11$) and powered ($n = 7$) foot conditions. Significance was defined as $\alpha = 0.05$. In the case of a significant effect of foot type, a follow-up paired *t*-test was performed to compare ΔW between the passive and powered foot conditions, pooled across all load conditions. In the case of a significant effect of load condition, Bonferroni-adjusted follow-up paired *t*-tests were performed to compare ΔW between the three load conditions, pooled across both foot types. In the case of a significant foot × load interaction, a set of nine Bonferroni-adjusted non-pooled follow-up *t*-tests were used to evaluate all pairwise comparisons of foot and load conditions.

3. Results

3.1. Ankle work symmetry

Ankle ΔW was positive (intact net ankle work > prosthetic net ankle work) for all load conditions, pooled across foot type, as well as for both foot types, pooled across load conditions. Ankle ΔW decreased for the Front Load condition compared to the No Load condition ($p < 0.001$, Cohen's *d*: 0.37), pooled across both foot types (Fig. 1A). There was also a trend of decreased ankle ΔW for the Front Load condition compared to the Back Load condition, but this trend was not significant with the Bonferroni adjustment ($p = 0.017$). In addition, ankle ΔW decreased for the powered foot condition compared to the passive foot ($p < 0.001$, Cohen's *d*: 1.33), pooled across load conditions (Fig. 1B). For both foot types, the intact ankle provided net positive work. When wearing the passive prosthesis, the prosthetic ankle demonstrated net negative work for all load conditions. However, when wearing the powered prosthesis, the prosthetic ankle demonstrated net positive work for all load conditions (Fig. 1B).

3.2. Knee work symmetry

Knee ΔW was positive (intact-side net knee work > prosthesis-side net knee work) for all foot and load conditions. There were no changes in knee ΔW between load or foot conditions (Fig. 2).

3.3. Hip work symmetry

Hip ΔW was negative (prosthesis-side net hip work > intact-side net hip work) for all foot and load conditions (Fig. 3). There was a trend of decreased hip ΔW magnitude for the Front Load condition compared to the No Load condition, pooled across foot types, but this trend was not significant (Fig. 3A). However, there was a significant interaction effect between load and foot condition ($p = 0.029$) and follow-up paired *t*-tests determined that the magnitude of hip ΔW decreased for the Front Load condition compared to the No Load condition when wearing the passive foot ($p = 0.005$, Cohen's *d*: 0.95) (Fig. 3B).

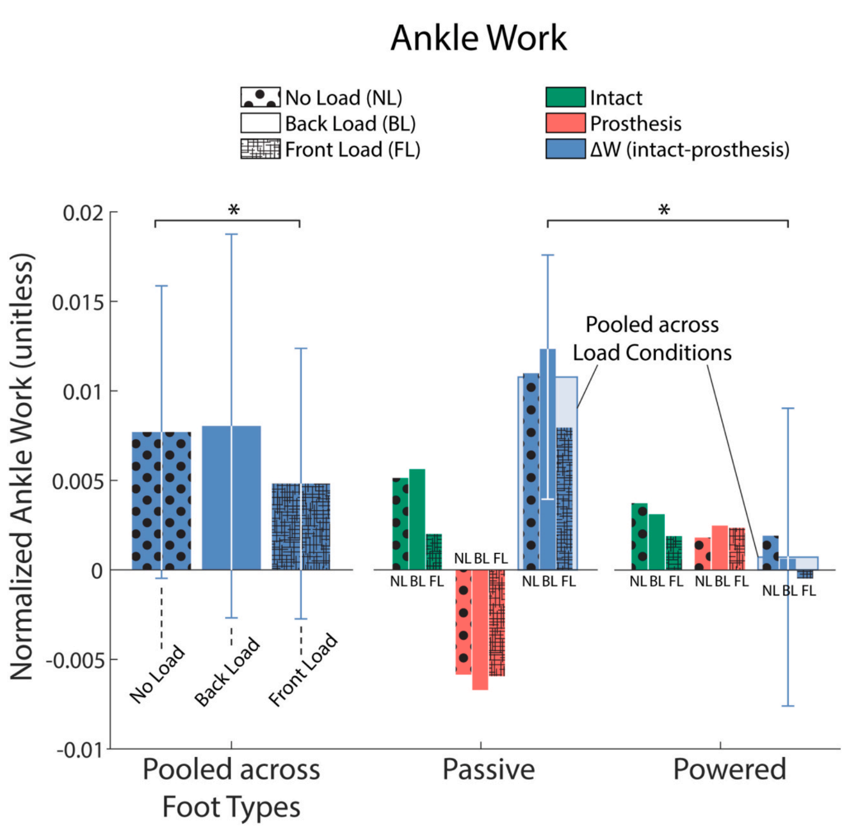

Fig. 1. Ankle ΔW pooled across foot types (left pane) decreased for the Front Load (hashed) compared to the No Load (dotted) condition. Ankle ΔW pooled across load conditions (wide bars in the center and right panes) decreased for the powered compared to the passive foot condition. Ankle ΔW was calculated as the difference between the intact and prosthesis-side net ankle work for each condition.

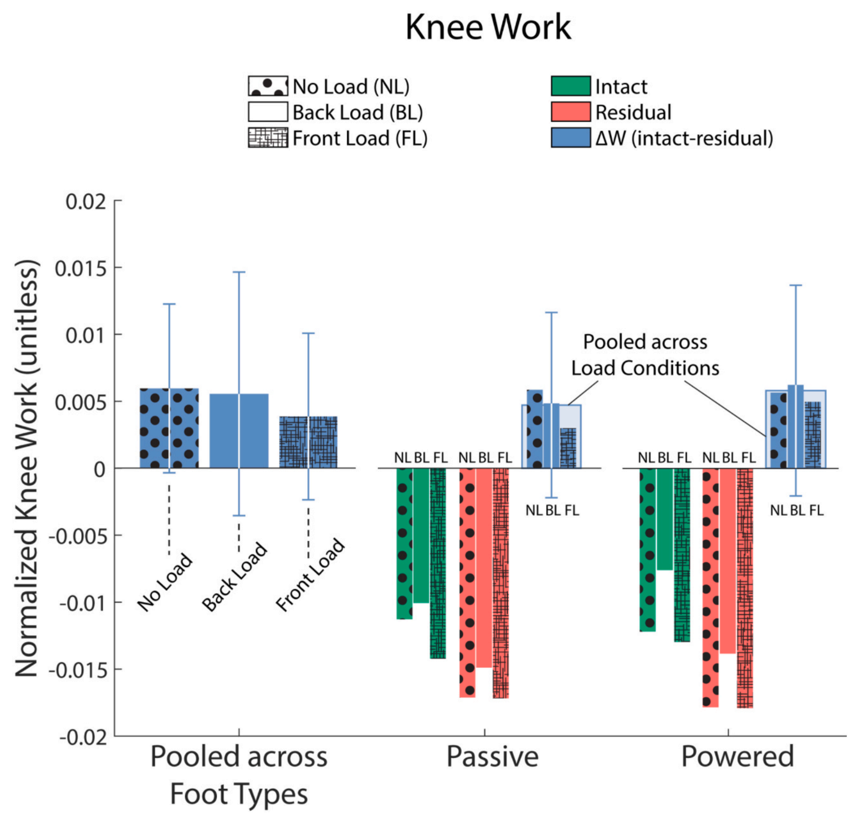

Fig. 2. Knee ΔW pooled across foot types (left pane) saw no difference between the No Load (dotted), Back Load (solid), and Front Load (hashed) conditions. Knee ΔW pooled across load conditions (wide bars in the center and right panes) saw no difference between the powered and passive foot conditions. Knee ΔW was calculated as the difference between the intact and residual net knee work for each condition.

4. Discussion

The objective of this study was to determine how the presence and placement of an added load affects lower-limb joint work symmetry for individuals wearing either a passive or powered prosthesis. We hypothesized that the presence and placement of an added load would affect joint work symmetry, expecting more asymmetry during the loaded conditions. Contrary to this hypothesis, ankle work symmetry improved during the Front Load condition across prosthetic foot types, and hip work symmetry improved during the Front Load condition while wearing the passive prosthesis. Our hypothesis that the load placement would affect joint work symmetry was supported given that the Front Load improved joint work symmetry compared to the No Load condition but the Back Load condition demonstrated no changes in joint work symmetry. Finally, we hypothesized that the powered foot would offer improved joint work symmetry, which was partially supported as the powered foot did improve ankle work symmetry for all load conditions.

4.1. Changes in joint work symmetry due to load carriage condition

Ankle ΔW was positive, indicating greater reliance on the intact ankle, for all load conditions (pooled across foot types) and for both foot types (pooled across load conditions). Conversely, hip ΔW was negative, indicating greater reliance on the prosthesis-side hip, across all load and foot conditions. These findings reflect previously described strategies to compensate for the loss of the ankle plantarflexors, including increased prosthesis-side hip power (Sadeghi et al., 2001) and work (Jeffers and Grabowski, 2017; Silverman et al., 2008) as well as increased intact ankle work (Jeffers and Grabowski, 2017; Seroussi et al., 1996; Silverman et al., 2008).

Work symmetry improved for both the ankle (decreased ankle ΔW) and the hip (decreased magnitude of hip ΔW) for the Front Load condition compared to the No Load condition. Surprisingly, the improvement in ankle work symmetry resulted from a reduction in intact ankle work, with few changes in prosthesis-side ankle work. In contrast to the changes observed at the ankle, net hip work for both the intact and prosthesis side increased in the Front Load condition, with a larger increase in the intact hip work resulting in improved symmetry. These results suggest that in the Front Load condition, there is a tradeoff between decreased intact ankle demand but increased intact hip demand, which may be indicative of a strategy to counteract the destabilizing moment caused by the front load.

When carrying a front load, the added load causes the total COM to shift anteriorly, resulting in a destabilizing moment due to gravity, and increased forward angular momentum. The ankle plantarflexors likewise contribute to forward angular momentum during walking whereas the hip extensors (e.g., hamstrings) contribute to backward angular momentum (Neptune and McGowan, 2011). Decreasing intact ankle work and increasing intact hip work during the Front Load condition may serve as a mechanism to counteract the forward angular momentum generated by the additional load, maintain sagittal plane balance control and mitigate fall risk. This finding is consistent with previous studies which report increased intact-side hip extensor (hamstrings, gluteus maximus) metabolic cost (Lefranc et al., 2025) as well as increased intact-side hamstring contribution to forward propulsion (Lefranc et al., 2024) and backward whole-body angular momentum (Templin et al., 2021) during anterior load carriage. These changes also align with previous observations of more cautious gait demonstrated by healthy young adults during anterior load carriage (Cain et al., 2016; Perry et al., 2010). Based on previous modeling work, although Front Load carriage generally increased overall metabolic cost compared to No Load or Back Load conditions, lower intact axial knee joint loads were observed for the Front Load condition compared to the Back Load (Lefranc et al., 2025), which may be due to the proximal redistribution

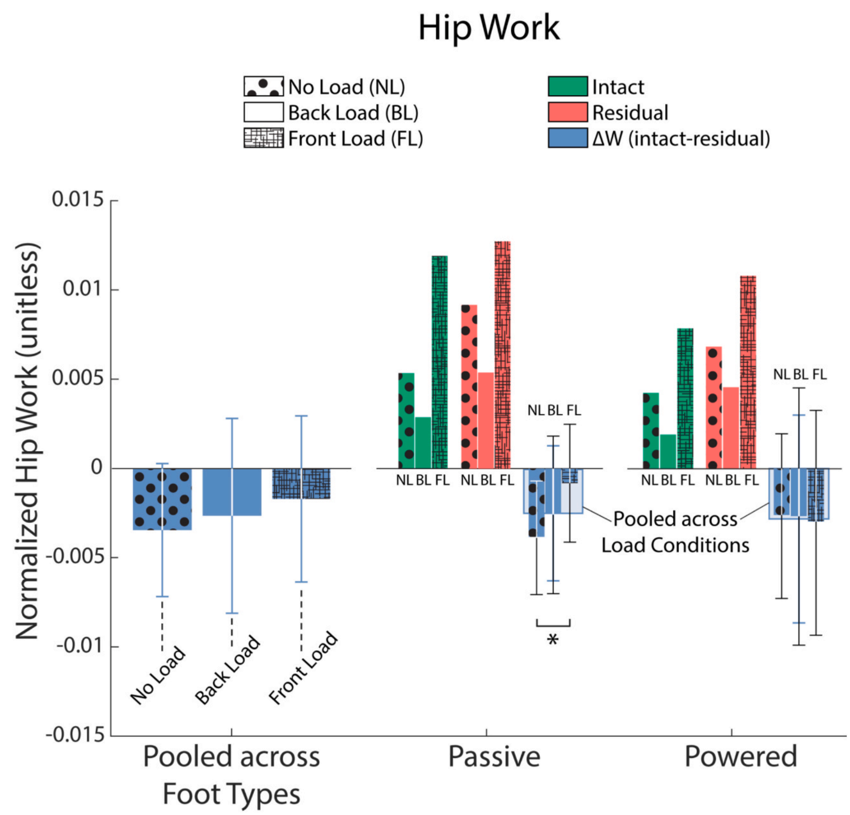

Fig. 3. Hip ΔW pooled across foot types (left pane) saw no difference between the No Load (dotted), Back Load (solid), and Front Load (hashed) conditions. Hip ΔW pooled across load conditions (wide bars in the center and right panes) saw no difference between the powered and passive foot conditions. There was a significant interaction effect between load and foot condition, in which the magnitude of hip ΔW decreased for the Front Load condition compared to the No Load condition only when wearing the passive foot. Hip ΔW was calculated as the difference between the intact and residual net hip work for each condition.

of intact-side net joint work observed in the current study. As such, it may be beneficial for individuals with transtibial prostheses to use an anterior load carriage strategy to minimize intact knee joint loading and osteoarthritis risk.

Unlike the changes in ankle and hip work symmetry demonstrated in the Front Load condition, no changes in joint work symmetry were demonstrated for the Back Load condition. Observed trends in individual leg net joint work suggests that during the Back Load condition, the magnitude of both intact and residual net knee and hip work decreased compared to the No Load condition. However, since net joint work for both legs responded similarly, joint work symmetry remained unchanged.

4.2. Changes in joint work symmetry due to prosthetic foot type

Across all load conditions, ankle work symmetry improved when wearing the powered foot compared to the passive foot, with a 93% reduction in ΔW (average passive ΔW: 0.010; average powered ΔW: 0.0007), largely attributed to the change in net prosthetic ankle work from negative to positive when switching from the passive to the powered foot. This finding is consistent with previous work that found improved net leg work symmetry with a powered prosthetic foot during incline walking (Montgomery and Grabowski, 2018). The ability of the powered foot to generate net positive ankle work more closely replicates the function of the biological ankle plantarflexors in generating forward propulsion at push-off (Neptune et al., 2001), thereby reducing the need for intact ankle compensations typically reported in challenging walking conditions (i.e., faster walking speeds) (Silverman et al., 2008). A post-hoc comparison of prosthetic positive ankle push-off work confirmed that the powered prosthesis did provide increased positive work at push-off compared to the passive device (passive: 0.134 J/kg, powered: 0.190 J/kg, $p < 0.001$). Together, the improved ankle work symmetry and the increased prosthetic ankle push-off work demonstrated while wearing the powered prosthesis suggest that the powered device can reduce intact ankle compensations and may reduce risk for intact-side overuse injuries (Morgenroth et al., 2011) during load carriage tasks. Previous modeling work also found that axial intact knee joint loading, a biomarker for osteoarthritis risk, was reduced when using a powered prosthesis compared to a passive ESAR device (Lefranc et al., 2025), further encouraging the use of powered prostheses during load carriage.

Surprisingly, we did not see an improvement in hip joint work symmetry with the powered prosthesis. Observed trends in net hip work for each leg demonstrate that the powered prosthesis did reduce the net hip work produced by the intact and residual legs. However, since net hip work changed for both legs similarly, hip joint work symmetry was unaffected by the powered foot. There were also no observable changes in net knee work, or resulting knee work symmetry, with the use of the powered prosthesis. Unlike the ankle and hip which produce net positive work, the knee demonstrated net negative work, indicating overall energy absorption. Furthermore, segmental power analyses have revealed that the knee muscles (e.g., rectus femoris and vasti) act to transfer energy between the leg and trunk throughout the gait cycle (Neptune et al., 2004). Given the role of the knee musculature in power transfer rather than power production, it is not surprising that there were no observed changes in net knee work despite the increased ankle work with the powered prosthesis.

5. Limitations

A potential limitation of this study was the relatively small sample size, given that that only seven participants were able to use the powered prosthetic foot due to residual limb length constraints. Although the number of participants for this study is consistent with previous similar prosthetic component studies (e.g., Clites et al., 2021; McDonald et al., 2021; Montgomery and Grabowski, 2018), unbalanced group sizes between the prosthetic foot conditions may reduce statistical power. As such, results for the passive versus powered foot analyses may not be as generalizable to a larger population. Furthermore, a limited number of strides were evaluated in each condition, as the force plate-embedded walkway only accommodates five continuous strides per trial. Multiple trials were recorded for each condition; however, the omission of strides without a single, complete foot contact on each force plate resulted in most participants having only a few successful strides on each foot for each condition. Moreover, our trial inclusion criteria required walking speed to be within ± 10% of each participant's self-selected speed to reduce the impact of walking speed variability as a confounding variable. However, it is possible that small stride-to-stride variability or differences in speed preferences between participants may still obscure work outcomes. Further investigation to determine a balance between acceptable walking speed variability and the number of strides required to achieve appropriate statistical power is warranted to better inform future experimental designs.

Another potential limitation was that this study focused on net joint work rather than the total work done at each joint. Net joint work may underestimate total energy expenditure due to cancellations of positive and negative work. However, comparing total work done at each joint may obscure directional asymmetries in the case of compensations during which one limb generates positive work and the other limb negative work of similar magnitude (e.g., as seen by the intact and passive prosthetic ankle). Finally, this work evaluated changes in between-limb symmetry for various foot-load combinations. Although useful for quantifying reliance on the intact limb, symmetry analyses do not capture changes that affect both limbs similarly. For example, the magnitude of both intact and residual net knee and hip work decreased during the Back Load condition compared to the No Load condition. However, since the net joint work for both legs responded to the load similarly, joint work symmetry remained unchanged.

6. Conclusion and recommendations for future work

This study evaluated intact limb reliance for different load carriage conditions while wearing either a passive or powered prosthesis. We identified a tradeoff between intact ankle and hip demand when carrying a front load, and found that the powered foot improved between-limb ankle work symmetry across all load conditions. These results provide insight into how compensatory strategies change in response to altered task demands and encourage the use of powered prostheses to reduce intact limb reliance during load carriage. Future studies should expand on this work and investigate individual leg responses to different foot-load combinations in order to compare between symmetric and asymmetric compensations. Furthermore, future work should evaluate total work and energy expenditure for different load carriage conditions to guide individuals with lower-limb amputations on recommended load carriage strategies. Finally, future work should explore a range of prosthetic ankle–foot stiffness profiles during various load carriage conditions to explore potential benefits of stiffness-adaptive prostheses.

CRediT authorship contribution statement

Stephanie L. Molitor: Writing – review & editing, Writing – original draft, Visualization, Validation, Project administration, Methodology, Formal analysis, Data curation, Conceptualization. **Krista M. Cyr:** Writing – review & editing, Validation, Investigation, Data curation. **Glenn K. Klute:** Writing – review & editing, Validation, Supervision, Resources, Project administration, Conceptualization. **Richard R. Neptune:** Writing – review & editing, Visualization, Validation, Supervision, Resources, Project administration, Methodology, Conceptualization.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2025.113047.

References

Ardianuari, S., Cyr, K.M., Neptune, R.R., Klute, G.K., 2024. Should individuals with unilateral transtibial amputation carry a load on their intact or prosthetic side? J. Biomech. 177, 112385. https://doi.org/10.1016/j.jbiomech.2024.112385.

Au, S.K., Herr, H.M., 2008. Powered ankle-foot prosthesis. IEEE Robotics & Automation Magazine 15, 52–59. https://doi.org/10.1109/MRA.2008.927697.

Bayram, H.A., Bayram, M.B., 2018. Dynamic Functional Stiffness Index of the Ankle Joint during Daily living. J. Foot Ankle Surg. 57, 668–674. https://doi.org/10.1053/j.jfas.2017.11.034.

Brandt, A., Liu, M., Huang, H.H., 2016. Does the impedance of above-knee powered prostheses need to be adjusted for load-carrying conditions?. In: In: 2016 38th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC). Presented at the 2016 38th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC), pp. 5075–5078. https://doi.org/10.1109/EMBC.2016.7591868.

Cain, S.M., McGinnis, R.S., Davidson, S.P., Vitali, R.V., Perkins, N.C., McLean, S.G., 2016. Quantifying performance and effects of load carriage during a challenging balancing task using an array of wireless inertial sensors. Gait Posture 43, 65–69. https://doi.org/10.1016/j.gaitpost.2015.10.022.

Clites, T.R., Shepherd, M.K., Ingraham, K.A., Wontorcik, L., Rouse, E.J., 2021. Understanding patient preference in prosthetic ankle stiffness. J. Neuroeng. Rehabil. 18, 128. https://doi.org/10.1186/s12984-021-00916-1.

de Leva, P., 1996. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. J. Biomech. 29, 1223–1230. https://doi.org/10.1016/0021-9290(95)00178-6.

Derrick, T.R., van den Bogert, A.J., Cereatti, A., Dumas, R., Fantozzi, S., Leardini, A., 2020. ISB recommendations on the reporting of intersegmental forces and moments during human motion analysis. J. Biomech. 99, 109533. https://doi.org/10.1016/j.jbiomech.2019.109533.

Doyle, S.S., Lemaire, E.D., Besemann, M., Dudek, N.L., 2014. Changes to level ground transtibial amputee gait with a weighted backpack. Clin. Biomech. (Bristol, Avon) 29, 149–154. https://doi.org/10.1016/j.clinbiomech.2013.11.019.

Fiedler, G., Zhang, X., 2016. Quantifying accommodation to prosthesis interventions in persons with lower limb loss. Gait Posture 50, 14–16. https://doi.org/10.1016/j.gaitpost.2016.08.016.

Frigo, C., Crenna, P., Jensen, L.M., 1996. Moment-angle relationship at lower limb joints during human walking at different velocities. J. Electromyogr. Kinesiol. 6, 177–190. https://doi.org/10.1016/1050-6411(96)00030-2.

Gailey, R., Allen, K., Castles, J., Kucharik, J., Roeder, M., 2008. Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use. J. Rehabil. Res. Dev. 45, 15–29. https://doi.org/10.1682/jrrd.2006.11.0147.

Grabowski, A.M., D'Andrea, S., 2013. Effects of a powered ankle-foot prosthesis on kinetic loading of the unaffected leg during level-ground walking. J NeuroEngineering Rehabil 10, 49. https://doi.org/10.1186/1743-0003-10-49.

Hedrick, E.A., Malcolm, P., Wilken, J.M., Takahashi, K.Z., 2019. The effects of ankle stiffness on mechanics and energetics of walking with added loads: a prosthetic emulator study. J. Neuroeng. Rehabil. 16, 148. https://doi.org/10.1186/s12984-019-0621-x.

Herr, H.M., Grabowski, A.M., 2011. Bionic ankle–foot prosthesis normalizes walking gait for persons with leg amputation. Proc. R. Soc. B Biol. Sci. 279, 457–464. https://doi.org/10.1098/rspb.2011.1194.

Jeffers, J.R., Grabowski, A.M., 2017. Individual Leg and Joint Work during Sloped walking for people with a Transtibial Amputation using Passive and Powered Prostheses. Front. Robot. AI 4. https://doi.org/10.3389/frobt.2017.00072.

Kern, A.M., Papachatzis, N., Patterson, J.M., Bruening, D.A., Takahashi, K.Z., 2019. Ankle and midtarsal joint quasi-stiffness during walking with added mass. PeerJ 7, e7487.

Leardini, A., Sawacha, Z., Paolini, G., Ingrosso, S., Nativo, R., Benedetti, M.G., 2007. A new anatomically based protocol for gait analysis in children. Gait Posture 26, 560–571. https://doi.org/10.1016/j.gaitpost.2006.12.018.

Lefranc, A.S., Klute, G.K., Neptune, R.R., 2025. The influence of load carriage and prosthetic foot type on measures of biomechanical demand. Journal of Biomechanics 193, 112992.

Lefranc, A.S., Klute, G.K., Neptune, R.R., 2024. The influence of load carriage and prosthetic foot type on individual muscle and prosthetic foot contributions to body support and propulsion. J. Biomech. 177, 112379. https://doi.org/10.1016/j.jbiomech.2024.112379.

McDonald, K.A., Teater, R.H., Cruz, J.P., Kerr, J.T., Bastas, G., Zelik, K.E., 2021. Adding a toe joint to a prosthesis: walking biomechanics, energetics, and preference of individuals with unilateral below-knee limb loss. Sci. Rep. 11, 1924. https://doi.org/10.1038/s41598-021-81565-1.

Montgomery, J.R., Grabowski, A.M., 2018. Use of a powered ankle–foot prosthesis reduces the metabolic cost of uphill walking and improves leg work symmetry in people with transtibial amputations. J. R. Soc. Interface 15, 20180442. https://doi.org/10.1098/rsif.2018.0442.

Morgenroth, D.C., Segal, A.D., Zelik, K.E., Czerniecki, J.M., Klute, G.K., Adamczyk, P.G., Orendurff, M.S., Hahn, M.E., Collins, S.H., Kuo, A.D., 2011. The effect of prosthetic foot push-off on mechanical loading associated with knee osteoarthritis in lower extremity amputees. Gait Posture 34, 502–507. https://doi.org/10.1016/j.gaitpost.2011.07.001.

Neptune, R.R., Kautz, S.A., Zajac, F.E., 2001. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. J. Biomech. 34, 1387–1398. https://doi.org/10.1016/S0021-9290(01)00105-1.

Neptune, R.R., McGowan, C.P., 2011. Muscle contributions to whole-body sagittal plane angular momentum during walking. J. Biomech. 44, 6–12. https://doi.org/10.1016/j.jbiomech.2010.08.015.

Neptune, R.R., Zajac, F.E., Kautz, S.A., 2004. Muscle force redistributes segmental power for body progression during walking. Gait Posture 19, 194–205. https://doi.org/10.1016/S0966-6362(03)00062-6.

Perry, C.J., Kiriella, J.B., Hawkins, K.M., Shanahan, C.J., Moore, A.E., Gage, W.H., 2010. The effects of anterior load carriage on lower limb gait parameters during obstacle clearance. Gait Posture 32, 57–61. https://doi.org/10.1016/j.gaitpost.2010.03.003.

Sadeghi, H., Allard, P., Duhaime, M., 2001. Muscle Power Compensatory Mechanisms in Below-knee Amputee Gait. Am. J. Phys. Med. Rehabil. 80, 25.

Schnall, B.L., Hendershot, B.D., Bell, J.C., MSE, Wolf, E.J., 2014. Kinematic analysis of males with transtibial amputation carrying military loads. J Rehabil Res Dev 51, 1505–1514. 10.1682/JRRD.2014.01.0022.

Seroussi, R.E., Gitter, A., Czerniecki, J.M., Weaver, K., 1996. Mechanical work adaptations of above-knee amputee ambulation. Arch. Phys. Med. Rehabil. 77, 1209–1214. https://doi.org/10.1016/S0003-9993(96)90151-3.

Silverman, A.K., Fey, N.P., Portillo, A., Walden, J.G., Bosker, G., Neptune, R.R., 2008. Compensatory mechanisms in below-knee amputee gait in response to increasing steady-state walking speeds. Gait Posture 28, 602–609. https://doi.org/10.1016/j.gaitpost.2008.04.005.

Sinitski, E.H., Herbert-Copley, A.G., Lemaire, E.D., Doyle, S.S., Besemann, M., Dudek, N. L., 2016. Center of pressure and total force analyses for amputees walking with a backpack load over four surfaces. Appl. Ergon. 52, 169–176. https://doi.org/10.1016/j.apergo.2015.07.014.

Smith, J.D., Ferris, A.E., Heise, G.D., Hinrichs, R.N., Martin, P.E., 2014. Oscillation and Reaction Board Techniques for estimating Inertial Properties of a Below-knee Prosthesis. J. Vis. Exp. 50977. https://doi.org/10.3791/50977.

Takahashi, K.Z., Kepple, T.M., Stanhope, S.J., 2012. A unified deformable (UD) segment model for quantifying total power of anatomical and prosthetic below-knee structures during stance in gait. J. Biomech. 45, 2662–2667. https://doi.org/10.1016/j.jbiomech.2012.08.017.

Templin, T.N., Klute, G.K., Neptune, R.R., 2021. The Influence of load Carriage on knee Joint Loading and Metabolic cost on walking with Lower-Limb Amputation: a Preliminary Modeling Study. J Prosthet Orthot 33, 118–124. https://doi.org/10.1097/JPO.0000000000000359.

Wanamaker, A.B., Andridge, R.R., Chaudhari, A.M., 2017. When to biomechanically examine a lower-limb amputee: a systematic review of accommodation times. Prosthet. Orthot. Int. 41, 431–445. https://doi.org/10.1177/0309364616682385.

Zmitrewicz, R.J., Neptune, R.R., Sasaki, K., 2007. Mechanical energetic contributions from individual muscles and elastic prosthetic feet during symmetric unilateral transtibial amputee walking: a theoretical study. J. Biomech. 40, 1824–1831. https://doi.org/10.1016/j.jbiomech.2006.07.009.