Clinical Biomechanics 122 (2025) 106440

Contents lists available at ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

Influence of prosthetic foot selection on walking performance during various load carriage conditions

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A R T I C L E I N F O

Keywords:
Lower limb amputation
Transtibial amputee
Gait
Biomechanics

A B S T R A C T

Background: Ambulatory individuals with lower limb amputations often face challenges with body support, body propulsion, and balance control. Carrying an infant, toddler, backpack, or other load can exacerbate these challenges and highlights the importance of prescribing the most suitable prosthetic foot. The aim of this study was to examine the influence of five different prosthetic feet on walking performance during various load carriage conditions.
Methods: Biomechanical data were collected from twelve participants wearing five different prosthetic feet (four passive, one powered) while walking with no added load and carrying a load of 13.6 kg in four different positions: posterior, anterior, prosthetic side, and intact side.
Findings: Based on our study population, a powered-ankle-foot offers additional body support when a load is carried posteriorly. If additional forward propulsion is needed while carrying a load anteriorly, a heel wedge is better than a stiffer foot. For individuals who may need additional sagittal plane balance control, no study foot was advantageous regardless of how the load was carried. For those who need additional frontal plane balance control during posterior load carriage, a heel wedge is better than a stiffer or powered foot. Lastly, the standard-of-care, heel wedge, and dual keel feet provided more frontal plane balance control than a powered foot when a load was carried anteriorly.
Interpretation: For individuals with lower limb amputation who carry loads, consideration of their preferred load carrying method may help select an appropriate prosthetic foot for body support, propulsion, and balance control.

1. Introduction

Carrying loads is an essential practice for everyday life, whether it's for an occupation, recreation, or necessity. For able-bodied individuals, lower limb muscles can activate and respond to added loads to enable seamless continuation of biomechanical functions. For example, the ankle plantar flexor muscles have been shown to be primary contributors to body support, forward propulsion, and dynamic balance control (Liu et al., 2006; Neptune et al., 2001; Neptune and McGowan, 2016). For individuals with lower limb amputation who no longer have this muscle group, sudden changes to weight-bearing loads (e.g., carrying infants, toddlers, backpacks, or other loads) can negatively impact walking performance since the properties of most prosthetic limbs do not dynamically change to suit varying load conditions. Frequent load carriage can also lead to injuries such as foot blisters, pain/inflammation of lower limb joints, and stress fractures (Knapik et al., 2004). Individuals with vulnerable limbs due to trauma or dysvascular etiology are hypothesized to be at even greater risk for these types of injuries while carrying loads.

There are several different strategies for carrying loads, the most common including posteriorly in a backpack, anteriorly in a sling or with arms, or asymmetrically with arms on either side (Coleman et al., 2015; Knapik et al., 2004). Load carriage magnitude and placement have been shown to impact postural stability and contribute to fall risk (Martin et al., 2023). Carrying loads in general can also cause an increase in oxygen consumption (Fallowfield et al., 2012), but loads

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https://doi.org/10.1016/j.clinbiomech.2025.106440
Received 23 October 2024; Accepted 17 January 2025
Available online 19 January 2025
0268-0033/

carried posteriorly are less metabolically costly (Abe et al., 2004). Despite higher metabolic cost, American children are most often carried on the front (Bouterse and Wall-Scheffler, 2018). While many recreational activities and occupations (including parenthood) require load carriage, there is little evidence to guide the prosthetic foot prescription practice for lower limb amputees to improve their mobility and balance during various load carriage conditions.

Current literature on individuals with lower limb amputation carrying loads has focused primarily on posterior loads or weighted vests (Brandt et al., 2017; Doyle et al., 2014; Koehler-McNicholas et al., 2018; Schnall et al., 2014, 2020; Sinitski et al., 2016). Some of these studies have shown that lower limb amputees respond to added load with increased deformation of the prosthetic ankle-foot in late stance and an increased reliance on their intact limb when walking over level ground (Doyle et al., 2014; Schnall et al., 2014). This increased deformation of prosthetic feet is in stark contrast to the unaltered kinematics of the ankle-foot motion in non-amputees (Birrell and Haslam, 2009), and suggests a significant short-coming in standard prosthetic devices during added load conditions (Koehler-McNicholas et al., 2018). A prosthetic foot with greater forefoot stiffness has been shown to significantly decrease ankle dorsiflexion (Klodd et al., 2010), which could be desirable during added load conditions. However, people carry loads in a variety of ways and there are many different prosthetic feet available for prescription. Current literature is lacking in understanding how different prosthetic feet may influence walking performance during various load carriage methods (e.g., posterior, anterior, prosthetic side, and intact side load carriage). Our study aims to provide insight on walking performance, specifically body support, forward propulsion and balance control, and to provide prosthesis prescription guidance for clinicians whose patients frequently carry infants, toddlers, backpacks, or other loads.

Clinicians have few options to facilitate mobility of individuals with lower limb amputation who routinely carry loads. Lower cost options include the prescription of a stiffer category foot, the addition of a heel wedge to stiffen the heel, or the prescription of a dual-keel prosthetic foot (Thrive, Freedom Innovations, Irvine, CA, USA). The dual-keel prosthetic foot has a full-length primary keel and a truncated secondary keel. As additional loads are applied, the primary keel engages the secondary keel such that the prosthetic foot becomes stiffer. A more expensive prescription option is a commercially available powered ankle-foot (Empower, Ottobock, Austin, TX, USA). The clinical value of a powered ankle-foot prosthesis is an open question with mixed results. Some research has suggested certain benefits such as reduced metabolic cost, increased ankle power, and potential comorbidity reduction (Grabowski and D'Andrea, 2013; Herr and Grabowski, 2012; Russell Esposito et al., 2016), while other studies report limited to no benefit (Russell Esposito and Wilken, 2014), including increased compensatory strategies at proximal joints (Ferris et al., 2012). It may be that a specific gait training and rehabilitation program is needed to unlock benefits for all powered foot users (Kannenberg et al., 2021). The efficiency of the powered ankle-foot device during load carriage has yet to be demonstrated, and there is little evidence to support prescription of any of these prosthetic options while carrying a load.

The purpose of this research was to provide evidence to guide prosthesis prescription for individuals with a lower limb amputation who frequently carry infants, toddlers, or other loads. The specific aim was to examine the influence of five different prosthetic feet on walking performance (e.g., body support, forward propulsion and balance control) during various load carriage conditions (no added load and with a load carried posteriorly, anteriorly, on the prosthetic side, and on the intact side). We hypothesized that a study prosthesis exists that: 1) maximizes vertical ground reaction force (GRF) impulse (a measure of body support), 2) maximizes anterior GRF impulse (a measure of body forward propulsion), 3) minimizes the peak-to-peak range of sagittal plane whole-body angular momentum (WBAM) (a measure of anterior/posterior balance control), and 4) minimizes the peak-to-peak range of frontal plane WBAM (a measure of medio-lateral balance control) within each load carriage condition.

2. Methods

2.1. Subjects

Twelve individuals with unilateral transtibial amputation (Table 1) provided informed consent to participate in this institutional review board (VA Puget Sound Medical Center, Seattle, WA, USA) approved protocol. All considered themselves moderately active community ambulators, had been fit with a prosthesis and used it for at least six months, and did not have any other disorders, pain, or injury that would interfere with their gait.

2.2. Instrumentation

Each subject was fit with a standard-of-care, clinically prescribed prosthetic foot (CP) for their specific size and category (Sierra, Freedom Innovations, Irvine, CA, USA), this same prosthetic foot with a heel-stiffening wedge (HW), this same prosthetic foot but one category stiffer (SF), a dual-keel prosthetic foot (DK) (Thrive, Freedom Innovations, Irvine, CA, USA) intended for load carriage applications, and a powered ankle-foot (PF) (Empower, Ottobock, Austin, TX, USA) purported to adapt to changing loads. Due to the build height restrictions and residual limb lengths, only seven subjects were able to be fit with the PF. A 13.6 kg (30 lb) load was created using sand inside a cylindrical pack, which was then placed inside a carrier (Ergobaby, Los Angeles, CA, USA) for the subjects to wear. This weight was chosen to simulate carriage of children, backpacks, or other loads during daily or occupational tasks (Fryar et al., 2021; Gittleman et al., 2016). Subjects carried the load in four different positions: posterior, anterior, prosthetic side, and intact side (Fig. 1). Subjects also performed a no-load condition.

A 16-camera motion capture system (Vantage V8, Vicon Motion Systems, Oxford, UK) recorded marker trajectories at 120 Hz and force plate data at 1200 Hz. All subjects were provided with tight fitting spandex shorts and shirts to wear during data collection. The same researcher placed 14 mm reflective tracking markers on each subject using Vicon's standard Plug-in-Gait marker set, with additional markers placed bilaterally on the medial elbow, medial knee epicondyle, medial malleolus, tibial tuberosity, fibular head, and first and fifth metatarsal heads. Clusters of four markers were also placed bilaterally on the upper arms and thighs. The markers on the prosthetic limb mirrored the intact limb. Lastly, three markers were placed to define the load segment (two on top, one on bottom).

2.3. Protocol

At the first study visit, each subject walked at their own pace 14 m down a straight hallway while wearing their as-prescribed prosthesis. An average of four trials were used to measure their self-selected walking speed (SSWS). Subject height, body mass, and anthropometrics were also measured. The study prosthetic feet were then fit in randomized order and aligned by a licensed and certified prosthetist using standard clinical procedures. Each subject used their own prescribed socket and suspension system with all study feet. Although there is no consensus on how much accommodation time is necessary for acclimation to a new prosthetic foot (Wanamaker et al., 2017), we allowed each subject at least 15 min to walk with and without the load in each position to acclimate to each study foot. Rest breaks were provided as needed.

After at least one overnight rest period, each subject came back for a second visit. For this visit, the five load carriage conditions were randomized for each subject. Within each load carriage condition, foot order was also randomized. A standing static trial was performed for each load condition. Each subject walked overground at their SSWS

Table 1
Subject demographics. Abbreviations: NA (not applicable), SD (standard deviation), SSWS (self-selected walking speed).

Subject	Gender	Age (yrs)	Mass (kg)	Height (m)	Amputation Side	Etiology	Residual Limb Length (cm)	Time Since Amputation (yrs)	SSWS (m/s)
01	Male	59	83.5	1.674	Left	Trauma	13	15	1.16
02	Male	40	102.3	1.805	Left	Trauma	15	10	1.67
03	Male	60	111.9	1.800	Right	Trauma	15	3	1.30
04	Female	58	96.6	1.690	Right	Congenital	19	5	1.12
05	Male	39	105.7	1.712	Right	Trauma	15	12	1.43
06	Female	25	52.6	1.564	Right	Congenital	16	24	1.37
07	Male	43	107.0	1.820	Left	Infection	22	1	1.32
08	Male	31	98.5	1.830	Right	Trauma	–	11	1.26
09	Male	36	110.4	1.856	Right	Trauma	20	15	1.20
10	Male	29	83.9	1.644	Left	Trauma	13	3	1.16
11	Male	53	106.6	1.859	Left	Diabetic	17	7	0.92
12	Male	75	93.6	1.831	Left	Trauma	17	55	1.17
Mean	NA	**46**	**96.1**	**1.757**	NA	NA	**17**	**13**	**1.26**
SD	NA	**15**	**15.9**	**0.092**	NA	NA	**3**	**14**	**0.18**

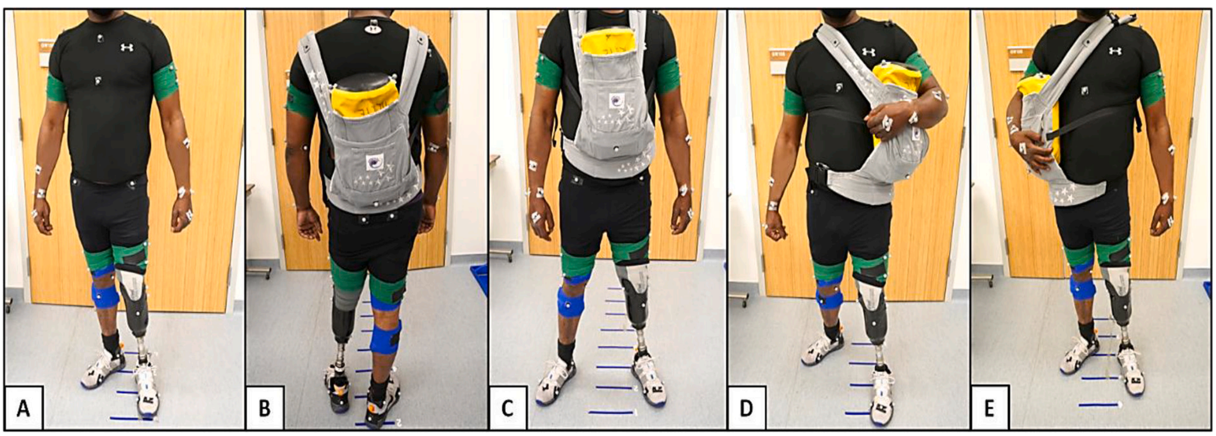

Fig. 1. A) No Load, B) Posterior Load, C) Anterior Load, D) Prosthetic Side Load, E) Intact Side Load.

across five embedded force plates (AMTI, Watertown, MA, USA) for a minimum of two repeated trials per load condition and foot type. Rest breaks were provided as needed.

2.4. Data analysis

The raw marker trajectory data was filtered in Vicon Nexus software using a Woltring filter with a mean-square-error of 10, which is based on a fifth-order spline-interpolating function (Woltring, 1986). Using Visual 3D software (C-Motion Inc., Boyds, MD, USA), a 15-segment whole body model (head, torso, Visual 3D Composite pelvis, and bilateral upper arm, forearm, hand, thigh, shank, and foot; plus a load segment when applicable) was created based on each standing static trial. Each segment's mass was estimated as a percentage of whole-body mass (Dempster, 1955), and the inertial properties and center of mass (CoM) positions were based on geometric approximations calculated in Visual 3D. The prosthetic shank mass was reduced to 35 % of the intact shank, and the prosthetic CoM location was moved 35 % closer to the knee joint (Smith et al., 2014). GRF data was low-pass filtered using a fourth-order Butterworth filter with a cutoff frequency of 25 Hz. Heel strike and toe off events were automatically detected using a combination method based on force plate loading threshold of 25 N and kinematic pattern recognition. These events were also inspected visually and corrected if needed. WBAM was calculated for the sagittal plane and frontal plane as the summation of the angular momentum of each segment about the total body CoM and was normalized by body mass (kg) including the mass of the load when applicable, body height (m), and $\sqrt{\text{gravity*body height}}$ (m/s) (Vistamehr et al., 2019). Normalized WBAM was then separated into prosthetic side gait cycle (prosthetic side heel strike to heel strike), and the range of WBAM was calculated as the peak-to-peak value for both sagittal and frontal planes separately.

GRF data was further processed using Matlab software (The MathWorks Inc., Natick, MA, USA). All GRFs were normalized by subject body mass (kg), including the mass of the load when applicable. Vertical GRF impulse was then calculated as the time-integral of the normalized vertical GRF across stance for the prosthetic limb, and anterior GRF impulse was calculated as the time-integral of the curve above zero for the normalized anteroposterior GRF across stance for the prosthetic limb.

2.5. Statistics

Linear mixed effects regression was used to assess differences in biomechanical outcomes (the dependent variable) by foot type for each load by combining all loading conditions in a single model and using a foot type by load condition interaction to estimate means and mean differences in outcomes by foot for each load separately. This model assumes a common variance in outcomes across load and foot. Thus, standard errors for the means will be similar across load and foot. Subject and foot type within subject were considered random effects. Several random effects structures were considered (from subject ID intercept only to a full subject by foot interaction model) with the best model chosen based on likelihood ratio tests. Hypothesis testing for the overall association between outcome and foot type were carried out using conditional F-tests for each load type. Additionally, the Benjamini Hochberg (BH) adjustment was applied to the F-tests for each model across the 5 load conditions to maintain a false discovery rate of 5 %. Hypothesis testing for pairwise differences across feet was carried out using Tukey's method to account for the inflation of the Type 1 error due to assessment of 10 pairwise differences for each load. Analyses were carried out using R 4.2.3 and packages tidyverse, lme4, emmeans and lmerTest.

3. Results

During the no load condition (shown in the left set of bars in each graph in Fig. 2, and supplementary material Table S1), the PF had a larger anterior GRF impulse compared to the SF by 0.044 Ns/kg (95 % confidence integral (CI) [0.003, 0.085], $p = 0.027$) and compared to the DK foot by 0.043 Ns/kg (CI [0.002, 0.084], $p = 0.036$) (Fig. 2B). The PF also had a smaller sagittal WBAM range compared to the CP foot by -0.0019 (CI [0.0008, 0.003], $p = 0.00007$), compared to the HW foot by -0.0012 (CI [0.0001, 0.0023], $p = 0.032$), and compared to the SF foot by -0.0012 (CI [0.0000, 0.0023], $p = 0.044$) (Fig. 2C). The CP foot had a larger anterior GRF impulse compared to the SF foot by 0.019 Ns/kg (CI [0.000, 0.038], $p = 0.048$) (Fig. 2B), as well as a larger sagittal WBAM range compared to the DK foot by 0.0012 (CI [0.0002, 0.0022], $p = 0.0075$) (Fig. 2C).

With the posterior load (second from left set of bars in each graph in Fig. 2, and supplementary material Table S2), the PF appears to have had a larger vertical GRF impulse compared to the CP foot by 0.26 Ns/kg (CI [0.02, 0.51], $p = 0.024$) and compared to the DK foot by 0.24 Ns/kg (CI [0.00, 0.49], $p = 0.046$) (Fig. 2A). The difference between the Benjamini Hochberg adjusted p value and the unadjusted value suggests a greater possibility of false discovery for this comparison (supplementary material Table S2). However, the differences in means are great enough to warrant consideration of at least a meaningful trend across these feet. The PF also had a larger frontal WBAM range compared to the HW foot by 0.0014 (CI [0.0001, 0.0027], $p = 0.028$), and the SF foot had a larger frontal WBAM range compared to the HW foot by 0.0013 (CI [0.0002, 0.0024], $p = 0.012$) (Fig. 2D). Although the F-test for an overall association between anterior GRF impulse and foot type was significant for the posterior load, there were no significant pairwise difference between any of the feet (supplementary material Table S2).

During the anterior load condition (third from left set of bars in each graph in Fig. 2, and supplementary material Table S3), the HW foot had a larger anterior GRF impulse compared to the SF foot by 0.023 Ns/kg (CI [0.004, 0.043], $p = 0.0085$) (Fig. 2B). The HW foot also had a smaller frontal WBAM range compared to the PF by -0.0017 (CI [-0.0029, -0.0004], $p = 0.0025$) (Fig. 2D). The CP foot had a smaller frontal WBAM range compared to the PF by -0.0015 (CI [-0.0028, -0.0003], $p = 0.0094$), and the DK foot also had a smaller frontal WBAM range compared to the PF by -0.0014 (CI [-0.0026, -0.0002], $p = 0.018$) (Fig. 2D).

For the prosthetic side load condition and the intact side load condition (right two sets of bars in each graph in Fig. 2, and supplementary material Tables S4 and S5 respectively), there were no significant pairwise differences between any of the feet. However, the F-test for an overall association between sagittal WBAM range and foot type was significant for the prosthetic side load, and the F-test for an overall association between anterior GRF impulse and foot type was significant for the intact side load.

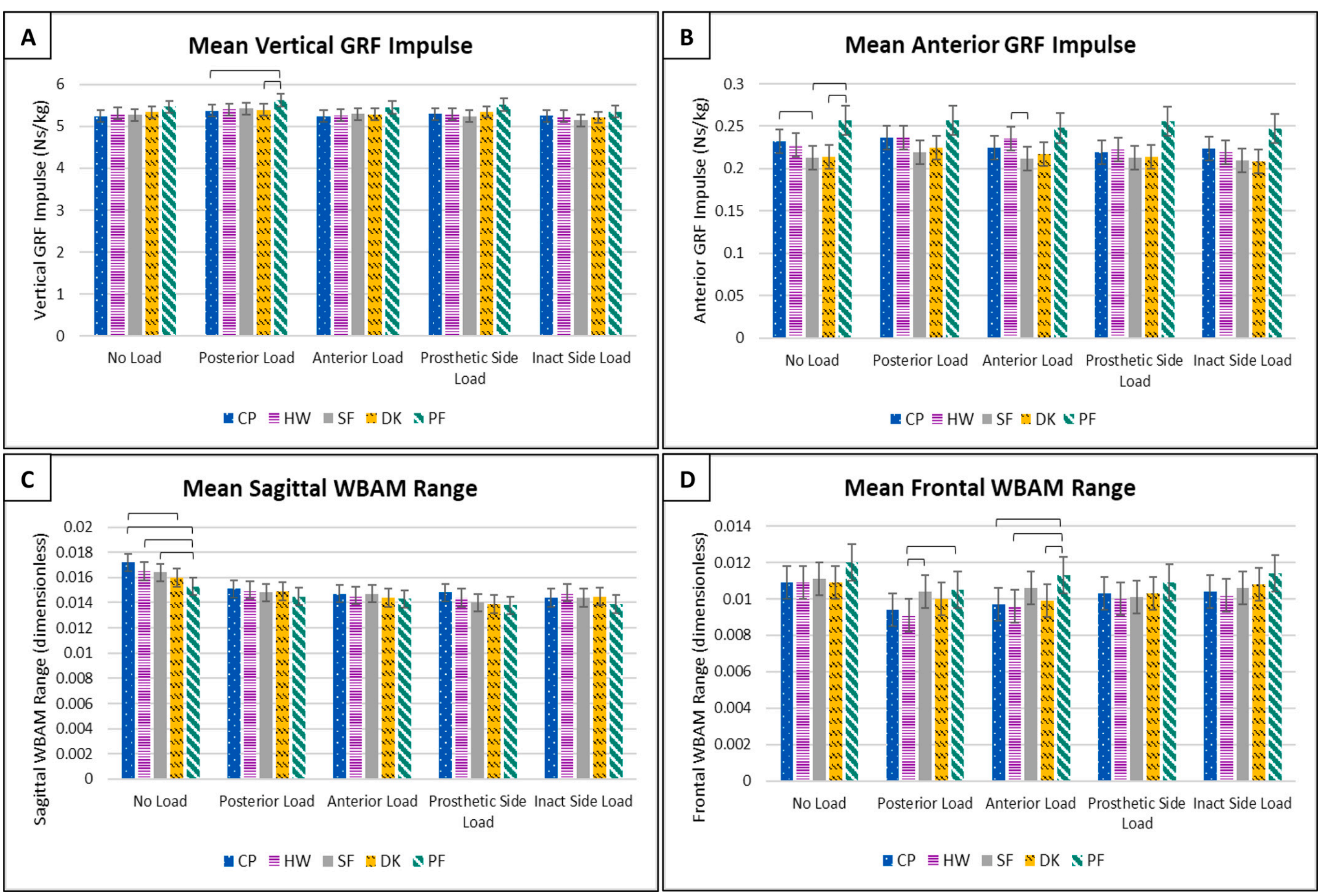

Fig. 2. Mean values, including standard error (SE) bars, for A) vertical GRF impulse (Ns/kg), B) anterior GRF impulse (Ns/kg), C) sagittal WBAM range, and D) frontal WBAM range. Horizontal brackets indicate pairwise significant differences. Abbreviations: clinically prescribed prosthetic foot (CP), clinically prescribed prosthetic foot with a heel-stiffening wedge (HW), clinically prescribed prosthetic foot but one category stiffer (SF), dual-keel prosthetic foot (DK), powered ankle-foot (PF).

4. Discussion

The purpose of this research was to examine the influence of five different prosthetic feet on walking performance during various load carriage conditions (no added load and with a load carried posteriorly, anteriorly, on the prosthetic side, and on the intact side). We hypothesized that a study prosthesis exists that: 1) maximizes vertical GRF impulse (a measure of body support), 2) maximizes anterior GRF impulse (a measure of body forward propulsion), 3) minimizes the peak-to-peak range of sagittal plane WBAM (a measure of anterior/posterior balance control), and 4) minimizes the peak-to-peak range of frontal plane WBAM (a measure of medio-lateral balance control) within each load carriage condition. This information may provide evidence to guide prosthesis prescription for individuals with a lower limb amputation who frequently carry infants, toddlers, backpacks, or other loads.

4.1. No load

With no load added, there were no differences in the vertical GRF impulse, suggesting the feet were equivalent in providing body support. However, there were differences in the anterior GRF impulse, namely, the PF provided significantly more forward propulsion (larger anterior GRF impulse) compared to the SF and DK feet. The PF also had a significantly smaller sagittal WBAM range compared to the CP, HW, and SF feet, which equates to the PF having improved sagittal balance control. Similarly, D'Andrea et al. found that transtibial amputees using a powered prosthesis were able to regulate their sagittal WBAM more effectively during level-ground walking compared to using a passive-elastic prosthesis (D'Andrea et al., 2014). However, their subjects with an amputation using the powered prosthesis were still not able to fully restore their WBAM ranges to those without amputation. While these are positive attributes of the PF for influencing walking performance, our study also found that the CP foot was also able to provide significantly more forward propulsion compared to the SF foot, and the DK foot was able to improve sagittal balance control compared to the CP foot. Depending on the needs of the amputee, their as prescribed prosthesis may provide adequate propulsion and support for walking without loads, and the more expensive PF may not be needed. However, if increased propulsion and sagittal balance control are of concern, the PF might be a good option to explore for level ground, unloaded walking.

For comparison, our CP foot, no load, anterior GRF impulses were somewhat higher than a previous study using a standard-of-care prosthetic foot (Silverman et al., 2008) as were our sagittal and frontal WBAM ranges (D'Andrea et al., 2014; Silverman and Neptune, 2011). There were some differences in amputation etiology between studies, but a post hoc analysis by Silverman et al. suggest etiology was responsible for only small differences and unlikely to be clinically meaningful (Silverman et al., 2008). There were also differences in the standard-of-care prosthetic feet worn between studies, as well as differences in the number of whole-body model segments, that may be responsible for the varied results. Our study used the same standard-of-care prosthetic foot for all subjects to reduce between subject variance.

4.2. Posterior load

When walking with an added posterior load, the PF appears to provide significantly more body support (larger vertical GRF impulse) compared to the CP and DK feet. There were no differences in anterior GRF impulse or sagittal plane WBAM range, suggesting the feet provided similar forward propulsion and sagittal balance control with a posterior load. The HW foot provided significantly improved frontal balance control (smaller frontal WBAM range) compared to PF and SF foot. For amputees who frequently carry posterior loads, adding a heel wedge to their existing prescribed prosthesis may be an easy, quick, and affordable option to improve frontal plane balance control without compromising body support. If body support is suffering significantly during posterior load carriage, a PF may be warranted, however frontal balance control may decrease as a result. A previous study has shown that the addition of a posterior load creates an increased need for stability enhancing strategies for transtibial amputees, such as decreased walking speed and step length, and increased double support time (Doyle et al., 2014). Therefore, the PF may not be the best option for posterior loads as it may only add to this instability.

4.3. Anterior load

During the anterior load condition, there were no differences in the vertical GRF impulse, suggesting the feet were equivalent in providing body support. The HW foot provided significantly greater forward propulsion (larger anterior GRF impulse) compared to the SF foot. There were no differences in sagittal plane WBAM, suggesting the feet provided similar sagittal balance control with an anterior load. The HW foot also provided significantly improved frontal balance control (smaller frontal WBAM range) compared to the PF. The CP and DK feet were also able to create significantly improved frontal balance control compared to the PF. The addition of a heel wedge stiffens the heel keel, preventing excessive heel keel deflection under added anterior load. This action may facilitate progression to mid-stance and add to forward propulsion.

4.4. Prosthetic side load and intact side load

Although there were no significant pairwise differences between any of the feet for the load on the prosthetic side or for the load on the intact side, similar trends were observed. The PF tended to have larger vertical and anterior GRF impulses compared to the other feet, suggesting a potential benefit for body support and forward propulsion during side load carriage conditions. The sagittal WBAM range was very similar across all feet. The HW tended to have a smaller frontal WBAM range compared to the other feet, suggesting improved frontal balance control while carrying side loads. A larger sample size is needed to confirm these trends.

4.5. Limitations

Limitations of this study should be considered when interpreting the results. The sample size was relatively small, mostly male, and mostly of traumatic etiology. Only seven out of the twelve subjects were able to complete the protocol with the PF due to build height and residual leg length restrictions. Having the same number of participants for all study prosthetic feet, and a larger sample size in general, might have confirmed some trends as statistically significant. There was no rationale for recruitment of participants by sex, but recruitment was conducted at a U.S. Department of Veterans Affairs medical facility where many Veteran patients are male. Females may be at greater risk for load carriage related injuries, suggesting further work examining sex/gender issues is warranted (Orr et al., 2020). While statistical significance was found for some outcomes, generalizing to a larger lower limb loss population should be done with caution as patients of diabetic/dysvascular etiology may have more difficulty with load carriage. The added load of 13.6 kg is another important limitation and was selected because many manufacturers use a 10 kg range to differentiate foot stiffness categories, so we sought to exceed that range. It was also selected to ensure all participants could complete all procedures. A heavier load would likely produce more significant differences in the study outcomes. Finally, this study did not measure cost-effectiveness or long-term outcomes. A future study with these metrics would provide additional evidence-based support for clinical practice.

5. Conclusions

This study provides insight into how different prosthetic feet influence walking performance during various load carriage methods. The PF showed some benefits during the no load (increased forward propulsion and sagittal balance control) and posterior load (increased body support) conditions. However, during the no load condition the CP foot was also able to provide increased forward propulsion and the DK foot was able to improve sagittal balance control. Depending on the biomechanical needs of the individual, any of these options could be viable. The HW foot was able to improve frontal balance control during both posterior load and anterior load conditions, as well as increased forward propulsion during the anterior load condition. The addition of a heel wedge to an amputee's existing prescribed prosthesis may be an easy, quick, and affordable option to improve frontal plane balance control and increase forward propulsion without compromising body support when adding a posterior or anterior load. These results may help guide clinical prosthesis prescription for individuals with a lower limb amputation who frequently carry infants, toddlers, or other loads. They may also encourage further development of active or semi-active prosthetic feet that can adapt to varying load conditions.

CRediT authorship contribution statement

Krista M. Cyr: Writing – review & editing, Writing – original draft, Methodology, Investigation, Formal analysis, Data curation, Conceptualization. **Richard R. Neptune:** Writing – review & editing, Visualization, Methodology, Investigation, Funding acquisition, Conceptualization. **Glenn K. Klute:** Writing – review & editing, Project administration, Methodology, Investigation, Funding acquisition, Conceptualization.

Declaration of competing interest

None.

Acknowledgements

Christina Carranza (certified prosthetist) fit and aligned the study prosthetic feet for all subjects, and Jane Shofer (biostatistician) performed all statistical analyses.

This work was supported in part by Merit Review Award Number I01 RX00318 and by Career Development/Capacity Building Award Number IK6 RX002974 from the United States Department of Veterans Affairs Rehabilitation Research and Development Service.

Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.clinbiomech.2025.106440.

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