The Influence of Load Carriage on Knee Joint Loading and Metabolic Cost on Walking with Lower-Limb Amputation: A Preliminary Modeling Study

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ABSTRACT

Introduction: For able-bodied individuals, the mechanical output from the ankle muscles is modulated to meet the altered demands of load carriage. However, for individuals with a lower-limb amputation, the stiffness properties of standard-of-care prosthetic feet do not change with varying load conditions. Thus, individuals with amputation often develop gait asymmetries during load carriage that increase their risk for developing overuse injuries such as in the intact knee and increase the metabolic cost of walking relative to able-bodied individuals. The purpose of this preliminary study was to assess the influence of load carriage technique on knee joint loading and metabolic cost during gait of an individual with a below-knee amputation using a forward dynamics simulation framework.

Methods: Simulations were generated to track the experimental walking data of individuals with amputation for 3 loading conditions (unloaded, front load, and back load).

Results: These simulations showed that individuals with amputation rely on their intact limb as a compensatory strategy to meet the increased demands of carrying a load. Carrying a back load was found to increase intact knee joint loading relative to carrying a front load but reduced metabolic cost.

Conclusion: The tradeoff between joint loading and metabolic cost should be considered when determining the appropriate load carriage technique. Future work should focus on improving prosthetic foot designs to help reduce joint loading asymmetry and elevated metabolic cost during different loading conditions for individuals with lower-limb amputation. (*J Prosthet Orthot*. 2021;33:118–124)

KEY INDEXING TERMS: Prosthetic feet, transtibial amputation, musculoskeletal model, forward dynamics simulations, Biomechanics

For individuals with a lower-limb amputation who are capable of locomotion, clinicians must choose from a wide range of available prosthetic feet when prescribing a prosthesis. Most major prosthetic foot manufacturers offer a range of stiffnesses that are delineated within up to nine stiffness categories. Typically, clinicians prescribe prosthetic foot stiffness based on the weight and activity level of the patient. A heavier and/or more active individual would be prescribed a stiffer foot. In general, each stiffness category is intended to be used by individuals within a 10-kg range (~7% of body weight). However, the load borne by a prosthesis can change suddenly during activities of daily living, such as when an individual carries a load in his or her arms or a backpack. If this load exceeds 10 kg, an immediate change to a prosthetic foot with increased stiffness would be recommended.

For able-bodied individuals, the mechanical output from the ankle muscles is seamlessly modulated to meet the altered demands of load carriage.[1] However, the properties of passive prosthetic feet, such as stiffness, are constant and are not modulated with varying load conditions. As a result, individuals with lower-limb amputations respond to the increased load with greater metabolic costs[2] and biomechanical asymmetries such as increased intact-limb power generation and absorption and increased prosthetic foot dorsiflexion during late stance.[3–5] These altered gait mechanics and asymmetries can lead to the early onset of joint disorders. In particular, individuals with below-knee amputation have an increased prevalence of osteoarthritis in their intact leg relative to their residual leg and able-bodied individuals.[6–8]

Previously, musculoskeletal modeling and simulation tools have been used to analyze individual muscle and prosthetic foot-ankle contributions to body support and forward propulsion during walking (e.g., Refs[9,10]). In addition, a series of experimental and modeling studies have shown how lower-limb muscles adapt to altered loading conditions.[1,11] Collectively, these studies have highlighted the critical role of the ankle plantarflexors in contributing to the vertical ground reaction force (GRF) impulse (body support), positive horizontal trunk work (forward propulsion), and modulation of mechanical output of the leg in response to increased need for body support and

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forward propulsion. Musculoskeletal simulations have also helped to quantify other biomechanical metrics that are difficult to analyze with experimental data alone such as knee joint loading[12,13] and metabolic cost.[14]

The purpose of this study was to gain insight into the relationships between load carriage technique (anterior versus posterior carriage) and joint loading and energy expenditure using a forward dynamics simulation framework. We expect that there is an optimal carriage technique that minimizes energy expenditure and joint loading for the different loading conditions. Understanding these relationships and their influence on walking performance of persons with amputation can help guide prosthetic foot prescription and improve the mobility of individuals with amputation.

METHODS

MUSCULOSKELETAL MODEL

A below-knee amputee musculoskeletal model was created by modifying the "gait2392" model in OpenSim,[15] which is an open-source software system for developing detailed musculoskeletal models and creating dynamic simulations of a wide variety of human movement tasks. Briefly, the segments distal to the right tibia were replaced with a transected tibia, pylon-socket, and ankle-foot prosthesis with inertial properties adapted from LaPrè et al.[16] The hip joint was modeled with three degrees of freedom (flexion/extension, internal/external rotation, and adduction/abduction), whereas the knee and intact ankle were modeled as one–degree of freedom pin joints. The six degrees of freedom between the transected tibia and pylon-socket segment were locked, and the prosthetic ankle motion was modeled as a one–degree of freedom pin joint.[9,17] All muscles crossing the ankle joint were removed and a coordinate actuator was added at the ankle joint to reproduce the prosthetic ankle torque. In the loading conditions, front and back loads were added to the torso segment of the model with inertial properties adapted from Dembia et al.[18] The body representing each load was fixed to the torso.

FORWARD DYNAMICS SIMULATIONS

Simulations were generated for three gait cycles in each loading and prosthetic foot condition for a total of nine simulations (three gait cycles $\times$ three loading conditions) using OpenSim 3.3.[15] To perform the simulations, the model was scaled based on experimentally measured marker data taken from a single individual with a below-knee amputation (see *Experimental Data* below). In the static pose, the maximum error between the virtual markers placed on the model and the experimental markers was 3.2 cm and the root-mean-square (RMS) marker error was 1.8 cm. Joint angles throughout each gait cycle were calculated using an inverse kinematics algorithm to minimize the difference between virtual and experimental markers.[15] Maximum marker error during the inverse kinematics trials was less than 3 cm and the RMS error was less than 1.5 cm. A residual reduction algorithm was used to alter the joint kinematics and model inertial properties to improve the dynamic consistency between experimentally measured kinematics and kinetics.[15] The suggested mass adjustments were applied to each segment and the position of the torso center of mass was modified to match the recommended position. After these adjustments were made, the residual reduction algorithm was performed an additional time. The weighting of the tracking parameters was fine-tuned until peak residual forces were less than 15 N, average residuals were less than 5 N, and RMS errors in coordinates were less than 2 degrees or 2 cm for rotational and translational coordinates, respectively. During the tuning process, optimal forces for residuals were kept low, whereas the weights of closely tracked coordinates were incrementally decreased. Computed muscle control was then used to solve for the muscle excitations that generated the individual's walking mechanics.[15] Computed muscle control solved the muscle redundancy problem at each joint by minimizing muscle activations squared while also accounting for muscle activation and deactivation dynamics.[19]

EXPERIMENTAL DATA

In this preliminary modeling study, the simulations tracked experimental data collected from one male individual with a below-knee amputation (age, 68 years; height, 176 cm; mass, 80 kg; amputated limb, right; postamputation, 1.5 years, modified patellar tendon-bearing socket with locking pin suspension) after providing informed written consent to an institutionally approved protocol. This subject was deemed representative because of his age,[20] height, and weight.[21] Sixty-two reflective markers were placed on the subject using a modified version of Vicon's Plug-in-Gait full-body model. The Plug-in-Gait model was modified by adding markers to the medial malleolus, medial elbow, and first and fifth metatarsal heads. Marker clusters were used to track the thigh and upper-arm segments instead of wands.[22] The shank segments were tracked with markers placed on the fibular head and tibial tuberosity. A 12-camera Vicon motion capture system (Vicon Motion Systems, Oxford, UK) collected kinematic data as the subject walked overground across five force plates (AMTI, Watertown, MA, USA) at his self-selected walking speed (SSWS). Ground reaction force and marker data were collected at 1200 and 120 Hz, respectively. Ground reaction force and marker data were filtered using a fourth-order low-pass Butterworth filter with cutoff frequencies of 20 and 6 Hz, respectively. Three successful overground walking trials were collected for each condition. A successful walking trial was defined as a trial in which foot strikes occurred on separate force plates. Before data collection, the SSWS was determined by asking the subject to walk down a 20-m hallway while wearing his current clinically prescribed prosthesis at his own pace. An average of three trials was used to define their SSWS.

STUDY PROTOCOL

Before data collection, the alignment and fit of the participant's clinically prescribed prosthesis were verified by a certified prosthetist using standard procedures. The subject then walked at his previously defined SSWS under three different loading conditions: unloaded (no load, NL) and with a pack carried posterior (back load, BL) and anterior (front load, FL) to his torso. Weights were placed inside the pack so that the total mass was 13.6 kg (30 lb). The pack was padded and included straps to secure the pack to the torso (Classic; Camelbak, Tetaluma, CA, USA). The subject was provided a minimum of 15 minutes to acclimate to each loading condition. Rest breaks were provided as needed at the subject's request.

JOINT LOADING AND METABOLIC COST

The axial tibiofemoral joint contact forces in both the intact and residual limbs were determined and expressed in the tibia reference frame and time integrated over the stance phase to provide the stance phase joint contact impulse for each leg. Instantaneous metabolic power for each muscle was determined based on the metabolic model by Umberger et al.[14,23] We then calculated average metabolic power by integrating the instantaneous metabolic power with respect to time and dividing by the gait cycle duration. To determine the metabolic cost, the total average metabolic power was determined by summing the contributions from the individual muscles. Joint contact impulses and metabolic cost were averaged over the three gait cycles for each load carriage condition.

RESULTS

JOINT LOADING

In response to the added load conditions (BL and FL), the intact knee joint contact impulses increased relative to NL (Figure 1A). The relative change in contact impulse between NL and BL (20%) was greater than the relative change between NL and FL (14%). The mean residual knee contact impulses were substantially lower than the mean intact impulses for all load carriage conditions. Similar to the intact limb, the residual knee contact impulses increased during BL and FL relative to NL (Figure 1B). However, the relative increases in residual knee contact impulse between NL and BL (13%) and NL and FL (13%) were less than the increases in the intact knee during BL and FL.

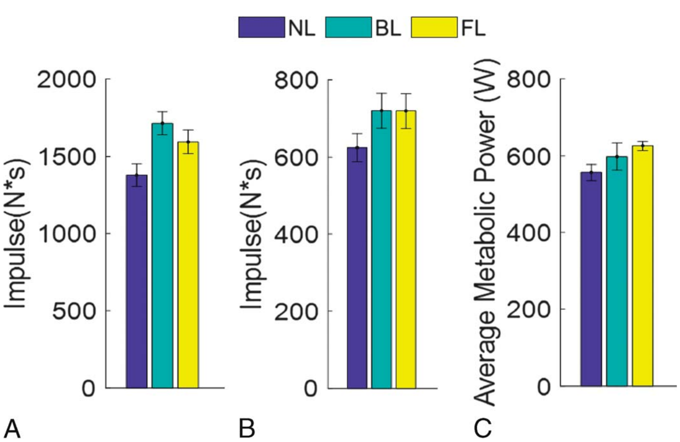

Figure 1. A, Mean stance phase intact knee contact impulses (N*s) ± 1 standard deviation for the three loading conditions. B, Mean stance phase residual knee contact impulses (N*s) ± 1 standard deviation for the three loading conditions. C, Mean ± 1 standard deviation of the total average metabolic cost for the three loading conditions. NL indicates no load; BL, back load; FL, front load.

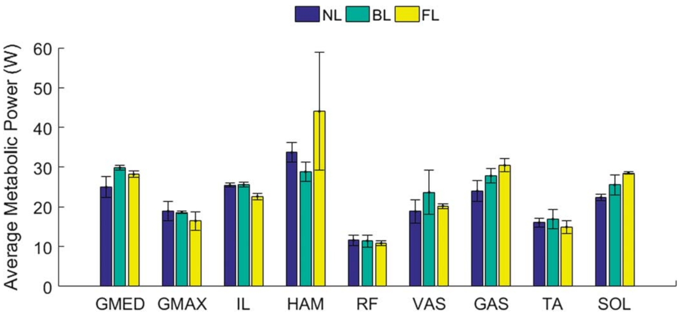

Figure 2. Mean ± 1 standard deviation of the total average metabolic cost of intact leg muscles throughout the gait cycle for the three loading conditions. NL indicates no load; BL, back load; FL, front load; GMED, gluteus medius; GMAX, gluteus maximus; IL, iliopsoas; HAM, biceps femoris long head, semimembranosus, semitendinosus; RF, rectus femoris; VAS, vastus medialis, vastus lateralis, and vastus intermedius; GAS, medial and lateral gastrocnemius; TA, tibialis anterior; SOL, soleus.

METABOLIC COST

Both BL and FL showed increased metabolic cost relative to NL (Figure 1C). The relative difference in metabolic cost between NL and BL (7%) was lower than the relative difference between NL and FL (11%). In addition, individual muscle contributions to metabolic cost were sensitive to the loading conditions. The intact leg GAS (medial and lateral gastrocnemius), SOL (soleus), VAS (vastus medialis, vastus lateralis, and vastus intermedius), and GMED (gluteus medius) showed increased cost in response to BL and FL (Figure 2). The average metabolic power consumed by SOL and GAS was greater during FL than BL. In contrast, GMED and VAS consumed more metabolic power during BL than FL. The largest change for any intact limb muscle in response to an added load was in HAM (biceps femoris long head, semimembranosus, semitendinosus) during FL. The GMAX (gluteus maximus), psoas, rectus femoris, and tibialis anterior showed minimal changes in response to the loads.

In general, the residual leg muscles consumed less average metabolic power and had a lower change in metabolic cost in response to BL and FL relative to the intact leg (Figure 3). Similar to the intact leg, the residual HAM showed the greatest increase in cost during FL. The GMED and GMAX showed increased cost relative to the intact limb and greater increases in response to BL and FL.

DISCUSSION

The purpose of this preliminary study was to analyze the influence of load carriage technique on knee joint loading and metabolic cost for an individual with a below-knee amputation using a forward dynamics simulation framework. In response to the added loads, we found increased joint loading asymmetry and increased intact limb muscle contributions to metabolic cost relative to the residual limb. We also found a lower average metabolic cost but higher knee joint loading during BL relative to FL.

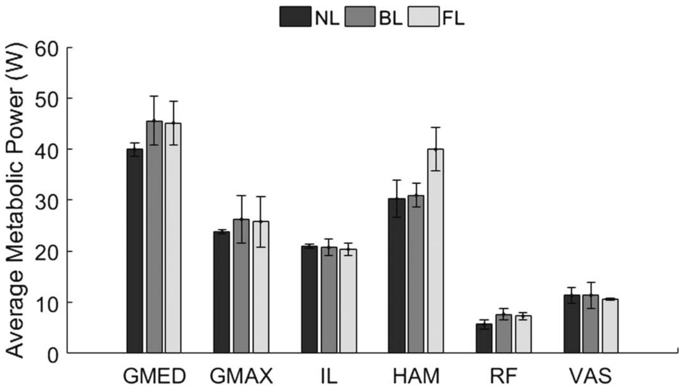

Figure 3. Mean ± 1 standard deviation of the total average metabolic cost of residual leg muscles throughout the gait cycle for the three loading conditions. NL indicates no load; BL, back load; FL, front load; GMED, gluteus medius; GMAX, gluteus maximus; IL, iliopsoas; HAM, biceps femoris long head, semimembranosus, semitendinosus; RF, rectus femoris; VAS, vastus medialis, vastus lateralis, and vastus intermedius.

JOINT LOADING

Previous studies have shown that individuals with amputation are at an increased risk of developing knee osteoarthritis in their intact limb.[6–8] Our results supported the expectation that the knee joint impulses experienced by the intact limb would be greater than those experienced by the residual limb in all loading conditions. In addition, we found that the imbalance between intact and residual knee loading was exacerbated in BL and FL relative to NL (Figure 1A, B). This increased asymmetric knee loading suggests that individuals with amputation rely on their intact limb as a compensatory strategy to meet the increased demands of carrying a load. Consequently, individuals with amputation who expect to carry various loads during activities of daily living may be at an increased risk of developing knee joint disorders and, ultimately, osteoarthritis.

We also found that joint loading during load carriage was dependent on the location of the load, as BL resulted in greater intact knee contact impulses compared with FL (Figure 1A, B). Previously, VAS has been shown to be a primary contributor to the compressive knee contact force during unloaded walking,[24] and VAS muscle activity and contributions to body support increase in response to added loads.[1] In addition, increased intact VAS muscle contributions to body support during the first half of stance have been associated with increased knee joint loading asymmetry in individuals with below-knee amputation.[17] Thus, we expected VAS to exhibit increased muscle activity and contributions to body support during FL than NL and an even greater increase during BL. In addition, during BL, the added posterior load acts to lean the body backward relative to its center of mass. Because VAS has been shown to contribute to forward angular momentum in early stance,[25] we expected that the BL condition would also require increased VAS output. To test these expectations, we performed additional analyses to determine the contribution of intact VAS to body support (i.e., vertical acceleration of the body center of mass) and sagittal plane angular momentum in all three loading conditions using previously described methods.[25,26] We identified muscle contributions to sagittal plane angular momentum by calculating muscle contributions to the time rate of change of angular momentum (i.e., external moment) as:

$$\dot{\overline{H}} = \overline{M} = \overline{r} \times \overline{F}_{\mathrm{GRF}}, \tag{1}$$

where $\overline{r}$ is the moment arm vector from the foot center of pressure on the foot to the body's center of mass and $\overline{F}_{\mathrm{GRF}}$ is the vector of the muscle's contribution to the GRF.

The results showed that during BL, the intact VAS muscle activity was increased and the contributions to body support and forward angular momentum were considerably higher relative to FL and NL (Figure 4). Thus, the reduced FL intact knee joint loading compared with BL was likely a result of a reduced demand placed on the intact leg VAS to provide body support and forward angular momentum.

METABOLIC COST

Our metabolic results are consistent with previous studies showing that metabolic cost increases during loaded walking relative to unloaded.[2,27] In particular, GAS and SOL showed markedly increased metabolic cost during BL and FL relative to NL (Figure 2). This is consistent with previous studies showing that GAS and SOL exhibit increased muscle activity during loaded relative to unloaded walking[11,28] and are the primary

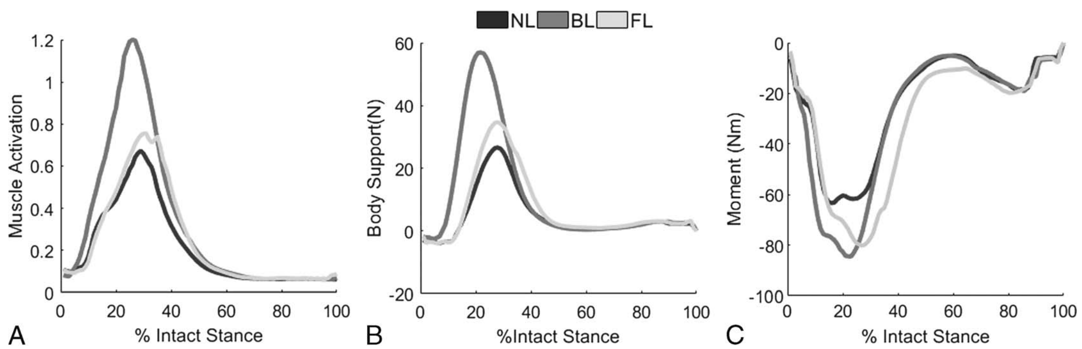

Figure 4. A, Estimated intact VAS muscle activity. B, Mean intact VAS contribution to body support over the stance phase for each of the three loading conditions. C, Mean intact leg VAS contributions to sagittal plane external moment about the center of mass of the body during the three loading conditions. Negative values indicate that the muscle is generating forward angular momentum. NL indicates no load; BL, back load; FL, front load; VAS, vastus medialis, vastus lateralis, and vastus intermedius.

muscles that respond to increased demand for body support and forward propulsion with increased loads.[1] Additional muscles that are responsible for body support, such as the intact and residual GMED,[1] also showed increased metabolic cost in both BL and FL (Figures 2 and 3).

Although FL resulted in lower knee joint loads relative to BL (Figure 1A, B), we found that BL produced lower metabolic cost relative to FL (Figure 1C). The primary contributor to the increased metabolic cost during FL was the intact HAM (Figure 2). During FL, the added anterior mass acts to lean the body forward relative to its center of mass. Because HAM has been shown to be a key contributor to generating backward angular momentum in early stance,[25] we expected that the FL condition would require increased output from HAM. An analysis of HAM contributions to sagittal plane angular momentum showed that increased HAM output was indeed required to control sagittal plane angular momentum during FL relative to both NL and BL (Figure 5), which ultimately contributed to the increased metabolic cost of FL.

Because both HAM and VAS contribute to providing body support, increased HAM output would reduce the need for VAS to contribute to body support.[1] Previous research has shown that HAM contributions to knee joint compressive forces are lower relative to VAS during walking.[12] Thus, the increased HAM output during FL may be an effective compensatory strategy to offload the intact knee joint loads.

LIMITATIONS AND FUTURE WORK

The primary limitation of this study was that the results may not be generalizable to all lower-limb prosthesis users as our preliminary study was based on analysis from one subject, and thus, we were not able to perform statistical analyses and could only observe trends. In addition, although this subject was deemed representative because of his age[20] and height and weight,[21] many other factors influence the gait mechanics of individuals with amputation. For example, prosthesis fit and the type of interface between the socket and residual limb (such as vacuum suspension and pin-locking) influence user comfort, and thus, a larger sample size would help validate the results of this study. We also only analyzed the influence of the subject's clinically prescribed passive elastic energy-storage-and-return (ESAR) foot on joint loading and metabolic cost. Studies have shown that joint loading and metabolic cost are influenced by changes in prosthetic foot stiffness and that there is an optimal stiffness that minimizes these metrics during unloaded walking.[17] Similarly, an alternative commercially available option to ESAR feet is powered ankle-foot prostheses. Some studies suggest that powered feet can provide substantial benefit to the population of individuals with amputation,[29–31] whereas other studies suggest mixed results.[32–35] Similarly, prosthetic feet with damping properties may also provide benefit to individuals with amputation.[36–38] Therefore, future work should investigate the ability of varying foot stiffness and these types of alternative devices to reduce joint loading and metabolic cost during various load carriage conditions. Future work should also be focused on not only anterior and posterior load carriage but also loads carried to the left or right of the body, which are common in activities of daily living.

An additional aspect of load carriage walking performance that should be studied is dynamic balance. With the functional loss of the plantarflexors, persons with amputation not only exhibit increased joint loading asymmetries and metabolic cost but also have a significantly altered angular momentum profiles relative to able-bodied individuals.[39] The present study showed that the different load conditions require altered control of whole-body angular momentum. Thus, future work should focus on how muscles work together in synergy to provide balance control in response to different loading conditions, which would provide evidence to assist clinicians in the prescription of prostheses for patients who are likely to carry various loads.

CONCLUSION

In summary, the increased asymmetric knee loading during load carriage observed in this study suggests that individuals with amputation rely on their intact limb as a compensatory strategy to meet the increased demands of carrying a front or back load. Carrying a front load reduced intact knee joint loading compared with carrying a back load through a reduced demand placed on the intact leg VAS to provide body support. However, the increased demand placed on the intact HAM during front load carriage increased the metabolic cost in comparison to back loads. This tradeoff between knee joint loading and

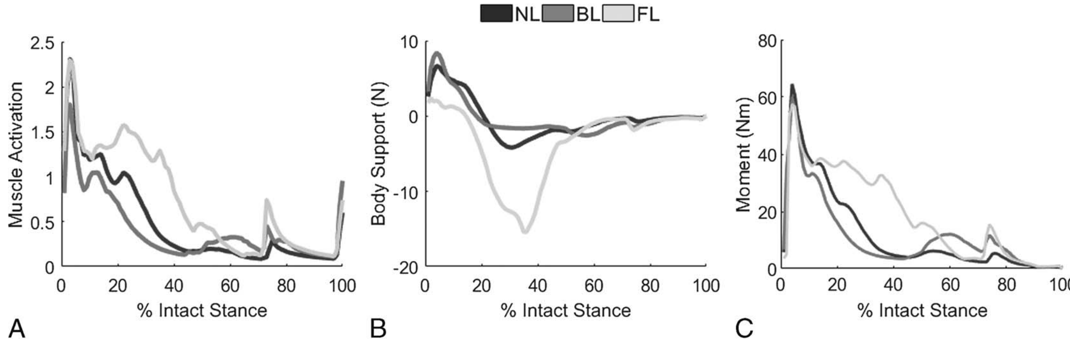

Figure 5. A, Estimated intact HAM muscle activity. B, Mean intact HAM contribution to body support over the stance phase for each of the three loading conditions. C, Mean intact leg HAM contributions to sagittal plane external moment about the center of mass of the body during the three loading conditions. Positive values indicate that the muscle is generating backward angular momentum. NL indicates no load; BL, back load; FL, front load; HAM, biceps femoris long head, semimembranosus, semitendinosus.

metabolic cost while carrying a back versus front load should be taken into consideration when determining load carriage technique. Future work should focus on improving prosthetic foot designs to help reduce joint loading asymmetry and elevated metabolic cost in response to different loading conditions for persons with lower-limb amputation.

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