Development of a Bimodal Ankle-Foot Prosthesis for Walking and Standing/Swaying

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The human ankle-foot system conforms to a circular effective rocker shape for walking, but to a much flatter effective shape for standing and swaying. Many persons with lower limb amputations have impaired balance and reduced balance confidence, and may benefit from prostheses designed to provide flatter effective rocker shapes during standing and swaying tasks. This paper describes the development and testing of an ankle-foot prosthesis prototype that provides distinctly different mechanical properties for walking and standing/swaying. The prototype developed was a single-axis prosthetic foot with a lockable ankle for added stability during standing and swaying. The bimodal ankle-foot prosthesis prototype was tested on pseudoprostheses (walking boots with prosthetic feet beneath) for walking and standing/swaying loads, and was compared to an Otto Bock single-axis prosthetic foot and to able-bodied data collected in a previous study. The height-normalized radius of the effective rocker shape for walking with the bimodal ankle-foot prototype was equal to that found earlier for able-bodied persons (0.17); the standing and swaying effective shape had a lower height-normalized radius (0.70) compared with that previously found for able-bodied persons (1.11). The bimodal ankle-foot prosthesis prototype had a similar radius as the Otto Bock single-axis prosthetic foot for the effective rocker shape for walking (0.17 for both), but had a much larger radius for standing and swaying (0.70 for bimodal, 0.34 for single-axis). The results suggest that the bimodal ankle-foot prosthesis prototype provides two distinct modes, including a biomimetic effective rocker shape for walking and an inherently stable base for standing and swaying. The radius of the prototype's effective rocker shape for standing/swaying suggests that it may provide inherent mechanical stability to a prosthesis user, since the radius is larger than the typical body center of mass's distance from the floor (between 50–60% of height). Future testing is warranted to determine if the bimodal ankle-foot prosthesis will increase balance and balance confidence in prosthesis users. [DOI: 10.1115/1.4024646]

Keywords: walking, prosthetic ankle, prosthetic foot, standing, posture

1 Introduction

Although the human ankle-foot system is complex, it has been shown that its functions for certain tasks are rather simple. For example, our group has shown that the human ankle-foot system conforms to a circular effective rocker shape (i.e., roll-over shape) during walking that remains consistent for different walking speeds [1], for added weight carried on the trunk [2], and for shoes of different heel heights and rocker profiles [3,4]. The effective rocker shape, or roll-over shape, is found by transforming the center of pressure of the ground reaction force into a shank-based coordinate system during the activity. The center of pressure provides an effective floor contact in the coordinate system of the shank, and the trajectory of the center of pressure suggests an effective rocker shape created by the ankle-foot system. The consistency of the roll-over shape for various conditions of walking has been used in the design of inexpensive ankle-foot systems that mimic able-bodied walking characteristics [5]. Measures derived from the roll-over shape have also been used as tools to evaluate commercially available prosthetic feet [6,7].

While the effective rocker shape of the ankle-foot system has been shown to be nearly circular and consistent for a wide variety of walking conditions, the effective rocker shape of the same system for standing and swaying has been shown to be much flatter (Fig. 1). Hansen and Wang [8] showed that the radius of the effective rocker shape is approximately 17% of body height for walking and about 110% of body height for standing and swaying. This finding suggests that the complex lower limb system conforms to a rather simple and stable effective surface for standing and swaying, i.e., a nearly flat base of support. Hansen and Wang [8] also developed a simple model to account for the different loading of prosthetic feet for walking (full body weight) and balanced standing/swaying (half body weight). Their model suggested that the stiffness needed in a prosthetic ankle to mimic standing/swaying function would need to be over three times that required to mimic walking function.

There are many activities of daily living that involve use of both hands during standing (e.g., washing the dishes or working at a standing workstation). During these tasks, the stability of persons who use lower limb prostheses depends upon their postural control and the mechanical characteristics of their prostheses. For many persons with lower limb amputations (e.g., older persons who have received their amputation as a result of diabetes or vascular disease), postural control can be impaired due to muscle weakness or other medical conditions, leading to reduced balance and balance confidence. Many of these people also have loss of sensation in their residual limbs and/or their remaining limbs, further reducing their ability to control their balance. Use of a

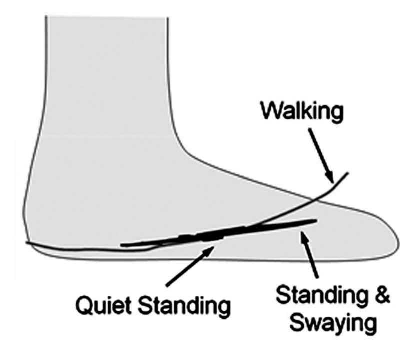

Fig. 1 Ankle-foot effective rocker shapes for walking (light gray), standing and swaying (dark gray), and quiet standing (black) (figure adapted from Ref. [8]). Effective rocker shapes of the ankle-foot system are found by transforming the center of pressure of the ground reaction force into a shank-based coordinate system.

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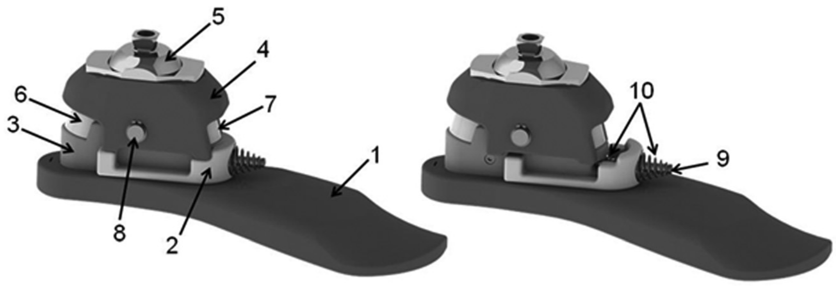


Fig. 2 CAD renderings of the final design in the locked standing/swaying mode (left) and in the unlocked walking mode (right). Major parts include the foot plate (1), slider (2), pillow block (3), ankle yoke (4), male pyramid (5), posterior bumper (6), anterior bumper (7), ankle shaft (8), small steel shaft (9), and slider springs (10). The slider is pushed and pulled by a small actuator located within the pillow block (see Fig. 3).

prosthesis with a flat effective rocker shape for standing could provide inherent stability to the body with less reliance on active control of the remaining joints. Miller et al. [9] showed that many lower limb prosthesis users have reduced balance confidence and also showed an association between balance confidence and social activities scores. Miller et al. also wrote that "Enhancement of balance confidence may improve mobility and social activity among the amputee population…" If prosthetic components can be developed that improve a person's balance and balance confidence for activities of daily living, they may participate in more social activities which may improve their quality of life, strength, and overall health.

There are many commercially available prosthetic ankle-foot systems. However, we are not aware of any commercially available systems that have distinct modes for walking and standing/swaying. One prototype foot incorporating a rigid rolling rocker for walking and a flat arched foot for standing was developed as part of a Ph.D. thesis [10]. However, subjects who used the prototype reported discomfort in early stance phase of walking, likely related to a lack of compliance in the heel [10]. The goal of this project was to develop and test a prosthetic ankle-foot system with separate modes for walking and standing/swaying and a simple method for switching between the two modes. Additionally, we sought to achieve the design goal while also incorporating heel compliance for shock absorption in the early stance phase of walking and flexibility in the forefoot for energy storage and return during late stance phase of walking. The rationale for the development of this system is for future testing with persons having lower limb amputations, particularly those with balance issues, to determine if balance and balance confidence can be improved.

2 Methods

2.1 Design Concept. As suggested earlier by Hansen and Wang [8], a simple design concept for creating both circular and flat effective rocker shapes is a single-axis ankle-foot prosthesis with a lockable ankle joint. In unlocked (walking) mode, the single-axis ankle-foot prosthesis should be designed to have the appropriate ankle stiffness to provide the biomimetic ankle-foot roll-over shape. In locked (standing/swaying) mode, the foot's keel should be sufficiently stiff, perhaps bending only to the biomimetic ankle-foot effective shape for standing/swaying (approximately 110% of body height).

2.2 Design Description. A locking single-axis ankle-foot prototype was designed to achieve a circular effective rocker shape for walking and a flat effective shape for standing. The overall design is similar to current single axis ankle-foot prostheses—i.e., having a rotating joint that interacts with rubber bumpers. However, the design has a novel sliding mechanism that mechanically blocks ankle movement for standing tasks.

A computer-aided design (CAD) rendering of the final prototype is shown in Fig. 2. The mechanical parts of the ankle-foot system include a foot plate, slider, pillow block, ankle yoke, male pyramid adapter, posterior bumper, anterior bumper, and ankle shaft. The pillow block is bolted to the foot plate and interacts with the ankle yoke via the ankle shaft. The posterior and anterior bumpers are positioned on angled ledges of the pillow block and resist plantarflexion and dorsiflexion movements, respectively. The slider rests on the foot plate and is pushed and pulled by an actuator (Firgelli PQ12 linear actuator, 20 mm, 12 V) connected to a small steel shaft (Figs. 3 and 4). The small steel shaft is connected to the slider using two springs. The actuator is strong enough to overpower the springs and go to its limits of movement if necessary. If the ankle is flexed to a non-neutral position that does not allow the slider to slide under the ankle yoke when trying to go from unlocked to locked mode, the spring can compress and then move the slider under the yoke after the user shifts their weight off of the ankle. When going from locked to unlocked mode, the user may have load on the ankle causing friction between the ankle yoke and the slider. In that case, the actuator can still extend to its limit and when the user removes load from the ankle, the spring will push the slider out from under the ankle yoke unlocking the joint.

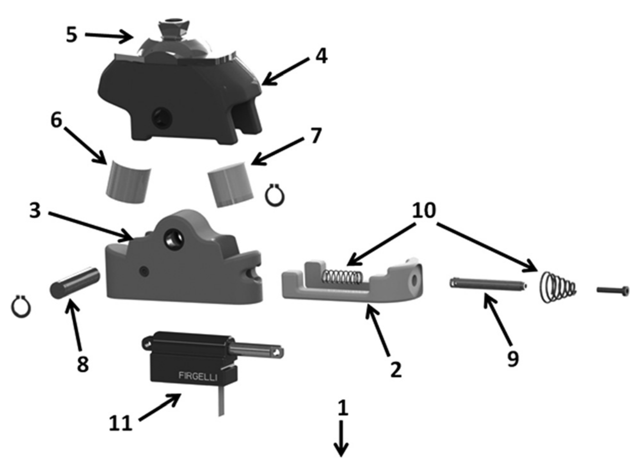


Fig. 3 Exploded view of the working parts of the bimodal ankle-foot system, labeled as per Fig. 2, with the linear actuator (11) shown. The ribbon cable from the actuator is cut short in the drawing. In the prototype, the ribbon cable folded into a track along the bottom of the foot plate (1–not shown) and came out a small slot in the posterior section of the foot plate. The slot is visible in the CAD renderings in Fig. 2.

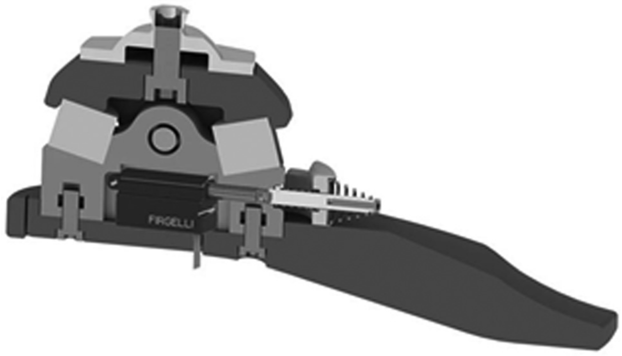


Fig. 4 Section view of the bimodal ankle-foot system showing the fully-assembled position of the parts, including the linear actuator which is located within a hollow space in the pillow block

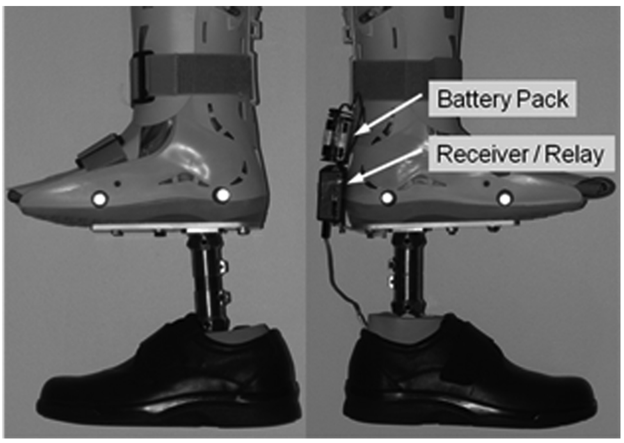
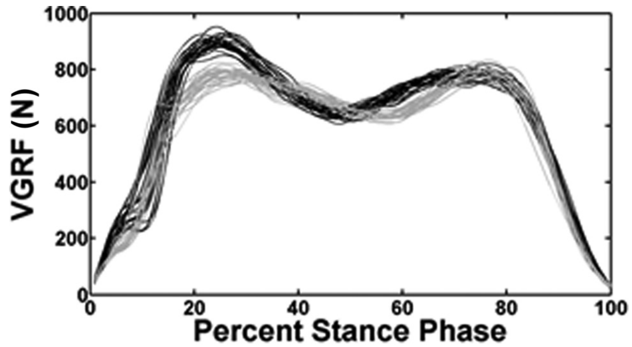


Fig. 5 Photographs of the pseudoprostheses connected to the Otto Bock single-axis prosthetic foot (left) and the bimodal ankle-foot prosthesis prototype (right). The wireless receiver, relay, and battery pack were taped to the pseudoprosthesis for testing of the prototype.

The ribbon cable of the actuator was passed through a channel and through a posterior slot in the foot plate. This cable was connected to a relay (OMI-SS-212D), which was controlled by a wireless receiver (Firgelli Automations 2-4 Channel Remote Control System) and powered by a battery pack (8-1.5 V-AA batteries in series) (Fig. 5). The wireless receiver received commands to switch states using a wireless key fob (not shown in Fig. 5). Upon receiving the command, the wireless receiver switched states on the relay, reversing the polarity of voltage applied to the actuator causing it to move the slider to lock or unlock the ankle joint.

The foot plate, slider, pillow block and ankle yoke were fabricated out of ULTEM® using a fused deposition layering rapid prototyping machine (Stratasys FDM 400). The bumpers were made from Shore 60 A polyurethane cylinders with 22 mm diameter. The bumpers were cut to a length that required a small amount of precompression. The ankle was assembled in a vise that provided the force to precompress the bumpers to align the ankle shaft with the ankle yoke and pillow block. The top of the ankle yoke was designed to interface with a standard endoskeletal male pyramid adaptor (Hosmer Dorrance SACH foot adaptor).

2.3 Experimental Protocol. The bimodal ankle-foot prosthesis prototype was tested under pseudoprostheses [11]. Pseudoprostheses are Aircast® walking boots that have been modified to allow prosthetic feet to be attached beneath (see Fig. 5). The bimodal ankle-foot prosthesis prototype was tested within a



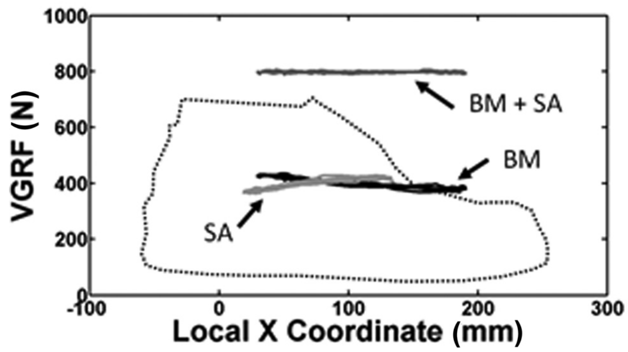


Fig. 6 Vertical ground reaction forces (VGRF) on bimodal ankle-foot prosthesis prototype (BM–black) and single-axis foot (SA–gray) for walking (top) and standing/swaying (bottom). The dark gray line in the standing/swaying plot (bottom) is the sum of the forces on both feet during standing/swaying.

commercially available cosmetic foot shell (College Park Industries, Fraser, Michigan), which was placed inside a shoe (Ambulator BV3000MM11). Two reflective markers were placed along the lateral edge of the pseudoprostheses (see Fig. 5) and a third marker was placed on the medial side of the pseudoprostheses approximately in the plane of the pseudoprosthesis attachment surface. A person (mass = 70 kg) walked at a speed of 1.0 ms/s with the pseudoprostheses on a split-belt instrumented treadmill (Bertec, Columbus, Ohio) with the ankle in unlocked mode and performed standing and fore-aft swaying movements with the ankle in locked mode, similar to the movements of able-bodied persons in [8]. The bimodal ankle-foot prosthesis prototype was tested under the right side pseudoprosthesis and an Otto Bock Single-Axis prosthetic foot was tested under the left side pseudoprosthesis for comparison. Reflective markers were tracked during the walking and standing/swaying trials using an 8-camera motion analysis system (Qualisys, Gothenburg, Sweden). The use of pseudoprostheses for testing new prosthesis prototypes by our research staff was reviewed by the Institutional Review Board at the Minneapolis VA Health Care System and was declared to be exempt from human subjects protection oversight.

2.4 Data Analysis. The center of pressure of the ground reaction force, measured by the instrumented treadmill, was transformed into a local coordinate system using the three markers on the pseudoprostheses. The three markers defined a plane approximately parallel to the attachment surface of the pseudoprosthesis. The two lateral markers defined the X direction of the coordinate system (pointing anteriorly) and the vector normal to the plane created by the three markers and pointing proximally defined the Z direction. The center of pressure trajectory in a shank-based coordinate system describes the effective rocker shape of the ankle-foot system during the activity (i.e., walking or fore-aft swaying [8]). The local coordinate systems on the pseudoprostheses were intended to represent shank-based reference frames for

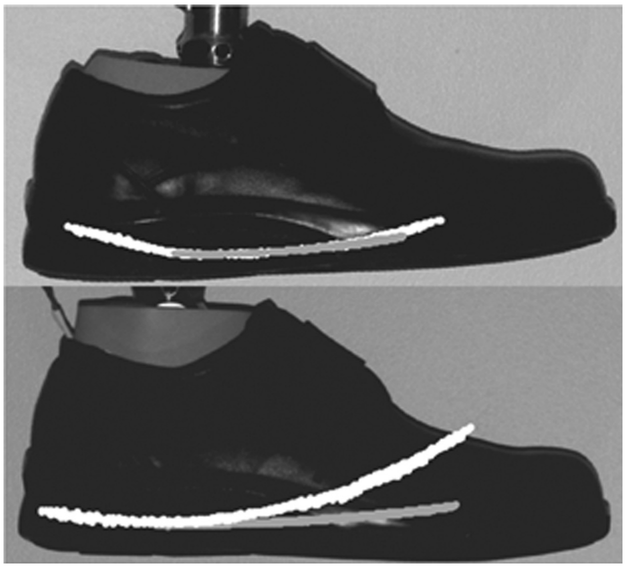


Fig. 7 Effective shapes of the single-axis prosthetic foot (top) and bimodal ankle-foot prosthesis prototype (bottom) for walking (white) and standing/swaying (gray). The bimodal ankle-foot prosthesis prototype appears to have a more distinct difference in effective shapes compared with the single-axis prosthetic foot, consistent with the distinct differences found earlier in able-bodied persons (Fig. 1). (Note that the top image has been flipped horizontally to facilitate comparison with able-bodied data (Fig. 1) and the bimodal ankle-foot prosthesis prototype.)

prostheses that would incorporate the tested components. The effective rocker shapes were overlaid on a photograph of the foot for qualitative evaluation, using the lateral markers in the photograph to approximate the scaling and positioning. The effective rocker shapes were also modeled as the lower arc of a circle, using the technique described in Ref. [1], for more quantitative analysis and for comparison with the radii of effective ankle-foot rocker shapes previously measured on able-bodied persons [8]. The best-fit radii were normalized by the average height of a person that would have a 27 cm foot length, using an anthropomorphic scaling table [12]. Specifically, we assumed persons that would use a 27 cm foot would have an average height of 177.6 cm (foot length = 0.152 × height).

The vertical loads on the prosthetic feet were examined for walking and standing/swaying experiments. For walking, the vertical ground reaction forces were plotted as a function of the stance phase cycle. For standing/swaying, the vertical ground reaction forces under each prosthetic foot were plotted as a function of the X direction of the local coordinate system, with lower X values representing heel loading and higher X values representing forefoot loading.

3 Results

The vertical ground reaction forces for walking showed a pattern consistent with that found in normal walking (Fig. 6(*a*)). For standing/swaying, the loading was nearly balanced between the feet (Fig. 6(*b*)). The effective shapes (walking and standing/swaying) for the bimodal ankle-foot prosthesis prototype and the single-axis prosthetic foot are shown in Fig. 7. The height-normalized radii of the best-fit circular rockers for the bimodal ankle-foot prosthesis prototype were 0.70 for standing/swaying and 0.17 for walking (Fig. 8). For the single-axis prosthetic foot, the height-normalized radii of the best-fit circular arcs were 0.34 for standing/swaying and 0.17 for walking. The standing/swaying radii for both prostheses tested were smaller than that measured in able-bodied persons previously (1.11), while the radii for their walking effective shapes matched that previously measured for able-bodied persons (0.17).

4 Discussion

The results of the experimental testing support the existence of two distinct functional modes for the bimodal ankle-foot prosthesis prototype. In particular, the bimodal ankle-foot prosthesis prototype conformed to an effective rocker shape that was much flatter during standing and swaying compared with that for walking. The radius of the effective rocker shape for walking with the bimodal ankle-foot prosthesis prototype was equal to the median radius used by able-bodied persons. The radius of the standing and swaying effective shape did not fall into the interquartile range of the corresponding measurement for able-bodied persons. However, the radius of the bimodal ankle-foot prosthesis prototype during standing and swaying was large enough to provide a mechanically stable system. The walking human has been

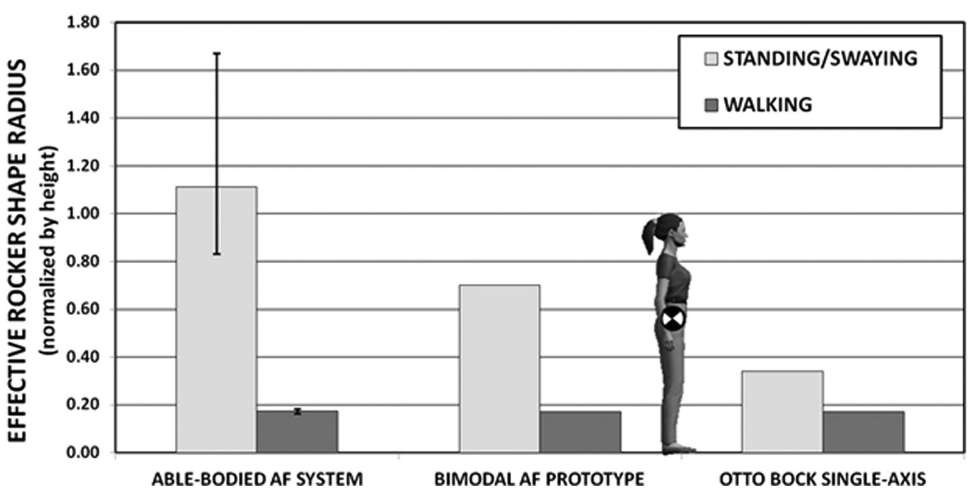


Fig. 8 Best fit radii to the effective rocker shapes of able-bodied ankle-foot systems compared with effective shapes measured for the bimodal ankle-foot (AF) prosthesis prototype and the Otto Bock single-axis prosthetic foot. Data for the able-bodied ankle-foot system are medians with error bars drawn between the first and third quartiles [8]. A picture of a human is drawn to indicate scaling to height as well as an indicator of the body center of mass, which is typically between 0.5 and 0.6 times body height. A rocker radius that is greater than the height of the body center of mass is mechanically stable.

modeled by some investigators as an inverted pendulum with a rocker shape for a foot [13–15]. Using this model, the body is inherently stable if the radius is larger than the height of the body center of mass from the floor. When the radius is less than the height of the body center of mass from the floor, the person is inherently unstable and will tend to fall to one direction or the other without application of external forces. The Otto Bock single-axis prosthetic foot also provided a higher radius for standing and swaying compared with that for walking, likely due to the reduced loading on the system during standing/swaying compared with walking. However, the difference was not enough to create an inherently stable system for standing because the radius for standing and swaying is less than the distance from the floor to the body center of mass.

The radius of the effective rocker shape of the bimodal ankle-foot prosthesis for standing and swaying could be increased by utilizing a stiffer foot plate and/or by designing a tighter lock. The locking mechanism of the prototype described in this paper consists of mechanical blockage of rotation of the ankle yoke with respect to the foot plate using a sliding mechanism. In order to allow the sliding mechanism to engage and disengage the lock, some clearance is required. This clearance inherently leads to some ankle motion during the standing and swaying activity and likely a smaller radius of the effective rocker shape. In future work, different locking mechanisms could be incorporated that provide tighter locking of the ankle joint. The current locking mechanism locks about only one ankle angle. Future locking mechanisms that could lock over a range of angles may be useful to lower limb prosthesis users for standing on uneven surfaces (e.g., uphill slopes).

A limitation of the study is that walking and standing/swaying loads were applied to the prosthetic ankle-foot systems using pseudoprostheses instead of within real lower limb prostheses. However, the loading shown in Fig. 5 suggests that the loads applied to the feet were similar to what would be expected for walking and for balanced standing/swaying with a prosthesis. Testing with pseudoprostheses suggests that the bimodal ankle-foot prosthesis works as designed, creating distinctly different modes for walking and standing/swaying. Testing in persons with lower limb amputation is the next step to see if the results are similar to those found in this study and also to see if the added locking feature of the bimodal ankle-foot prosthesis improves balance and balance confidence of lower limb prosthesis users over commercially available single-axis ankle-foot prostheses. Testing with prosthesis users in the future should also help to determine best approaches for switching between modes. More advanced electronics could be incorporated into the bimodal ankle-foot prosthesis to automatically switch between walking and standing/swaying modes, while still allowing the user to override the automatic control with a key fob, for example. Human testing is also needed to determine if any added benefit of the standing/swaying mode in terms of stability will overcome the burden of charging batteries to control the system and any additional weight of the bimodal ankle-foot prosthesis compared with other prosthetic feet for prosthesis users in the lowest functional levels.

In conclusion, a bimodal ankle-foot prosthesis prototype was developed and tested in this study. The testing was designed to mimic usage of the prototype for walking and standing/swaying activities. The results of the study suggest that the prototype worked as intended, providing a biomimetic effective rocker shape for walking and a stable base for standing/swaying. The next steps in the project include testing of the device by lower limb prosthesis users and commercialization of the technology.

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References

[1] Hansen, A., Childress, D., and Knox, E., 2004, "Roll-Over Shapes of Human Locomotor Systems: Effects of Walking Speed," Clin. Biomech. (Bristol, Avon), **19**(4), pp. 407–414.
[2] Hansen, A., and Childress, D., 2005, "Effects of Adding Weight to the Torso on Roll-Over Characteristics of Walking," J. Rehabil. Res. Dev., **42**(3), pp. 381–390.
[3] Hansen, A., and Childress, D., 2004, "Effects of Shoe Heel Height on Biologic Roll-Over Characteristics During Walking," J. Rehabil. Res. Dev., **41**(4), pp. 547–554.
[4] Wang, C. C., and Hansen, A. H., 2010, "Response of Able-Bodied Persons to Changes in Shoe Rocker Radius During Walking: Changes in Ankle Kinematics to Maintain a Consistent Roll-Over Shape," J. Biomech., **43**, pp. 2288–2293.
[5] Sam, M., Childress, D., Hansen, A., Meier, M., Lambla, S., Grahn, E., and Rolock, J., 2004, "The Shape & Roll Prosthetic Foot (Part I): Design and Development of Appropriate Technology for Low-Income Countries," Med. Confl. Surviv., **20**(4), pp. 294–306.
[6] Hansen, A., Sam, M., and Childress, D., 2004, "The Effective Foot Length Ratio (EFLR): A Potential Tool for Characterization and Evaluation of Prosthetic Feet," J. Prosthetics Orthotics, **16**(2), pp. 41–45.
[7] Curtze, C., Hof, A. L., van Keeken, H. G., Halbertsma, J. P., Postema, K., and Otten, B., 2009, "Comparative Roll-Over Analysis of Prosthetic Feet," J. Biomech., **42**(11), pp. 1746–1753.
[8] Hansen, A. H., and Wang, C. C., 2010, "Effective Rocker Shapes Used by Able-Bodied Persons for Walking and Fore-Aft Swaying: Implications for Design of Ankle-Foot Prostheses," Gait and Posture, **32**, pp. 181–184.
[9] Miller, W. C., Deathe, A. B., Speechley, M., and Koval, J., 2001, "The Influence of Falling, Fear of Falling, and Balance Confidence on Prosthetic Mobility and Social Activity Among Individuals With a Lower Extremity Amputation," Arch. Phys. Med. Rehabil., **82**, pp. 1238–1244.
[10] Adamczyk, P. G., 2008, "The Influence of Center of Mass Velocity Redirection on Mechanical and Metabolic Performance During Walking," Ph.D. thesis, University of Michigan, Ann Arbor, MI.
[11] Hansen, A. H., Childress, D. S., and Knox, E. H., 2000, "Prosthetic Foot Roll-Over Shapes With Implications for Alignment of Trans-Tibial Prostheses," Prosthetics Ortho. Int., **24**(3), pp. 205–215.
[12] Winter, D., 1990, *Biomechanics and Motor Control of Human Movement*, Wiley, New York.
[13] Gard, S., and Childress, D., 2001, "What Determines the Vertical Displacement of the Body During Normal Walking?" J. Prosthetics Ortho. **13**, pp. 64–67.
[14] McGeer, T., 1990, "Passive Dynamic Walking," Int. J. Robot. Res., **9**, pp. 62–82.
[15] Morawski, J., and Wojcieszak, I., 1978, "Miniwalker—A Resonant Model of Human Locomotion," Proceedings of the 6th Int. Congress of Biomechanics, Vol. 2A, E. Asmussen, and E., K. Jorgensen, eds., University Park Press, Baltimore, MD, pp. 445–451.