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Ability of a Robotic Ankle Prosthesis to Augment Effective Foot-Ankle Stiffness relative to Standalone Prosthetic Feet

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***Abstract*—Well-prescribed prosthetic feet are critical to restore gait for persons with limb loss. Criteria for prescription can be difficult to define due to differences in individuals and few measurements defining the mechanical properties of prostheses. The use of a robotic ankle in series with a prosthesis can provide biological levels of mechanical power during gait. In addition, active ankle prostheses can adapt to different use cases. To assess how this paradigm influences user assistance, we quantified the effective stiffness of standalone feet of varying clinical stiffness categories in comparison to a robotic ankle in series with a fixed category level prosthetic foot. We hypothesized that control of a powered ankle across its range of stiffness and damping parameters can expand the effective stiffness range offered by commercially available passive feet, and better explain the effective stiffness rendered during loading. Benchtop compression loading was completed on energy storage and return feet of manufacturer-defined stiffness category levels (4-9), as well as an integrated prosthetic foot (category 9) and robotic ankle system. Force-displacement data were used to characterize stiffness in toe- and heel-only loading, at low (~0-50% body weight) and high (~50-100%) end levels. Control of the ankle captured well most of the profiles of standalone feet, as well as responses outside of these behaviors at low stiffness. Generally, there were stronger linear relationships between effective stiffness and category level of standalone feet (r=~0.9), and less so between the stiffness gain of the robotic ankle and effective stiffness (r=~0.8). The *exception* was for high-end toe-only loading of the standalone and robotic conditions (r=0.76 and 0.92, respectively).**

I. Introduction

Clinical intervention with prosthetic feet requires understanding the widely-varying needs of individuals with lower-limb loss. Clinicians often consider factors such as how an individual's age, weight, amputation level and mobility level should affect the prescription of their prosthetic components. Prosthetic feet are a significant consideration when prescribing components of lower-limb prostheses due to their multitude of designs and effects on biomechanics during ambulation. Individuals with lower-limb loss experience several biomechanical differences regarding ambulation as compared to non-amputees, which include lower self-selected walking speed [1], increased intact limb loading [2, 3], increased gait asymmetry [4], and increased metabolic cost [1, 4, 5]. Prescription of a suitable prosthetic foot is vital to mitigating theses negative effects so that mobility can be restored, as mobility has been shown to be strongly related to quality of life for these individuals [6].

In the United States, the classification for functional levels for individuals with amputation ranges from K level 0 (no ability or potential to ambulate) to K level 4 (active individual or athlete). This functional level often determines the appropriate foot-ankle system to be prescribed due to the justification required by third-party payers. Lower K level individuals would typically be prescribed a less flexible foot to encourage balance and stability, while higher K level individuals would be prescribed a more flexible foot, e.g. an energy storage and return (ESAR) or often noted as a dynamic elastic response foot, to allow for more advanced ambulation behaviors. ESAR feet store and release mechanical elastic energy during stance in an attempt to restore the lost function of the ankle and surrounding musculotendon and orthopedic structures. Furthermore, there are clinical stiffness category levels of ESAR feet that descriptively characterize the perceived stiffness of the foot. The stiffness of prosthetic feet has been shown to impact the amount of body support and propulsion provided by the prosthesis, and consequently the activity of remaining muscles and effort required during ambulation [7-10]. Thus, it is important to not only prescribe the appropriate foot-ankle system, but also to prescribe the appropriate foot stiffness.

Prosthetists utilize manufacturer recommendation tables that consider a patient's body weight and activity level along with empirical knowledge to inform prescription. However, consistent quantitative measures of the mechanical properties of prosthetic feet across manufacturers and foot types are not generally made, which can complicate this prescription process. A few previous studies have reported inconsistent incremental changes in measured effective stiffness across stiffness categories of prosthetic feet [11, 12]. This makes prescription of prosthetic feet that the prosthetist does not have prior experience with more challenging and likely results in prescription of a relatively small portion of the commercially-available options. Finally, some individuals may also benefit from a stiffness outside of the commercially-available options for optimal biomechanical outcomes or one that is adaptive to their needs and use cases.

While well-prescribed ESAR feet have been shown to improve some of the previously-mentioned biomechanical outcomes as compared to less flexible options [13, 14], they are unable to completely restore the lost function of ankle muscles that are largely responsible for mechanical energetics

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of walking. They also have fixed stiffness values that cannot be changed throughout the stance phase or across different forms of gait or ambulation. To further encourage improving the biomechanical assistance provided to these individuals and offer adaptation of stiffness to different terrains and tasks, a multitude of semi-active and active ankles have been introduced, either commercially or within the research space [15-18]. One of these designs is the Open-Source Leg (OSL) which is a powered (i.e., robotic) knee-ankle system that was developed for open-source adoption to encourage advancement and reproducibility of control systems [19]. In this study, we use the ankle portion of the OSL in series with an ESAR foot to examine effective foot-ankle stiffnesses when these components are connected in series. Prosthetic ankles and knees that contain motorized actuators were first introduced around 2008-2010 [20, 21]. It has been shown in some previous studies that powered prostheses can restore metabolic energy [16] to normal levels, offload the contralateral limb [16] and help return to normal some time-varying measures of dynamic balance [22]. In addition, these devices are often highlighted for their ability to provide modulated or adaptive biomechanical assistance across variable walking terrains such as ramps and stairs [21, 23, 24].

In order to assess how prescribing commercially-available prosthetic feet or using a robotic ankle prosthesis in series with a prosthetic foot may influence the biomechanical assistance provided to users, the goal of this study was to quantify the effective stiffness of standalone commercially-available feet of varying clinical stiffness category levels in comparison to a robotic ankle in series with a prosthetic foot of a fixed category level. We hypothesized that control of the powered ankle across its range of stiffness and damping parameters can expand the effective stiffness range offered by commercially available passive feet, and better explain the effective stiffness that is rendered during loading.

II. Methods

A. Prosthetic Components and Testing Conditions

Ossur Vari-Flex low profile (LP) prosthetic feet were selected for this study. The Vari-Flex LP is a commonly prescribed ESAR carbon fiber foot with a reduced build height for individuals with long residual limbs or in this case that allows for attachment to a robotic ankle (Fig.1). Feet in stiffness category levels 4, 6, 7, and 9 were selected for testing based on availability and to represent a relatively standard clinical range of stiffnesses for feet of similar sizes. All feet tested ranged between sizes 26 and 27 cm. The powered ankle is part of the Open-Source Leg v1.0 originally developed at the University of Michigan. The Open-Source Leg consists of knee and ankle joints that operate independently which allows for individual joint testing. The Open-Source Ankle (OSA) specifically consists of two belt stages and a four-bar linkage which has a 30˚ range of motion. The ankle contains a linkage rocker that couples the transmission to the attached prosthetic foot. The angle of the linkage rocker corresponds to the ankle joint angle [19]. The OSA robotic ankle prosthesis was connected to the category 9 Vari-Flex LP foot, which was the stiffest of the standalone feet, and programmed across its recommended range of linear rotational stiffness and damping gains with a commanded equilibrium position of 0° resulting in torques (τ, mN*m) defined by (1); where k is the OSA gain, b is the damping, θ is the ankle angle, θ_{eq} is the equilibrium angle and $\dot{\theta}$ is the motor velocity [25].

$$\tau = 0.146(7.812 \times 10^{-4}k\left(\theta - \theta_{eq}\right) + 2.844 \times 10^{-4}b\dot{\theta} \quad (1)$$

Six different gains were commanded ranging from 300-10000. The specific gains are listed in Table 1. Damping (B) was paired with each gain to result in zero overshoot, as previously performed [25]. The highest and lowest ankle controller gains were chosen to represent the values that either saturated the ankle motor current for high-end stiffness conditions at max deflection or produced comparisons across conditions within the range of motion of the ankle joint (i.e., avoiding interaction with hard stops at the end range of the actuator and thus harm to the device), respectively.

B. Mechanical Characterization

Loading was completed on a benchtop uniaxial materials testing system (Instron Series 3345, Norwood, MA, USA). The feet and foot-ankle system were connected by their pyramid adaptors to the force transducer of the materials testing system with a custom machined adaptor. The bottom of the feet were aligned with a flat surface prior to testing. Two orientations were tested to characterize both the toe and heel

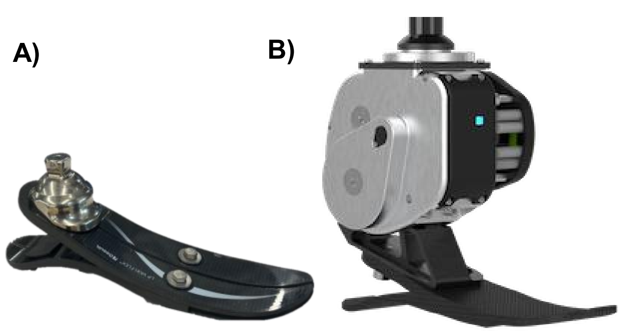

Figure 1. **A)** Ossur Vari-Flex Low Profile carbon fiber prosthetic foot. **B)** Open-Source Ankle attached to a low profile foot.

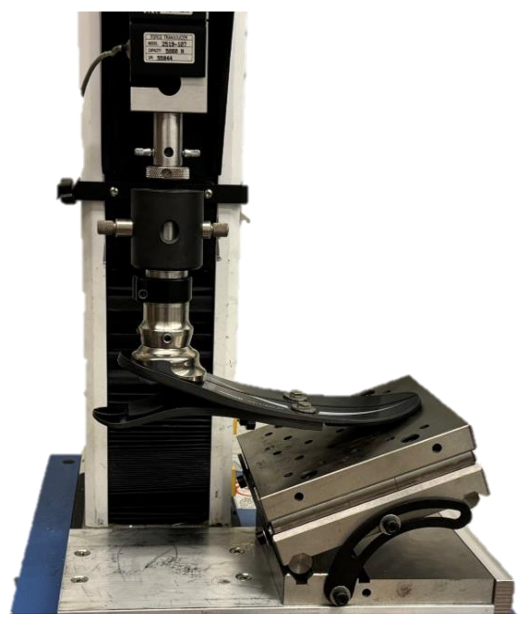

Figure 2. Benchtop mechanical testing setup shown in toe loading condition. The sine plate is set to a 10˚ incline under the toe of the prosthetic foot. The prosthetic foot is connected to the force transducer with a pyramid adapter and custom machined part.

portions of the feet and foot-ankle system. A sine plate was placed under the toe and heel portions at 10° incline and decline to achieve toe-only and heel-only loading, respectively. Three trials were completed for each condition (i.e. category level and OSA gain) in which each trial consisted of loading to 825 N at a rate of 5 mm/s downward into the sine plate. The maximum force was chosen to represent the average adult body weight [26]. Instantaneous force and displacement data throughout loading were recorded by the materials testing system software at 100 Hz.

C. *Analyses*

Linear regression was completed on the lower and upper 50% of the force vs. displacement data above the minimum force threshold of 50 N. Displacement was zeroed at this 50 N. Per published guidelines, 50 N is the recommended minimum threshold for structural testing of prosthetic foot components [27, 28]. The slopes resulting from these linear regressions were averaged across the three trials for toe and heel loading of each condition to characterize stiffness under low- and high-end loading scenarios. Toe and heel stiffnesses were correlated to category level and OSA stiffness gain to produce Pearson correlation coefficient (r) values (Table I). Regression models were also developed to predict effective stiffness during toe- and heel-only loading at low- and high-end loading levels from stiffness category level or ankle stiffness parameter. For standalone foot conditions, a linear model was used. For OSA conditions, a logarithmic model was shown to have the best agreement (Table II).

III. Results

Force vs. displacement data for all trials of the standalone feet and OSA-foot system were shown to be well modeled

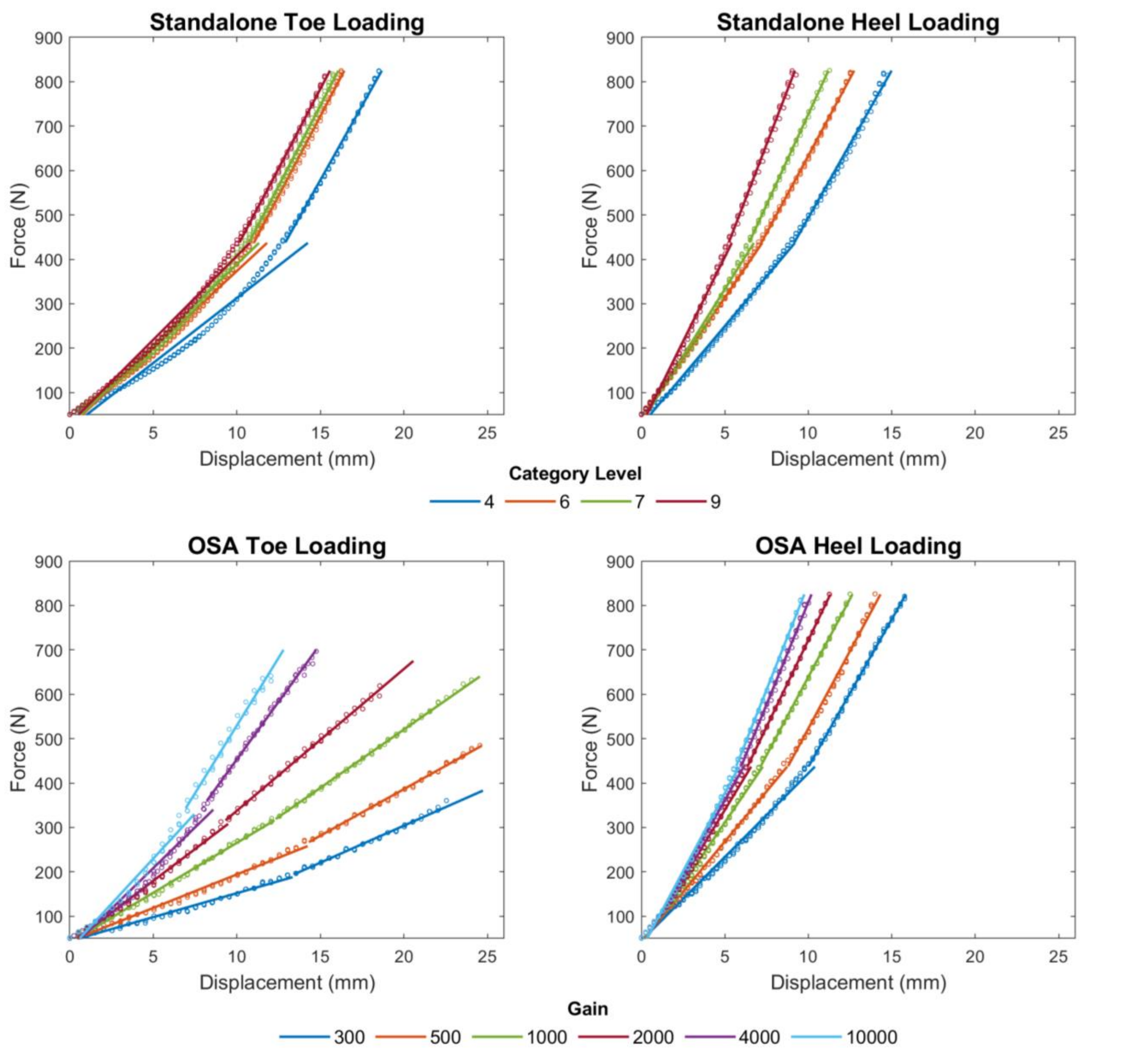

Figure 3. Force vs. displcement data with linear regression models overlaid for the loading of the standalone feet and the Open Source Ankle system. The measured force-displacement data of all three trials is represented by the circular markers. Solid lines represent the mean linear regression of the lower and upper 50% of the data of the three loading trials for each condition.

using a linear fit on the upper and lower half of the data (Fig. 3). The calculated stiffness values and standard deviations for all conditions from this data (Table I and Fig. 4) was sensitive to loading the toe or heel. Toe stiffness was usually less than heel stiffness in all conditions across both altered stiffness category and different OSA gains.

A. Standalone Prosthetic Feet of Varying Stiffness Category

Toe stiffness of the standalone ESAR feet ranged from 29.23 mean (0.06 standard deviation) N/mm at the lowest category level (4) to 37.73 (0.27) N/mm at the highest category level (9) under low-end loads and from 67.03 (0.49) to 71.45 (0.81) N/mm under high-end loads. The heel stiffnesses of the standalone feet ranged from 44.61 (0.38) at the lowest category level to 75.95 (1.29) N/mm at the highest category level under low-end loads and from 66.47 (0.99) to 97.77 (2.09) N/mm under high-end loads. There was a high linear correlation between stiffness and category level for both the low- and high-end heel loading (r = 0.992 and 0.952 respectively). The low-end toe stiffnesses were also correlated with category level (r = 0.941). However, high-end toe stiffness was less correlated with category level (r = 0.755).

B. Open Source Leg Prosthesis

The use of the robotic ankle and category 9 foot augmented the effective stiffnesses during toe loading ranging from 10.85 (0.13) N/mm to 40.78 (1.44) N/mm at low-end loads and from 16.68 (0.59) to 60.98 (0.65) N/mm under high-end loads. Overall, the lowest achieved toe stiffness with the OSA was over 50% less than the toe stiffness of the lowest category level foot. The highest OSA toe stiffness was ~16% less than the toe stiffness of the highest category level foot. Heel stiffnesses using the OSA ranged from 37.70 (0.08) to 70.28 (0.05) under low-end loads and from 66.52 (0.29) to 94.86 (0.06) N/mm under high-end loads. The lowest OSA heel stiffness overall was ~17% less than the lowest category level foot and the highest OSA heel stiffness was ~3% less than the highest category level foot. In addition, a logarithmic model was found to best fit effective stiffness vs. stiffness gain for both toe- and heel-only loading in the low- and high-end load ranges of the OSA (Table II).

IV. Discussion

The purpose of the study was to compare the ability of a robotic ankle prosthesis to augment effective foot-ankle stiffnesses relative to clinically-available standalone prosthetic feet. We hypothesized that control of the powered ankle across its recommended range of stiffness and damping parameters could expand the effective stiffness offered by available passive prosthetic feet. We generally accept this hypothesis at both low-end and high-end loading because the ability to control a robotic ankle in series with a high category prosthetic foot could reduce the effective stiffness of the foot-ankle system and extend beyond the effective stiffness range of this set of commercially-available prosthetic feet in both toe- and heel-only loading conditions. Effective stiffness with the OSA extended beyond the standalone stiffness range in all categories except high-end (50-100% body weight) heel-only loading where the lowest stiffnesses were approximately equal (Table I). For low-end loads (i.e., 0-50% body weight), the stiffnesses achieved with the OSA also covered the standalone range for both toe- and heel-only loading. On the high-end, the device spanned the standalone range of stiffnesses during heel loading but toe stiffnesses were somewhat under this range. In most conditions, the standalone category 9 foot was slightly stiffer than could be achieved with the highest commanded gain on the OSA connected to the category 9 foot. This was expected as the robotic ankle introduces compliance into the system and therefore should never achieve higher stiffness values than the category 9 foot alone. Using a stiffer foot in series with the OSA would likely allow for higher effective stiffnesses of the system if desired.

We also hypothesized that the control of the powered ankle across its range of stiffness and damping parameters would better explain the effective stiffness that is rendered during loading. We generally did not accept this hypothesis since higher linear correlation coefficients (r) between effective stiffness and category level of the standalone feet were observed as compared to controller gains of the robot (Table I). However, the exception is that for loading of the toe only at high-end loading levels, as the control parameters of the OSA were more strongly linearly related to effective stiffness than category level (Table I). In addition, we found that although non-linear for most loading conditions of the OSA (Fig. 4), there is a significant relationship that can be derived (Table II), essentially as a potential calibration relationship

TABLE I. Effectve Stiffness Values: Mean (Std. Dev.)

Standalone (N/mm)					**OSA (N/mm)**				
	Low End		**High End**			**Low End**		**High End**	
Category	***Toe***	***Heel***	***Toe***	***Heel***	***Gain***	***Toe***	***Heel***	***Toe***	***Heel***
4	29.23 (0.06)	44.61 (0.38)	67.03 (0.49)	66.47 (0.99)	300	10.85 (0.13)	37.70 (0.08)	16.68 (0.59)	66.52 (0.29)
6	34.97 (0.39)	55.07 (0.51)	71.62 (0.35)	69.76 (1.18)	500	14.88 (0.21)	45.17 (0.06)	20.97 (0.37)	69.94 (0.18)
7	36.32 (0.33)	59.99 (0.38)	72.38 (0.50)	81.60 (1.18)	1000	20.81 (0.12)	55.24 (0.07)	26.28 (0.60)	70.45 (0.05)
9	37.73 (0.27)	75.95 (1.29)	71.45 (0.81)	97.77 (2.09)	2000	28.62 (0.43)	61.95 (0.11)	31.94 (0.60)	78.06 (0.17)
					4000	36.64 (0.34)	66.74 (0.62)	51.97 (0.76)	92.04 (0.87)
					10000	40.78 (1.44)	70.28 (0.05)	60.98 (0.65)	94.86 (0.06)
r	0.941	0.992	0.755	0.952	r	0.852	0.777	0.923	0.885

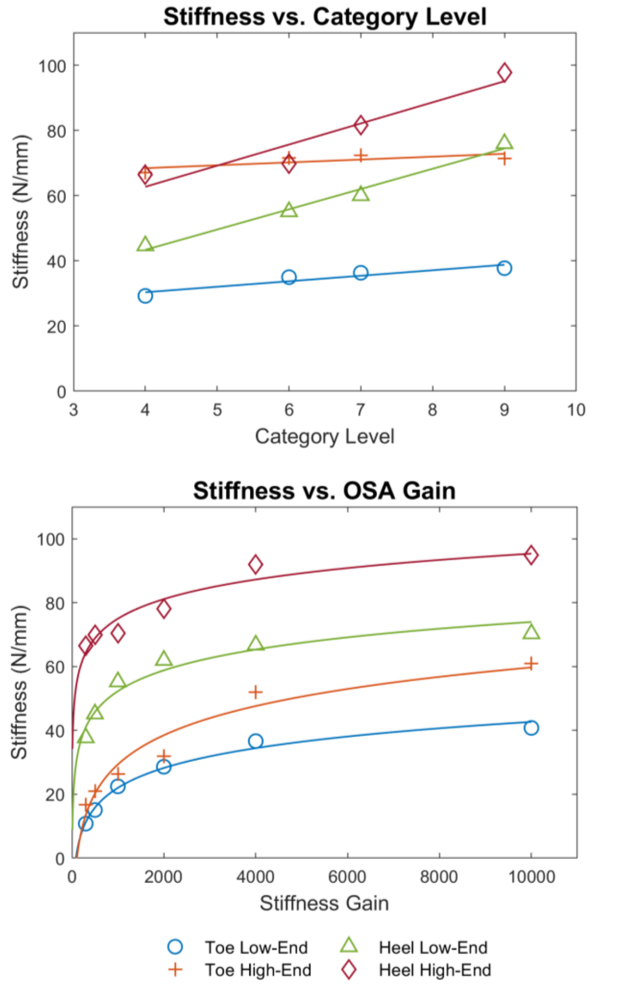

Figure 4. Measured effective toe and heel stiffnesses across category levels of standalone feet for low-end and high-end loads with linear fits (top). Measured effective toe and heel stiffnesses across programmed gains of the Open-Source Ankle for low-end and high-end loads with logarithmic fits (bottom).

for human subject experiments. Once fit with a respective regression model, there was better overall model agreement for the OSA conditions relative to the standalone feet with coefficients of determination being >~0.90 for all loading conditions of the active device, and in most cases >~0.95 (Table II).

A. Implications

The ability of a robotic ankle to offer a potentially wider range and more precise effective stiffnesses to the user may improve user satisfaction. Indeed, previous work has shown that individuals with limb loss can sense small differences in stiffness during walking [29]. Other related work has attempted to address prescription difficulties by using a prosthetic foot emulator in a matching paradigm to replicate toe-only loading (i.e., forefoot) stiffness of different prosthetic foot models, allowing individuals to trial different emulated commercial options while walking [30]. This study differs from the current study in several aspects most notably in the robot actuator (linear as opposed to rotational) and in the different goals of either matching a given commercial device or testing the range of effective stiffness that can be rendered (i.e., the goal of our study). In addition, the current study also characterized effective stiffness during heel-only loading. Importantly, not only could user perception improve with well-matched or well-optimized ankle-foot stiffness, but biomechanical outcomes may also improve. Previous work that systemically varies foot stiffness shows that optimal body support and propulsion can reduce intact limb loading, encourage gait symmetry, allow for increased walking speeds and reduce metabolic cost during ambulation [8, 9].

Another potential benefit of the OSA paradigm that was not specifically tested in our study is the potential for the ability to continuously modulate ankle-foot stiffness during a given stride. Passive prostheses have "fixed" stiffnesses that may be altered with the addition of wedges or shoes, but have a limited range of stiffness alterations and cannot explicitly change stiffness throughout the stride or across other activities. This makes transitioning from level ground walking to ramps and/or stairs challenging and may contribute to the increased fear and incidence of falling, which is prevalent in this population [31]. Continuous or adaptive stiffness modulation, if optimized for specific individuals, could improve these outcomes and contribute to their balance and/or safety. Previous studies have demonstrated the ability to use continuously-variable torque [32] or impedance-based control strategies for joint stiffness or equilibrium angle modulation of powered knees and ankles during the stride and across ambulation modes [33]. More recently, continuous control strategies were tested on this specific device in combination with myoelectric control input [34]. Understanding how effective foot-ankle stiffness is influenced by using the OSA in series with a commercially-available foot such as the Ossur Vari-Flex foot used in the present study is important to future development of control strategies for modulating assistance during the stride, across terrains and between users.

B. Limitations

To characterize these relationships, some assumptions were made. For example, the loading response of the standalone feet and foot-ankle systems was non-linear, especially at lower force values which is consistent with the findings of previous studies [11, 35-37]. Some studies have

TABLE II. EFFECTIVE STIFFNESS ACROSS CATEGORY LEVEL AND GAIN

Stiffness vs. Category Level				
$y = ax + b$		***a***	***b***	R^2
Toe	***Low-End***	1.69	23.60	0.886
	High-End	0.88	64.90	0.570
Heel	***Low-End***	6.22	18.50	0.984
	High-End	6.48	36.81	0.906
Stiffness vs. OSA Gain				
$y = a*log_{10}(x) + b$		***a***	***b***	R^2
Toe	***Low-End***	20.65	-39.92	0.987
	High-End	30.37	-61.78	0.947
Heel	***Low-End***	21.67	-12.72	0.944
	High-End	20.37	13.86	0.899

used linear relationships computed over the entire loading curve to characterize stiffness of prosthetic feet [11, 36, 37]. To account for the non-linearity, we chose to fit linear regressions separately to the lower and upper halves of the loading data. This allowed us to intuitively compare the range of effective stiffnesses of the standalone feet with the robotic ankle-foot system at both low and high loads. Quasi-static mechanical testing of the OSA-foot system in the toe-only condition at low stiffness resulted in dorsiflexion of the foot to the end of the ankle's range of motion. In addition, toe-only loading at high stiffness reached the recommended torque limits of the device. During ambulation, we expect that for either of these cases, there would be a physical interaction of the foot with the hard stop of the ankle, and thus the effective stiffness of the system would be determined only by the properties of the foot attached, and whatever footwear was used. Finally, although not tested, stiffness category levels 1-3 exist for this prosthetic foot model, but were inaccessible at the time of this study. We expect lower effective stiffness values could be achieved for these foot models if available. However, we also expect that these choices of category levels at this foot size are relatively uncommon for most users, and based on the relationships we have found, the floor of the commanded stiffness values of the OSA could be decreased to surpass the lowest stiffness rendered by the standalone feet. Lastly, the quasi-static nature of these comparisons differ from the real and often transient dynamics of gait. Thus, assessing these outcomes during human use is important future work.

C. Conclusions

These findings highlight the complexity of rendering high fidelity effective stiffness of the prosthetic foot-ankle system with standalone or robotic ankle prostheses, as well as the potential influence of clinical prescription on biomechanical outcomes of individuals with limb loss. We found a robotic ankle in series with a high stiffness category level prosthetic foot can be controlled to continuously span most of the profiles of lower category level prosthetic feet, as well as produce mechanical behaviors that extend beyond those capabilities. Prosthetic users could benefit by providing an effective stiffness that is not currently covered in the clinical range and/or is continuously modulated on different terrains, or during different phases of gait.

REFERENCES

[1] L. J. Mengelkoch, J. T. Kahle, and M. J. Highsmith, "Energy costs and performance of transfemoral amputees and non-amputees during walking and running: A pilot study," *Prosthet Orthot Int,* vol. 41, no. 5, pp. 484-491, 2017.

[2] R. J. Zmitrewicz, R. R. Neptune, J. G. Walden, W. E. Rogers, and G. W. Bosker, "The effect of foot and ankle prosthetic components on braking and propulsive impulses during transtibial amputee gait," *Arch Phys Med Rehabil,* vol. 87, no. 10, pp. 1334-9, 2006.

[3] R. Gailey, K. Allen, J. Castles, J. Kucharik, and M. Roeder, "Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use," *J Rehabil Res Dev,* vol. 45, no. 1, pp. 15-29, 2008.

[4] S. J. Mattes, P. E. Martin, and T. D. Royer, "Walking symmetry and energy cost in persons with unilateral transtibial amputations: Matching prosthetic and intact limb inertial properties," *Arch of Phys Med and Rehabil,* vol. 81, no. 5, pp. 561-568, 2000.

[5] R. L. Waters and S. Mulroy, "The energy expenditure of normal and pathologic gait," *Gait & Posture,* vol. 9, no. 3, pp. 207-231, 1999.

[6] S. R. Wurdeman, P. M. Stevens, and J. H. Campbell, "Mobility Analysis of AmpuTees (MAAT I): Quality of life and satisfaction are strongly related to mobility for patients with a lower limb prosthesis," *Prosthet Orthot Int,* vol. 42, no. 5, pp. 498-503, 2018.

[7] N. P. Fey, G. K. Klute, and R. R. Neptune, "Optimization of prosthetic foot stiffness to reduce metabolic cost and intact knee loading during below-knee amputee walking: a theoretical study," *J Biomech Eng,* vol. 134, no. 11, p. 111005, 2012.

[8] N. P. Fey, G. K. Klute, and R. R. Neptune, "Altering prosthetic foot stiffness influences foot and muscle function during below-knee amputee walking: A modeling and simulation analysis," *Journal of Biomechanics,* vol. 46, no. 4, pp. 637-644, 2013.

[9] M. J. Major, M. Twiste, L. P. J. Kenney, and D. Howard, "The effects of prosthetic ankle stiffness on ankle and knee kinematics, prosthetic limb loading, and net metabolic cost of trans-tibial amputee gait," *Clinical Biomechanics,* vol. 29, no. 1, pp. 98-104, 2014.

[10] P. G. Adamczyk, M. Roland, and M. E. Hahn, "Sensitivity of biomechanical outcomes to independent variations of hindfoot and forefoot stiffness in foot prostheses," *Hum Mov Sci,* vol. 54, pp. 154-171, 2017.

[11] N. D. Womac, R. R. Neptune, and G. K. Klute, "Stiffness and energy storage characteristics of energy storage and return prosthetic feet," *Prosthet Orthot Int,* vol. 43, no. 3, pp. 266-275, 2019.

[12] T. R. Ruxin *et al.*, "Comparing forefoot and heel stiffnesses across commercial prosthetic feet manufactured for individuals with varying body weights and foot sizes," *Prosthet Orthot Int,* vol. 46, no. 5, pp. 425-431, 2022.

[13] V. Agrawal, R. S. Gailey, I. A. Gaunaurd, C. O'Toole, A. Finnieston, and R. Tolchin, "Comparison of four different categories of prosthetic feet during ramp ambulation in unilateral transtibial amputees," *Prosthet Orthot Int,* vol. 39, no. 5, pp. 380-389, 2015.

[14] H. Houdijk, D. Wezenberg, L. Hak, and A. G. Cutti, "Energy storing and return prosthetic feet improve step length symmetry while preserving margins of stability in persons with transtibial amputation," *J Neuroeng Rehabil,* vol. 15, no. Suppl 1, p. 76, 2018.

[15] F. C. Sup and M. Goldfarb, "Design of a Pneumatically Actuated Transfemoral Prosthesis," in *ASME 2006 Intl Mech Eng Congress and Expo*, 2006, vol. Dynamic Systems and Control, Parts A and B, pp. 1419-1428.

[16] H. M. Herr and A. M. Grabowski, "Bionic ankle-foot prosthesis normalizes walking gait for persons with leg amputation," (in eng), *Proc Biol Sci,* vol. 279, no. 1728, pp. 457-64, Feb 7 2012.

[17] J. Geeroms, L. Flynn, R. Jimenez-Fabian, B. Vanderborght, and D. Lefeber, "Ankle-Knee prosthesis with powered ankle and energy transfer for CYBERLEGs α-prototype," (in eng), *IEEE Int Conf Rehabil Robot,* vol. 2013, p. 6650352, Jun 2013.

[18] M. Grimmer *et al.*, "A powered prosthetic ankle joint for walking and running," *Biomed Eng Online,* vol. 15, no. Suppl 3, p. 141, Dec 19 2016.

[19] A. F. Azocar, L. M. Mooney, J.-F. Duval, A. M. Simon, L. J. Hargrove, and E. J. Rouse, "Design and clinical implementation of an open-source bionic leg," *Nature Biomedical Engineering,* vol. 4, no. 10, pp. 941-953, 2020.

[20] S. Au, M. Berniker, and H. Herr, "Powered ankle-foot prosthesis to assist level-ground and stair-descent gaits," *Neural Networks,* vol. 21, no. 4, pp. 654-666, 2008.

[21] F. Sup, H. A. Varol, and M. Goldfarb, "Upslope Walking With a Powered Knee and Ankle Prosthesis: Initial Results With an Amputee Subject," *IEEE Trans on Neural Sys and Rehabil Eng,* vol. 19, no. 1, pp. 71-78, 2011.

[22] N. T. Pickle, A. K. Silverman, J. M. Wilken, and N. P. Fey, "Statistical analysis of timeseries data reveals changes in 3D segmental coordination of balance in response to prosthetic ankle power on ramps," *Scientific Reports,* vol. 9, no. 1, p. 1272, 2019.

[23] A. M. Simon *et al.*, "Configuring a powered knee and ankle prosthesis for transfemoral amputees within five specific ambulation modes," *PLoS One,* vol. 9, no. 6, p. e99387, 2014.
[24] L. J. Hargrove *et al.*, "Intuitive Control of a Powered Prosthetic Leg During Ambulation: A Randomized Clinical Trial," *JAMA,* vol. 313, no. 22, pp. 2244-2252, 2015.
[25] Dephy, Inc., "Impedance Control Gains." https://dephy.com/start/#impedance-control-gains (accessed June 5, 2024).
[26] C. C. Fryar, MD. Gu, Q. Afful, J. Ogden, CL. "Anthropometric reference data for children and adults: United States, 2015–2018," National Center for Health Statistics. *Vital Health Sta*t vol. 3, no. 46, pp. 8-9, 2021.
[27] ISO 10328:2016. Prosthetics-structural testing of lowerlimb prostheses-requirements and test methods.
[28] ISO 22675:2016. Prosthetics-Testing of ankle-foot devices and foot units.
[29] M. K. Shepherd, A. F. Azocar, M. J. Major, and E. J. Rouse, "Amputee perception of prosthetic ankle stiffness during locomotion," *J of NeuroEng and Rehabil,* vol. 15, no. 1, p. 99, 2018.
[30] E. G. Halsne, C. S. Curran, J. M. Caputo, A. H. Hansen, B. J. Hafner, and D. C. Morgenroth, "Emulating the Effective Ankle Stiffness of Commercial Prosthetic Feet Using a Robotic Prosthetic Foot Emulator," *J Biomech Eng,* vol. 144, no. 11, 2022.
[31] W. C. Miller, M. Speechley, and B. Deathe, "The prevalence and risk factors of falling and fear of falling among lower extremity amputees," *Arch of Physl Med and Rehabil,* vol. 82, no. 8, pp. 1031-1037, 2001.
[32] T. Lenzi, L. Hargrove, and J. Sensinger, "Speed-Adaptation Mechanism: Robotic Prostheses Can Actively Regulate Joint Torque," *IEEE Robotics & Automation Magazine,* vol. 21, no. 4, pp. 94-107, 2014.
[33] N. P. Fey, A. M. Simon, A. J. Young, and L. J. Hargrove, "Controlling Knee Swing Initiation and Ankle Plantarflexion With an Active Prosthesis on Level and Inclined Surfaces at Variable Walking Speeds," *IEEE J Transl Eng Health Med,* vol. 2, 2014.
[34] R. R. Posh, J. P. Schmiedeler, and P. M. Wensing, "Finite-State Impedance and Direct Myoelectric Control for Robotic Ankle Prostheses: Comparing Their Performance and Exploring Their Combination," *IEEE Trans Neural Syst Rehabil Eng,* vol. 31, pp. 2778-2788, 2023.
[35] H. W. L. Van Jaarsveld, H. J. Grootenboer, J. De Vries, and H. F. J. M. Koopman, "Stiffness and hysteresis properties of some prosthetic feet," *Prosthet Orthot Int,* vol. 14, no. 3, pp. 117-124, 1990.
[36] M. D. Geil, "Energy Loss and Stiffness Properties of Dynamic Elastic Response Prosthetic Feet," *JPO,* vol. 13, no. 3, pp. 70-73, 2001.
[37] M. J. Major, J. Scham, and M. Orendurff, "The effects of common footwear on stance-phase mechanical properties of the prosthetic foot-shoe system," *Prosthet Orthot Int,* vol. 42, no. 2, pp. 198-207, 2018.