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Preferred Ankle Stiffness of a Variable-Stiffness Prosthesis Across Five Activities

Nicholas J. Pett[1,2,3], Nundini D. Rawal[1,2,3], Varun S. Shetty[1,3], Leslie Wontorcik[4], Elliott J. Rouse[1,2,3]

***Abstract*— Common existing ankle-foot prostheses are carbon-composite springs that cannot vary their mechanics to meet the biomechanical demands of differing activities. This lack of adjustment leads to discomfort and compensatory movements, which ultimately lower mobility and quality of life for people with amputation. Emerging variable-stiffness ankle prostheses have the potential to address this challenge by intelligently changing their stiffness between activities. However, the appropriate stiffness for each user and activity remains an open question. One potential strategy is to modulate the stiffness to the preferred settings of the wearer. Prosthesis users are sensitive and consistent in the selection of their preferred stiffness settings. However, previous work has primarily focused on level walking; a detailed analysis of preferred stiffness variation across multiple activities remains unexplored. In this preliminary study, we quantified the preferred prosthetic ankle stiffness of four participants with below-knee amputation across five different activities. Preferred stiffness settings varied substantially between activities. When averaged across participants, the preferred ankle stiffness differed between activities by 31.8% of the preferred stiffness for level walking. This difference reflects changes in stiffness that span up to four categories of commercial prostheses. In addition, prosthetic ankle kinematics varied across activities and stiffness, with mean peak dorsiflexion reaching as great as 15.3° during incline walking. The differences in preferred stiffness across activities, coupled with the corresponding changes in kinematics, underscore the potential of user preference and its implications in variable-stiffness prostheses.**

I. INTRODUCTION

One challenge to the widespread impact of modern prostheses is their ability to meet the biomechanical demands of differing activities. Differing activities, such as stair descent or incline walking, require differing efforts and motions from the lower limbs [1], [2]. The standard of care for people with below-knee amputation is passive prostheses [3], [4], known as Energy Storage and Return (ESR) feet, which have fixed mechanics that cannot be adjusted following manufacture (or can be altered slightly through the use of plastic “wedges” [5]). This forces people with amputation to use fixed prosthesis mechanics irrespective of the biomechanical demands specific to each activity. The inappropriate mechanics may, in part, account for the mobility challenges faced by these individuals. Common deficits include asymmetric gait biomechanics [6], [7], compensatory motions [6], [8], and greater exertion during ambulation [9], [10]. Consequently, over 60% of lower-limb amputees are “not sufficiently active” [11]. This reduction in mobility lowers quality of life and often leads to secondary conditions, such as heart disease and depression [12]–[14]. One important part of the solution to these challenges is the advancement of intelligent prostheses that can adjust their mechanical behavior to meet the constantly-evolving biomechanical demands of community life.

A new class of prostheses is being developed that enables step-to-step variation of prosthetic ankle stiffness; this type of prosthesis allows for manual or automatic stiffness adjustments between steps or activities. Among the first of these prostheses was a quasi-passive pneumatic foot that stored energy in a pneumatic cylinder during dorsiflexion and could adjust its equilibrium position via a solenoid valve [15]. This design was succeeded by the Variable-Stiffness Prosthetic Ankle-Foot (VSPA Foot), which utilizes a cam-based transmission and a simply-supported leaf spring with an adjustable support location to modulate ankle stiffness [16]. Glanzer and Adamczyk’s Variable Stiffness Foot uses a similar principle that includes an adjustable fulcrum and leaf spring [17]. The Variable Stiffness Ankle developed by Lecomte et al. also relies on a leaf spring with an adjustable support, but the spring is vertically-mounted and this ankle unit can be attached to a commercially-available ESR foot [18]. The ELSA foot, developed by Heremans et al., implements a lockable, stiffness-adjustable parallel elastic element (nylon rope); an adjustable clamp modulates rope length and consequently stiffness [19]. Most recently, Rogers-Bradley et al. developed a prosthesis that allows for discrete stiffness adjustments by locking relative motion between varying combinations of parallel leaf springs [20]. Although specific implementation and available stiffness range vary, the underlying objective of these technologies is to enable prosthesis stiffness to be individualized to best accommodate different users and activities.

To intelligently adjust stiffness for these new prostheses, we must have criteria by which optimal stiffness can be selected. Quantified user preference has recently emerged as a potential “meta-criterion” that can be quickly evaluated in a clinical setting, serving as an alternative to objectives that are time, equipment, and labor-intensive [21]–[23]. For variable-stiffness prostheses user preference is a feasible criterion that facilitates interaction between the clinician and patient [21]–[23], building on their existing communication in the prescription process. In addition, preference has shown promise stemming from its downstream biomechanical effects. In

[1]Neurobionics Lab, University of Michigan, Ann Arbor, 48109, USA
[2]Department of Mechanical Engineering, University of Michigan, Ann Arbor, 48109, USA
[3]Department of Robotics, University of Michigan, Ann Arbor, 48109, USA (npett@umich.edu, ejrouse@umich.edu)
[4]Department of Physical Medicine and Rehabilitation, Michigan Medicine, University of Michigan Orthotics and Prosthetics Center, Ann Arbor, MI, 48104, USA

particular, it has been demonstrated that kinematic symmetry at the ankle joint was greatest at the users' preferred stiffness [22].

Abstract perceptions such as comfort, smoothness of motion, and balance are difficult to optimize using traditional metrics [21]–[23]; user preference enables the wearer to balance these objectives, culminating in prosthesis behavior that is personalized to the wearer and activity. If users are reliable in the perception of their preference, clinicians can put more emphasis on their input when evaluating potential prostheses. Past research indicates that users are more consistent than clinicians when evaluating their preferred prosthesis stiffness [24], further underscoring its potential.

While user preference has shown promise for selecting prosthesis stiffness, existing work has principally assessed the metric during level-ground walking. There is some indication that preferred stiffness varies on inclines and stairs [16], [25], and that changing stiffness settings may yield kinematic [16], [26] or metabolic [27] benefits. A gap remains in the formal quantification of preferred ankle stiffness for non-level activities. Given that these activities account for a considerable portion of community ambulation [28], addressing this gap is important. With this information, individuals' preferred stiffness values can be used to adjust the settings of variable-stiffness prostheses in real time, upon detecting that the wearer has transitioned to a new activity. In this preliminary study, we use the VSPA Foot to quantify the preferred ankle-foot stiffness of four individuals with below-knee amputation across level walking, inclined ramp walking, declined ramp walking, stair ascent, and stair descent. We measure how preferred prosthetic ankle stiffness varies with activity, and examine the associated changes in ankle kinematics. Based on the known variation in biomechanics between locomotion activities [1], [2], we hypothesize that there will be significant variation in preferred ankle stiffness between activities. We anticipate that changes greater than the perceptual threshold (i.e. Just-Noticeable Difference, 7.7% [21]) would be clinically meaningful.

II. METHODS

A. Experimental Design

1) VSPA Foot: This experiment was conducted using the Variable-Stiffness Prosthetic Ankle-Foot (VSPA Foot, originally described in Shepherd and Rouse, 2017, and used in several other studies of variable-stiffness prostheses [16], [21], [22], [24]). The VSPA Foot uses a titanium leaf spring with a cam-based transmission and a motor-driven simple support that can be repositioned between steps to allow for automatic adjustment of ankle stiffness during locomotion.

The device features a customizable cam profile within the cam-based transmission able to achieve a desired torque-angle relationship at the ankle joint; in this work we used a cam with a linear torque-angle relationship, which can achieve a stiffness range of 200 to 1060 $\frac{Nm}{rad}$ (Fig. 1). The use of a linear torque-angle curve allows for more direct comparison with previous VSPA Foot studies [21], [22], [24] and enables straightforward reporting of joint stiffness.

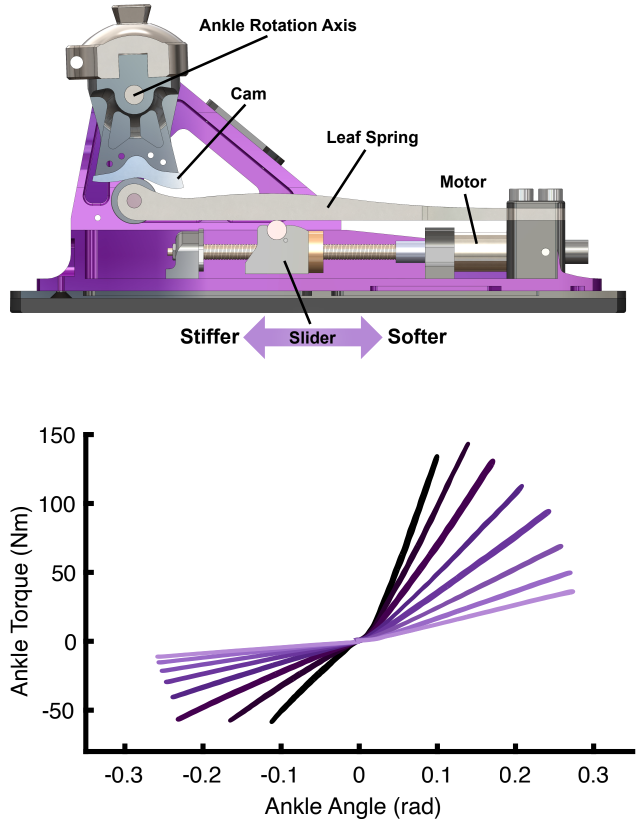

Fig. 1. The VSPA Foot allows for variable ankle stiffness using a leaf spring and a cam-based transmission. Changing the position of the support under the leaf spring (i.e., the slider) allows the torque-angle curve to be shifted to increase or decrease the stiffness of the ankle joint. The linear torque-angle relationship seen here (bottom) is determined by the cam profile, which can be programmed before manufacture.

Nominal stiffness values reported in this paper refer to the dorsiflexion stiffness of the device. The ratio of plantarflexion stiffness to dorsiflexion stiffness in this cam is 1:3, which is consistent with prior work [21], [22], [24]. The selection of a lower plantarflexion stiffness stemmed from feedback with prosthesis users and minimizing prosthesis "foot slap."

Due to the low-power motor used for stiffness adjustment, the VSPA Foot is intended to adjust its stiffness during the swing phase of gait, when the spring is unloaded. For this study, we implemented a Bayesian gait state estimator based on the work of Medrano et al. to identify stance and swing phase in level, incline, and decline walking activities [29]. This estimator employs an Extended Kalman Filter, which uses real-time foot orientation in the sagittal plane from an inertial measurement unit (IMU) (see Mechatronics section). Real-time gait phase information was obtained using the IMU measurements and a gait model that was trained using previously collected VSPA Foot biomechanical data from Clites et al. [22], [29]. Heuristic swing detection based on ankle angles and foot velocity was employed for stair ascent and descent.

The design of the VSPA Foot facilitates studies of user experience. The VSPA Foot's step-to-step ankle stiffness adjustment allows the wearer to seamlessly experience a range of prosthesis mechanical behaviors, eliminating the inconvenience of donning and doffing multiple devices. This efficient transition between mechanical behaviors minimizes time between options, allowing users to more quickly identify their preferences. In addition, the VSPA Foot is lightweight (1.1 kg), comparable to some commercial ESR feet (0.5 to 0.9 kg [30]–[32]) and lighter than emerging microprocessor-controlled ankles (e.g., Ottobock Meridium, 1.3 kg without footshell [33]).

2) Participants: Four people with unilateral, transtibial amputation (2 male, 2 female) participated in this study. Participants ranged in age from 35 to 70 years old (51.8 ± 14.4) and in weight from 83.4 to 108.8 kg (93.0 ± 11.7). All participants were required to be community ambulators, based on self-reported mobility in a pre-screening questionnaire. In addition, participants were at least nine months post-amputation and had no injuries that would affect gait or prevent them from walking for up to 15 minutes continuously. Each participant provided written, informed consent before testing began. This human subject study was approved by the Institutional Review Board for the University of Michigan Medical School.

3) Acclimation to Lab Environment and Prosthesis: The VSPA Foot was fitted to each participant by a certified prosthetist, and alignment was tuned by walking overground as is standard in the clinic. Participants were given time to walk on the treadmill in their take-home prosthesis as well as in the VSPA Foot to get familiar with the setting and the new device. Formal testing did not begin until participants confirmed that they were acclimated to the novel prosthesis.

B. Preferred Stiffness

During preferred stiffness testing, participants donned an electronic dial on a belt around the waist. Rotating the dial enabled the wearer to adjust the stiffness of the VSPA Foot. The dial allowed for infinite rotation and had no absolute reference, so that participants could not identify a specific orientation correlated with their preferred stiffness. At the extrema of the stiffness range of the VSPA Foot, the dial could rotate but stiffness was saturated. Participants could adjust the dial at any point in the gait cycle, and stiffness changes were made during the next identified swing phase. Dial operation was discussed with participants prior to testing, until they clearly understood its functionality.

Preference testing consisted of five activities, progressing through level walking, incline walking, decline walking, and finally stair ascent and descent. The order of the activities was fixed for all participants due to practical aspects of running the study. For each activity, participants were asked to complete seven preference identification trials. After each trial, the starting stiffness was reseeded to prevent the user from repeating the same pattern of dial adjustment. Stair testing was conducted on a separate day to limit the impact of participant fatigue[1]. As a safety precaution, participants donned a safety harness during treadmill-based testing and walked with a gait belt, accompanied by a member of the research team, during stair testing.

Once setup was complete and participants understood how to adjust the stiffness of the VSPA Foot, the following process was conducted for each activity:

1) Identification of Preferred Walking Speed: For each treadmill-based activity (level, incline, and decline walking), participants walked at a fixed, self-selected pace. This decision was made to be consistent with literature [22] and to make the study possible for a broader population of ambulators. The treadmill was initially set to a speed based on the participant's reporting of their own mobility. The participant then walked and instructed the research team to either increase or decrease speed until preferred speed was identified. This process typically took 30 to 90 seconds for level walking, and was quicker for subsequent activities. Stair testing was self-paced on a static staircase to ensure participant safety and comfort with the activity.

2) Preference Selection: Once walking speed was established, participants completed the specified locomotion activity while adjusting the stiffness of the prosthesis with the dial. Participants were prescribed a starting stiffness, and then encouraged to explore the full stiffness range, from "uncomfortably soft" to "uncomfortably stiff" during each trial. They were informed verbally if they had saturated at the maximum or minimum stiffness. Preference identification was self-paced to permit participants to explore stiffness settings for as long as they desired (mean time to convergence was 53.7 ± 22.8 seconds). When the participant expressed that they had found their preferred stiffness, the treadmill was stopped, and the individual preference trial concluded. Each participant successfully completed seven preference identification trials for each activity, with the exception of Participant 1, who only completed six incline trials due to a participant-reported complication. Participants were allowed to rest as long as desired between trials to avoid fatigue.

For each locomotion activity, there was a predetermined starting stiffness for the first trial. For level walking, this starting stiffness was the mean preferred stiffness for level walking at a self-selected pace obtained from Clites et al., 2021 ($634.4\frac{Nm}{rad}$) [22]. For incline and decline walking, the starting stiffness was the participant's own mean preferred stiffness from their level walking trials. For stair ascent and descent, the participant's first stiffness was their mean preferred stiffness from incline and decline walking, respectively. In subsequent trials for each activity, the starting stiffness was pseudo-randomly re-seeded by ±25% from the preferred stiffness for the previous trial. In some instances this re-seeding exceeded the maximum stiffness of the device, in which case the starting stiffness was set as the maximum.

[1]This is an example of the practical considerations that motivated the fixed order of activities. Scheduling of the stair ascent and descent trials needed to be simultaneous, and they were placed on a separate day to limit any impact of fatigue.

C. Locomotion Activities

1) Level Walking: Baseline user preferences were established on a level, split-belt treadmill (Bertec, Columbus, OH, USA). This allowed participants to first explore their preferred ankle stiffness in an activity with which they were most experienced. During all activities, participants were allowed to hold the handrails if they desired.

2) Incline and Decline Walking: Incline and decline testing were conducted on the treadmill at a 4.7º slope. This inclination was chosen as the maximum angle the treadmill could achieve without exceeding the ADA limit for a running slope (1:12, approximately 4.76º) [34]. By using a slope that people with amputation are likely to encounter in the community, our results are more easily translated out of the laboratory.

3) Stair Ascent and Descent: Stair testing was conducted on a large, multi-flight staircase in the Ford Motor Company Robotics Building (2505 Hayward St., University of Michigan, Ann Arbor, MI, 48109, USA). This staircase encompasses 6 flights with 19 steps in the first flight and 12 steps in subsequent flights. Each step has a height of 178 mm and a depth of 305 mm. Participants were encouraged to use as many flights as necessary to establish a preference in ascent or descent, and were asked not to adjust the stiffness of the device or consider the sensation of the device as they traversed the landings between successive flights. In general, preferences were established in 1 to 2 flights of stairs. To most efficiently complete testing and limit participant fatigue, participants continued to ascend for 2 to 3 consecutive trials (the quantity depended on how many flights the individual used to select their preference) before reversing to complete 2-3 consecutive descent preference trials. After selecting a preference for a trial, participants stopped at the next landing to rest and let the research team set the starting stiffness for the following trial. This alternation of ascents and descents continued until seven trials had been completed for each activity. The use of a static staircase assured participants could stop at any point if needed, improving their comfort and more accurately reflecting the manner in which they traverse stairs day-to-day. All participants chose to hold onto the handrail with one hand while on the stairs (Fig. 2).

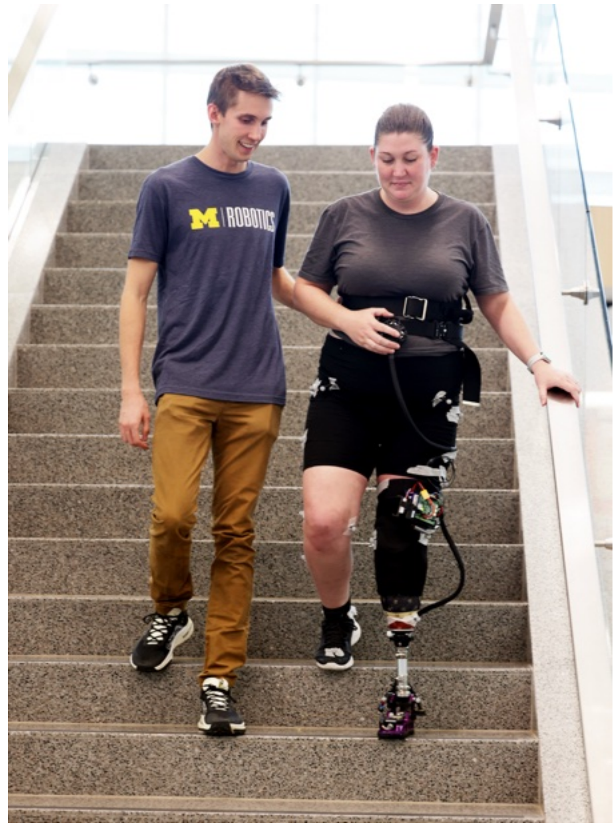

Fig. 2. A stair descent preference trial. The participant descends the staircase, accompanied by a researcher holding onto her gait belt. As she descends she rotates the dial, which initiates a slider movement to alter the stiffness of the prosthetic ankle between steps.

D. Mechatronics

Several mechatronic systems were implemented to facilitate control of the VSPA Foot. As mentioned above, an IMU (model: 3DM-CX5-AR, MicroStrain, Williston, VT, USA) provided data at 190 Hz via UART at 115.2 kbaud; the IMU was affixed to the rigid frame of the VSPA Foot and was used for gait state estimation. The motor used to actuate the simple support beneath the leaf spring was commanded using two brushed motor drivers operating in parallel (model: BD65496MUV, ROHM Semiconductor, Kyoto, Japan). A 14-bit absolute magnetic angle encoder (model: AS5048B, AMS, Premstaetten, Austria) was used to track the position of the dial as participants explored stiffness settings. The dial angle was acquired at 190 Hz via I^2C communication, and mapped linearly to slider position to control the stiffness of the prosthesis.

In addition to control of the foot, prosthetic ankle kinematic data were recorded during all trials to investigate the relationship between kinematics and preferred stiffness across activities. An identical magnetic angle encoder, positioned on the ankle rotational axis of the VSPA Foot, was used to record ankle kinematic data at 190 Hz via I^2C communication. The zero angle was defined as the average angle of the VSPA Foot during swing phase. Ankle encoder and IMU data were post-processed using a second-order Savitzky-Golay filter with a window length of ten samples. All data analyses were completed in MATLAB (MathWorks, Natick, MA, USA) and Python.

These systems were all operated via an onboard, single-board computer (model: Raspberry Pi 4B, Raspberry Pi Foundation, Cambridge, UK), with which the research team could communicate remotely using an SSH connection. A custom printed circuit board provided an interface between the sensors and the single-board computer. These two boards were contained in an additively manufactured housing (material: Onyx, Markforged, Waltham, MA, USA) and affixed to the prosthetic socket of the participant (see Fig. 2). The entire electronic module, including boards and sensors, was powered using an 11.1 V, 1300 mAh lithium polymer battery

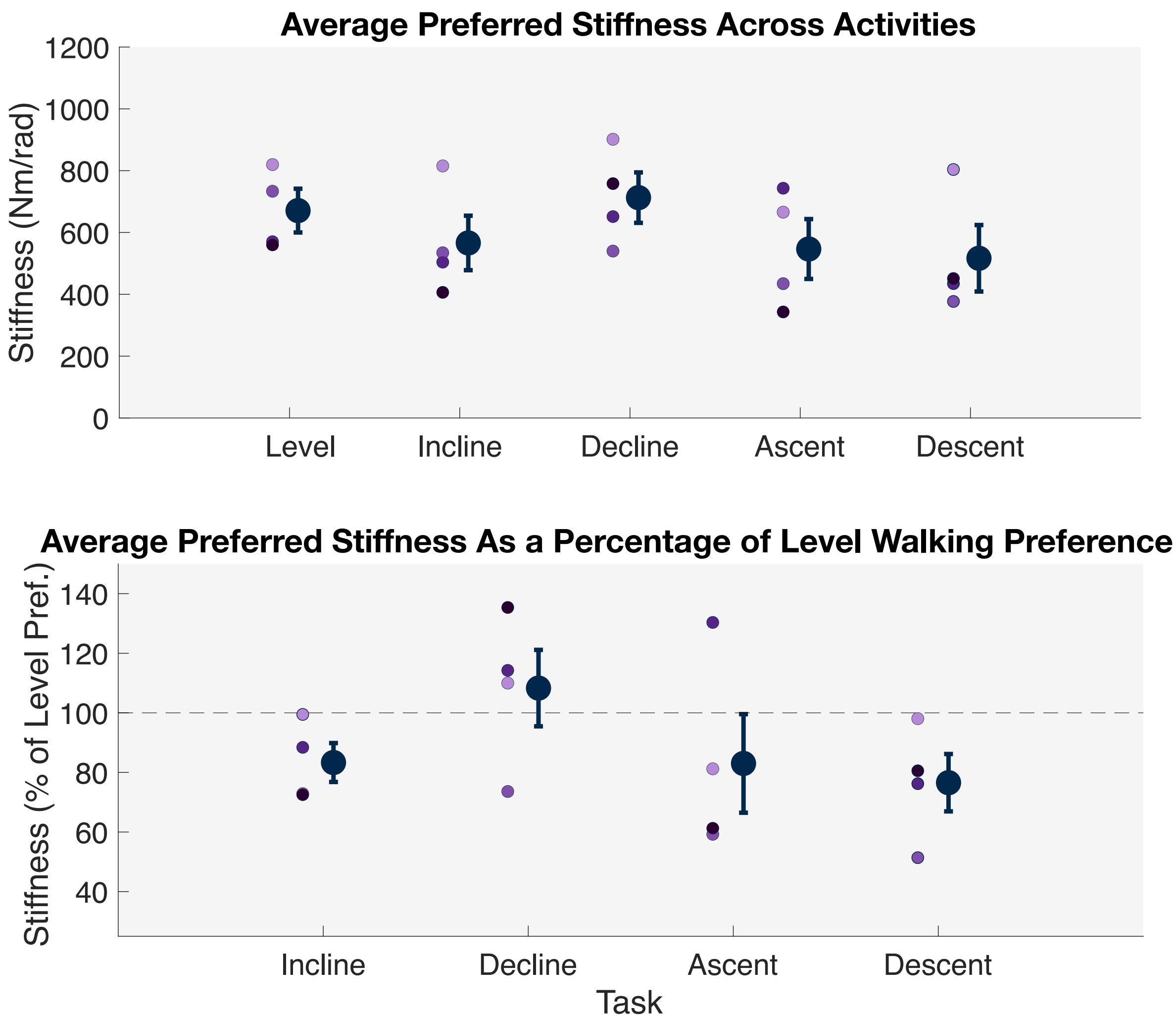

Fig. 3. Average preferred ankle stiffness for each of five locomotion activities (top) and level-walking-normalized preferred ankle stiffness for the incline, decline, ascent, and descent activities (bottom). For both graphs, the purple points depict the individual participant average preferred stiffness for each activity (each shade denotes a different participant), while the larger blue points depict the inter-participant average with standard error bars. Level-walking-normalized preferred ankle stiffness was $83.3 \pm 6.5\%$ for incline, $108.3 \pm 12.8\%$ for decline, $83.0 \pm 16.5\%$ for ascent, and $76.6 \pm 9.6\%$ for descent.

(model: Nano-Tech 1300mAh 25-50C, Turnigy Power Systems, Hong Kong). The software used to control the VSPA Foot was written in Python 3.9.

III. RESULTS

A. *Preferred Stiffness*

On average, preferred ankle stiffness for each of incline walking, decline walking, stair ascent, and stair descent was measurably different from the preferences for level walking. The average preferred ankle stiffness was greatest for decline walking ($712.8 \pm 81.6 \frac{Nm}{rad}$), followed by level walking ($670.9 \pm 70.8 \frac{Nm}{rad}$), incline walking ($566.2 \pm 88.1 \frac{Nm}{rad}$), stair ascent ($546.6 \pm 96.8 \frac{Nm}{rad}$), and stair descent ($516.6 \pm 107.6 \frac{Nm}{rad}$). Relative to level walking, individuals on average preferred 8.3% stiffer settings during decline walking, and preferred less stiff settings during incline walking, stair ascent, and stair descent by 16.7%, 17.0%, and 23.4%, respectively (Fig. 3).

Between participants, ankle stiffness preferences varied widely within an activity, while within-participants preferences generally remained consistent within each activity. Participants' preferences were more closely aligned in some activities than others. We saw the least variation in level walking, where the mean difference in preferred stiffness between two participants was $157.0 \pm 95.9 \frac{Nm}{rad}$. We observed greater variance for stiffness preferences during decline walking (mean difference of $198.6 \pm 98.7 \frac{Nm}{rad}$), incline walking ($209.6 \pm 145.9 \frac{Nm}{rad}$), stair descent ($216.2 \pm 185.2 \frac{Nm}{rad}$), and stair ascent ($238.6 \pm 131.0 \frac{Nm}{rad}$). Within-participants, some individuals were more consistent in their preferences. Participants 1-3 had within-participants coefficients of variation (CV) ranging from 6.0% to 18.0% for all activities, while Participant 4 had CV values as large as 32.0% for individual activities. The mean within-participants CV across all activities was $14.0 \pm 3.7\%$.

In addition to the variation seen in preferred stiffness

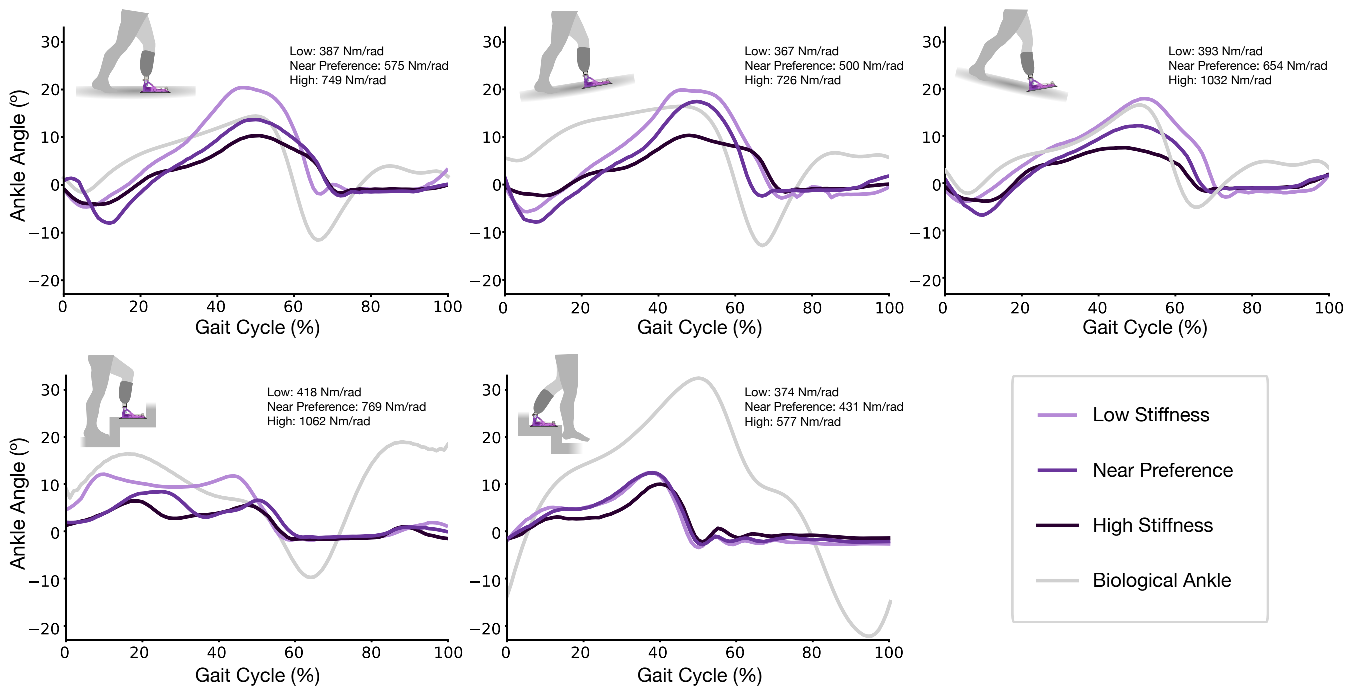

Fig. 4. Ankle angles across all five tested activities for a representative participant (Participant 2). The purple lines show the ankle angle trajectory for gait cycles below, near, and above preferred stiffness for each activity; the grey line shows the average biological ankle angle [35]. Across all activities, the participant's ankle range of motion was inversely related to prosthetic ankle stiffness.

between activities, the consistency of preference selections varied with activity as well. As a group, the four participants were least consistent during stair descent, with a mean within-participants CV of $17.6 \pm 10.4\%$. This was followed by stair ascent ($15.3 \pm 6.3\%$), decline walking ($13.5 \pm 3.4\%$), and level walking ($11.7 \pm 4.9\%$). Incline walking ankle stiffness preferences were most consistent with a mean within-participants CV of $11.7 \pm 4.3\%$.

B. Kinematics and Gait Speed

Clear distinctions were evident in ankle kinematics between different locomotion activities. The mean peak ankle dorsiflexion varied across activities, with the highest average observed during incline walking ($15.3 \pm 3.8°$), followed by level walking ($14.1 \pm 3.1°$), decline walking ($13.6 \pm 4.4°$), and stair descent ($12.2 \pm 5.6°$). The minimum average peak dorsiflexion was observed during stair ascent ($10.2 \pm 2.3°$). This between-activity variability in peak dorsiflexion is shown for a representative participant (Participant 2) in Fig. 4.

Self-selected speeds for level, incline, and decline walking were obtained to set the pace for each activity. Participant self-selected speeds were fastest on level ground ($0.69 \pm 0.20\frac{m}{s}$) and slowest during incline walking ($0.57 \pm 0.16\frac{m}{s}$), when averaged across all participants. Mean self-selected speed for decline walking was $0.68 \pm 0.20\frac{m}{s}$.

IV. DISCUSSION

A. Implications of Results

The objective of this preliminary study was to understand ankle stiffness preferences of below-knee amputees across differing, common activities. To this end, we leveraged the VSPA Foot, a variable-stiffness prosthesis that allows individuals to adjust the prosthesis stiffness between each step. Four participants with unilateral, transtibial amputation were recruited to walk on level, inclined, and declined treadmill configurations at a constant self-selected pace. Additionally, they were tasked with ascending and descending a static staircase. For each of these activities, participants selected their preferred prosthetic ankle stiffness seven separate times. We present initial findings on user preferences and the associated kinematic changes across activities that accompany ambulation with a variable-stiffness prosthesis.

Prosthesis users prefer differing ankle-foot stiffness across activities, highlighting the need for prostheses that can continuously vary their mechanical behavior. The mean preferred stiffness varied substantially across activities; the greatest difference was between decline walking and stair descent, where the preferences differed by 31.8% of the preferred stiffness during level walking. The differences in preferred ankle stiffness between level walking and all other activities exceeded the perceptual threshold of 7.7%. The implication of exceeding this threshold is that prosthesis users can notice that their feet feel "too stiff" or "too soft" for activities other than walking on level ground. These findings should be examined in the context of existing commercial ankle-foot prostheses. The changes in preferred stiffness between activities would necessitate changes to the prosthesis "category," a manufacturer-assigned numeric value used to grossly represent the prosthesis stiffness. Typically, prosthesis users are only able to use a single prosthesis with a fixed category chosen based on the user's weight, activity level, and preference. According to data from Womac et al., an increase by one unit on these scales corresponds to a 12.5% change in maximum stiffness on average [5]. Thus, the aforementioned 31.8% difference between decline walking and stair ascent suggests that people with amputation would, on average, prefer prostheses that are different by 2 - 3 stiffness categories for these two activities. Moreover, when examining preferred stiffness variation within individuals, the average within-participants difference between the most and least stiff activity preferences was $51.4 \pm 18.7\%$ of their preferred stiffness for level walking $(329.0 \pm 75.9 \frac{Nm}{rad})$[2]. This difference is equivalent to a change of approximately four prosthesis categories. These clear distinctions underscore the promise of prostheses that are able to vary stiffness across activities and users. Conventional ESR prostheses, which are inherently constrained to a single stiffness category, are unable to meet this need. Further, the broad variation in preferred stiffness between activities highlights the need for variable-stiffness prostheses with substantial stiffness ranges.

Our results agree with previous studies investigating user-preferred prosthesis stiffness during walking. The average preferred ankle stiffness found in this study for level walking, $670.9 \frac{Nm}{rad}$, is comparable but greater than that found by Shepherd et al. $(504.0 \frac{Nm}{rad})$ and Clites et al. $(634.4 \frac{Nm}{rad}$ at self-selected walking speed) [21], [22]. In addition, the mean within-participants coefficient of variation (CV) indicates more consistent level walking preferences in this study $(11.7 \pm 4.9\%)$ compared to the CV found by Shepherd et al., 2017 $(14.0 \pm 1.7\%)$ [21], but less consistency than was found by Shepherd et al., 2020 (5.6 ± 1.2) [24]. Earlier investigation of preferred stiffness for ramps and stair tasks found users may prefer softer settings for non-level activities [16], and that preferred stiffness on inclines varies widely between individuals [25]. These differences likely stem from inherent variations in user preferences as well as the small sample sizes. Further research is needed to encompass the broad variation demonstrated in the area of prosthesis user preference.

Changes in VSPA Foot stiffness accompanied changes in ankle kinematics that can improve mobility. When the VSPA Foot stiffness was on the softer settings, the joint's range of motion increased. This allowed users to achieve varying kinematic patterns across activities. When averaged across participants, the maximum difference in peak ankle dorsiflexion between activities was $5.2°$ (stair ascent and incline walking). This difference is notable, as it corresponds to 31.2% of the mean range of motion across all participants and activities. Across level, incline, and decline walking, preferred ankle stiffness was inversely related to peak dorsiflexion angle when participants used the prosthesis at their preferred stiffness. This relationship also aligned with observations by Major et al. and Ármannsdóttir et al. [26], [27]. Interestingly, this result did not extend to stair ascent and descent. In stair activities, participants exhibited both the lowest ankle dorsiflexion and their lowest preferred ankle stiffness. This may stem from the differing biomechanical demands of stair activities compared to other ambulatory tasks. The influence of stiffness on ankle kinematics, along with the varying kinematic demands of different activities, reflect the importance of developing intelligent prostheses that can continuously adjust their mechanical behavior.

[2]For example, Participant 3's preferred stiffness was highest for decline walking and lowest for stair ascent. These two preferred stiffness values differed by 74.1% of level walking preferred stiffness (a raw difference of $415.1 \frac{Nm}{rad}$).

B. Limitations

The VSPA Foot has several unique characteristics that affect its mechanical behavior, which may have affected user preferences. It has a flat bottom with a 10-mm-thick layer of shoe sole material (*SolFlex crepe*, Shore A durometer 50-55, SoleTech, Nahant, MA, USA) for traction, a decision consistent with the literature [21], [22], [24]. It was worn without a cosmesis or shoe during testing. These decisions limit compounding compliance and ensure consistent conditions for each participant; however, most participants addressed the lack of compliance in the heel as an attribute of the device that they did not find comfortable. We also did not change the length of the prosthesis to account for variable length of individuals' intact feet. Inconsistencies in prosthesis alignment and the presence of minor backlash at the ankle joint (less than $1.5°$) could also potentially affect preferred stiffness and ankle kinematics.

Beyond the prosthesis itself, some aspects of the experimental protocol may have influenced the results. Participants consistently opted to hold onto a handrail with one hand while adjusting the dial with the other. This was permitted in the interest of participant safety and stability. It is unclear whether the use of the handrail had a meaningful impact on preference; however, it is worth noting that most participants expressed that they use handrails whenever possible in their daily lives. Additionally, participants walked at just a single, relatively mild slope for incline and decline walking activities. This was due to practical limitations on the time of each study session; the specific slope was chosen because of its applicability to community ambulation. More extreme grades, which may increase the differences in preferred stiffness when compared to level ground walking, should be investigated in future work. It is also possible that randomizing the sequence of activities would have impacted the results, but doing so was not feasible due to practical considerations related to scheduling and participant fatigue. Further, the brief time period over which each individual was observed limited user adaptation to the VSPA Foot. Several participants expressed a desire to take the device home, so that they could establish familiarity with the VSPA Foot over several days. It is unclear what effect a longer acclimation period would have on user preferences; this should be explored in future work.

V. CONCLUSION

This preliminary work indicates that preferred prosthetic ankle stiffness in unilateral, transtibial amputees varies noticeably between locomotion activities. The differences in preference reflect the distinct kinematics of different activities, as well as less quantifiable attributes of user comfort and balance. These distinctions underscore a fundamental need for variable-stiffness behavior in ankle-foot prostheses. Further, the preferences and sensitivity of individual participants varied, highlighting the need for more individualization of prosthesis behavior. In future work, we believe that biomechanical analyses may reveal underlying factors that drive user preference across different activities and people.

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