The Footropter: A Passive Prosthetic Prescription Tool With Adjustable Forefoot and Hindfoot Stiffness

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***Abstract*— Commercially available prosthetic feet are fabricated to have a fixed forefoot and hindfoot stiffness that cannot be changed in a clinical setting. This does not allow for patients to quickly compare multiple prosthetic foot stiffnesses to choose the stiffness they like the most while walking. In this paper, we present the Footropter, a passive prosthetic foot prescription tool that allows Certified Prosthetists (CPs) to rapidly change both the forefoot and hindfoot stiffnesses. The forefoot stiffness is changed by repositioning a spring clamp along a length of unbonded fiberglass layers and the hindfoot stiffness is changed by repositioning a single heel spring support. We introduce the design and working principles, characterize the ranges of available forefoot and hindfoot stiffnesses, and demonstrate the utility of the Footropter through two preference and perception studies with two unilateral transtibial prosthesis users. The Footropter, when paired with a preference optimization algorithm, can enable CPs to integrate patients' experiential input into the clinical prescription process.**

***Index Terms*— Assistive technology, prosthetic limbs, variable-stiffness, prosthesis design.**

I. Introduction

OVER 1 million people in the United States live with a lower limb amputation [1], with 150,000 lower limb amputations occurring each year [2]. Individuals with new amputations will receive a prescription for a custom lower limb prosthesis, which includes components such as a socket and a prosthetic foot. Those who already have a prosthesis receive prescriptions every few years to replace their prosthetic foot as a result of wear and tear [3].

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There are over 100 prosthetic feet available on the market, but their mechanical principles are largely similar. The most commonly prescribed prosthetic feet, making up 72% of the those prescribed, are fundamentally spring-like [4]. The heel, or *hindfoot spring*, helps to absorb shock during initial contact with the ground; this spring is often accompanied with a viscoelastic bumper which dissipates some of the absorbed energy. The keel, or *forefoot spring*, stores elastic potential energy in early stance, and returns this energy to the user in late stance [5], [6], [7], [8].

The forefoot and hindfoot stiffnesses are two of the most defining mechanical characteristics of a prosthetic foot. Matching foot stiffness to each patient's level of function, personal goals, and natural biomechanics is crucial for restoring gait. Inappropriate stiffness can lead to clear and measurable biomechanical impairments or compensations. For instance, an overly stiff hindfoot can lead to increased ground reaction force loading rate and stance phase knee flexion in early stance, while an overly stiff forefoot hinders forward progression of the tibia during midstance, leading to increased knee extension and reduced prosthetic side push off work in late stance [7], [9]. Though these biomechanical features may at times be unavoidable and may not be simultaneously achievable for given device settings, they are common for clinicians to target during prescription. Long-term, an inappropriate stiffness for a single patient may be associated with knee and back pain, and an overall decrease in quality of life [6], [7], [8], [10].

Despite the importance of prosthesis mechanics in determining patient gait and overall health, the current clinical process for matching a patient with a prosthesis is typically limited in duration and lacking a consistent methodology. Certified Prosthetists (CPs) choose prosthetic feet for their patients based on a combination of information from patient evaluations, clinical knowledge, manufacturer recommendations, foot costs and insurance reimbursements, and familiarity with particular prosthetic feet [8], [11], [12], [13], [14]. The process for selecting or comparing foot models is hindered by manufacturers, who do not publish data, such as stiffness, describing the mechanics of their prostheses. Partially due to the inability to compare models, many CPs only consistently prescribe a small handful of models that they are familiar with [15]. Once a prosthetic foot model is chosen, CPs use a manufacturer-provided chart to determine stiffness, or "category", from two input variables: a patient's weight, and a coarse notion of activity level [5], [7], [10]. For one of the most popular foot models, the Ossur Variflex, these categories have approximate spacing of 14% for the forefoot stiffness and 14% for the hindfoot stiffness, though manufacturing tolerances may make this spacing inconsistent even within the same model [9], [16]. This spacing is 2-3 times larger than the variability associated with patient preferences, though differences between biomechanical performance may not be clearly visible to an observing prosthetist [6], [15], [17]. Typically, due to barriers in quantitatively comparing feet, limited clinical time, and the burden of returning multiple feet to their manufacturers, only one prosthetic foot is trialed by each patient, and this foot will be worn for years.

In this decision-making process, there are few to no opportunities for patients to give experiential input into deciding what prosthetic foot should be chosen. Patient preference is, however, relied upon for other mechanical variables that are currently adjustable—namely, alignment, or the six degrees of freedom that describe the translation and orientation of the foot relative to the socket. We argue the incorporation of patient preference should similarly be available when deciding the stiffness of the prosthetic foot [6], [17], [18]. To find a patient's preference for prosthetic foot stiffness, the patient would compare multiple foot stiffnesses in clinic to determine which foot characteristics they like the most. Recent research has employed robotic tools to show that individuals with unilateral transtibial amputations can repeatedly and accurately state their preference for their prosthetic foot's stiffness [6], [15], [17], [19]. High repeatability does not necessarily suggest that their preferences are beneficial for them long-term, or that their preferences would not change beyond a single clinical visit. Additional patient education may be required to confer any benefits associated with patient preferences, and formal usage of preference may not work as well with patients with more limited mobility. In these cases, prosthetists may need to perform trial and error and rely more on patient-reported sensations (e.g., "I feel like I'm falling forward") or their visual observation of gait. Still, early results suggest that utilizing patient preference to guide prosthetic foot selection, in conjunction with prosthetist observations, may have distinct advantages due to being an efficient and immediately perceivable outcome metric, and a holistic determinant of optimal device behavior [18].

However, despite recent research showing the clinical potential of user preference in determining prosthetic foot stiffness, a key limitation remains: Certified Prosthetists do not have the ability to rapidly change stiffness of a prosthetic foot in a clinical setting. Additionally, because preferences rely on sequential comparisons, it is critical that patients can quickly and efficiently trial multiple stiffnesses or feet to be able to fairly compare the sensations they feel [20], [21].

Currently, CPs do not have the resources to purchase multiple prosthetic feet for their patients to trial and compare. Even if this was a possibility, patients would likely not be able to fairly compare the feet due to the time it takes to remove a prosthetic foot and attach a new one. Comparing multiple prosthetic feet may also be challenging due to each foot having different build heights and alignments. Sagittal plane ankle alignment in particular may have an interaction effect with desired stiffness, as both variables affect tibial progression and roll-over shape [22].

As a response to this problem, prosthetic foot emulators have been created to allow for quickly changing prosthetic foot stiffnesses. However, these emulators have seen limited clinical adoption. For example, the Caplex emulator system (Humotech, Pittsburgh, USA) [8], has tethers connecting a prosthesis end effector to large actuators through Bowden cables and is limited to treadmill use only. Other emulators have been created specifically for experimental use [7], [22], or with the focus of varying prosthetic foot stiffness for different walking speeds, ramps, and stairs [7], [8], [23], [24], [25]. The MyFlex uses a knob and variable linkage to modify stiffness, but changing stiffness also changes alignment, and the range of stiffness adjustment is approximately a factor of 2 (within 10° of neutral) [26]. The VSPA Foot [23] is a semi-active variable stiffness device, and has a ∼10x range of stiffness variation, but suffers from backlash and an overly heavy and rigid foot structure. Rogers-Bradley et al presented a semi-active variable stiffness prosthesis that uses parallel leaf springs with solenoid-driven locking between them [25]. It has spring-like properties that closely mimic traditional feet (rather than a rotational ankle) but has only discrete stiffness options that span ∼36%. Though the ability to rapidly change stiffness through small motor-driven systems may help CPs and patients rapidly trial multiple stiffnesses, their clinical potential is currently limited by their robustness, difficulty to repair, weight, cost, and fidelity to mimic standard energy storage and return foot behavior. Thus far, no active or semi-active emulators have seen commercial success within clinical settings.

To solve this problem, we have created a novel prosthetic foot emulator that can be easily used in a clinical setting. The Footropter is a passive prosthetic foot that allows the CPs or researcher to independently adjust the forefoot and hindfoot stiffnesses with basic hand tools, and without removing the prosthesis. This device was inspired by the *phoropter*, a tool used in the eye care industry to determine a patient's eye prescription. The phoropter facilitates A-B comparisons of vision prescriptions until the patient converges on a prescription that allows for their preferred vision. Similarly, the Footropter was created to allow a patient to make comparisons of prosthetic foot stiffnesses to converge on a preferred prosthetic foot stiffness. Here, we describe the design, working principles, and benchtop characterization of the Footropter. Finally, we demonstrate the ability to use the Footropter for 2D preference optimization in two individuals with unilateral transtibial amputations, simulating a potential clinical encounter.

II. Design

The Footropter (Fig. 1) can vary both its forefoot and hindfoot stiffness using two separate mechanisms. To serve as a useful clinical tool for selecting prosthetic foot stiffness, both the forefoot and hindfoot components must exhibit a

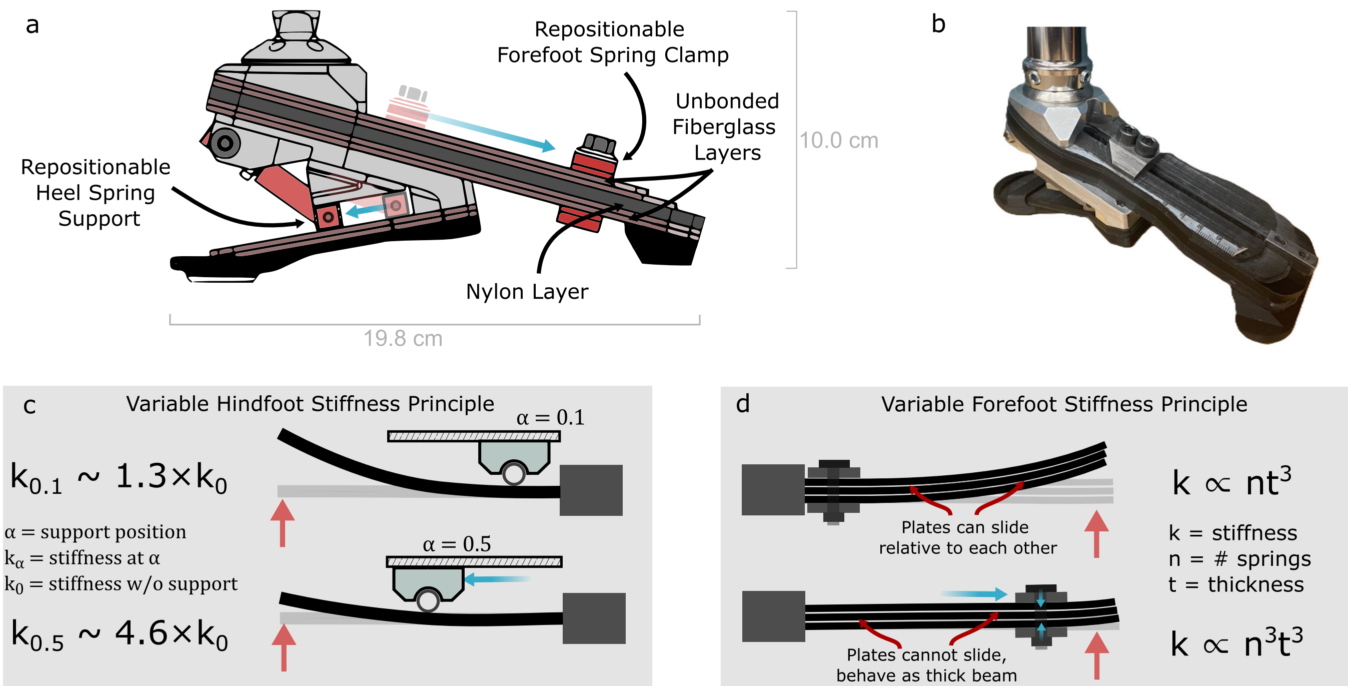

Fig. 1. (a) Footropter design. (b) Photograph of the fabricated Footropter. (c) Hindfoot stiffness is modified by changing the positioning of a simple support under a cantilever beam. The hindfoot spring is composed of two parallel unbonded plates (one plate shown for clarity) (d) Forefoot stiffness is modified by changing the location of a clamp. The plates behind the clamp are assumed to have a zero horizontal transverse shear constraint due to static fraction from the clamp.

large dynamic range of stiffnesses that encompasses the range of prosthetic forefoot and hindfoot stiffnesses observed in commercial products (20-50 N/mm for forefoot and 30-70 N/mm for hindfoot) [9].

To enable low stiffness behavior, each of the forefoot and hindfoot components is constructed from unbonded layers of fiberglass. Fiberglass is selected as the compliant material due to its high strain energy density as well as low mass density, allowing for a lightweight structure that can sustain large deflections without yielding. The unbonded layers of fiberglass allow for low stiffness behavior to be achieved without exceeding material stress limits. In a simple cantilevered beam, stress is concentrated at the outer surface of the beam, leaving the material within the interior of the beam unstressed. By stacking unbonded layers of compliant material, stresses are more evenly distributed throughout the material, enabling more compliant behavior as demonstrated in Bartlett et al. [27].

The hindfoot is configured as a propped cantilevered beam in which the beam support can be repositioned using a screw-based prismatic joint (Fig. 1). By repositioning the beam support, the stiffness of the hindfoot can be varied. An anterior positioning of the beam support results in a compliant hindfoot while a more posterior positioning of the beam support results in a stiff hindfoot. Similar mechanisms have been used in designs of other variable stiffness prosthetic devices [10], [23], [28]. Propped cantilever beams are statically indeterminate, but their stiffness can be approximated using the superposition method [29]. The stiffness can be defined according to the relative location of the adjustable simple support as:

$$K = K_0 * \frac{-4}{\alpha^3 - 6\alpha^2 + 9\alpha - 4} \tag{1}$$

where K_0 is the stiffness of the cantilever beam without the simple support, and α is the relative distance of the simple support along the beam, with 0 defined as coincident with the fixed support, and 1 defined as the end of the beam (see Appendix for derivation). The Footropter is capable of a range of α = [0.1, 0.5], which theoretically results in an approximate stiffness range of 3.5X. To fit a compliant spring in the hindfoot (while maintaining a relatively short beam length), we used two stacked fiberglass springs. These springs are in parallel with a 3D printed nylon heel, reinforced by continuous carbon fiber strands, and designed to fit in a 26cm foot shell.

The forefoot stiffness variation mechanism uses a different physical phenomenon in which the area moment of inertia of the forefoot is manipulated to modulate the forefoot stiffness (Fig. 1). Namely, a clamping mechanism enforces a zero horizontal transverse shear constraint between stacked compliant elements, changing the structure's stiffness. The unbonded layers of composite material can be clamped together at various locations. Posterior to the clamp, the layers act as a single, fused, stiff beam. Anterior to the clamp, the unbonded layers act like a much softer compliant element. By repositioning the clamp, the overall stiffness of the beam can be changed by modifying the proportion of the forefoot that is stiff vs. compliant. This clamp-based mechanism was selected to vary the forefoot stiffness and to minimize the mass and complexity of the stiffness varying mechanism located on the anterior of the prosthesis.

The stiffness of a stack of unbonded cantilevered beams is described by (2) in which n is the number of unbonded layers, E is the flexural modulus of the material, L is the cantilever length, t is the beam thickness, and w is the beam width [27], [29]. As shown in (2), the stiffness of the unbonded layered beams is proportional to the number of stacked beams, n.

$$K_{unbonded} = \frac{Ewnt^3}{4L^3} \propto n \tag{2}$$

In the bonded case, the effective thickness of the bonded cantilevered beam is equal to the number of stacked beams, n, multiplied by the thickness of each beam, t, resulting in the bonded beam stiffness shown in (3). As shown in (3), the stiffness of the bonded beam structure is proportional to n^3.

$$K_{bonded} = \frac{Ew(nt)^3}{4L^3} \propto n^3 \tag{3}$$

The range of achievable forefoot stiffnesses can then be found by taking the ratio of the fully bonded case to the fully unbonded case (4). As shown in (4) the range of achievable forefoot stiffnesses can be approximated by the square of the number of layers (n^2).

$$\Delta_K = \frac{K_{bonded}}{K_{unbonded}} = n^2 \tag{4}$$

This ideal scaling factor of n^2 is only achievable under small ranges of applied load and deflections, due to slipping between the compliant elements as well as the series stiffness of the toe portion of the prosthesis [30]. Consequently, the range of achievable stiffnesses is expected to be lower than the n^2 scaling factor. The forefoot stiffness variation mechanism is similar in some ways to laminar jamming (also called layer jamming) devices in which stacks of flexible material are forced together to increase the stiffness of the structure [30], [31], [32]. In fact, other works on laminar jamming have found the same n^2 stiffness variation factor [32]. However, unlike prior laminar jamming works in which the layered beams can either be in a high or low stiffness state, the movable clamping mechanism employed in this work allows for the stiffness of the forefoot to be continuously variable between its highest and lowest stiffness configurations.

Due to the above limitations that limit the range of achievable stiffness variation, an addition design element was introduced to extend the range of achievable forefoot stiffnesses: a nylon plate located in the middle of stack of forefoot plates. In the unbonded state, the fiberglass plates act like parallel cantilevered beams, and their location relative to one another has little effect on stiffness. In the bonded state, however, the distance of the fiberglass plates from the bonded beam neutral axis has a large effect on stiffness. Consequently, by placing a nylon beam in the middle of the composite beams, when the beam stack is clamped, the bonded beam is very stiff (the fiberglass plates are far from the neutral axis). This approach is used to increase the range of achievable forefoot stiffnesses.

The prototype device was fabricated using custom-machined aluminum and fiberglass components, as well as carbon-fiber reinforced 3D printed nylon (Mark II; Markforged, Waltham, MA, USA). Both the hindfoot and forefoot stiffness adjustment mechanisms were designed to be operated with standard Allen wrenches, which are commonly available in a clinical setting. The forefoot clamp consists of two screws, one of which is left-hand threaded such that the CP can simultaneously torque the two screws in opposite directions to tighten the clamp, canceling the net torque that would otherwise be applied to the foot and resisted by the patient. A scale is applied to the top of the foot to denote clamp location. The hindfoot stiffness is adjusted using an Allen key, and the sliding support location is visible next to a scale taped to the side of the foot.

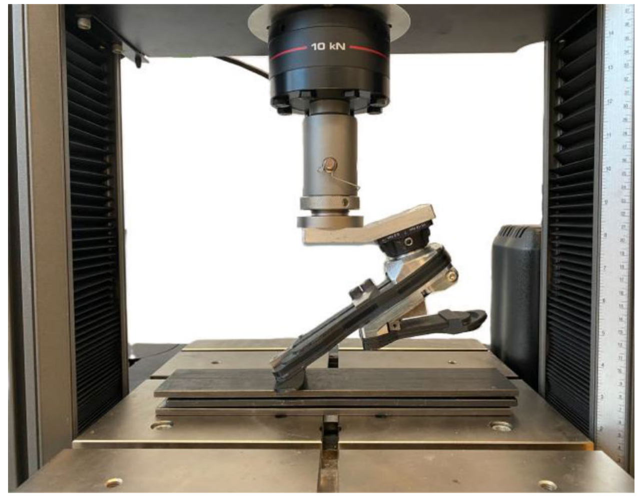

Fig. 2. Benchtop stiffness test setup. The forefoot is loaded at 20°, and the hindfoot is loaded at 15°, both up to 533 N (forefoot) and 580 N (hindfoot) at an approximate load rate of 200 N/s.

III. Benchtop Characterization

The final fabricated device was characterized using an Instron 5966 and 10 kN load cell (Fig. 2). The forefoot and hindfoot components were mounted via custom-machined adapters at 20 deg and 15 deg, respectively. The point of contact was placed on a bearing plate, such that force applied (and measured) was in line with direction of travel. The Footropter device was loaded to 533 N (forefoot) and 580 N (hindfoot), representing expected loads for our second subject, at a loading rate of approximately 200 N/s, and the forefoot and hindfoot stiffnesses were varied across their full ranges. Stiffness was defined by dividing 500 N by its associated deflection on the loading curve. Results of this benchtop characterization are shown in Figure 3 in which the forefoot and hindfoot stiffnesses are plotted against clamp location and hindfoot fulcrum location, respectively. The forefoot stiffness range is 20.4-73.2 N/mm (factor of 3.6X stiffness variation) while the hindfoot stiffness range is 34.4-66.7 N/mm (factor of 1.9X stiffness variation). Importantly, these ranges of achievable stiffnesses encompass the range of forefoot and hindfoot stiffnesses observed in commercially available prosthetic feet [9]. Hysteresis for both the forefoot and hindfoot was also quantified. Across stiffness conditions, the Footropter had a mean of 72% and 78% energy return for the forefoot and hindfoot, respectively.

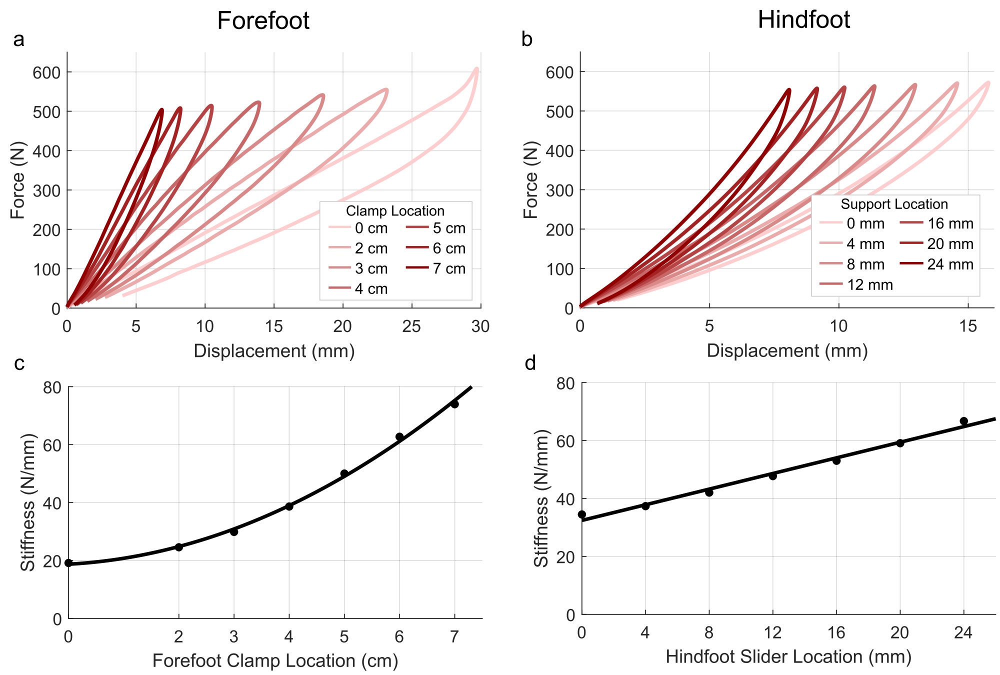

Fig. 3. Forefoot and hindfoot characterization results. (a) Forefoot force-displacement curves, (b) hindfoot force-displacement curves, (c) forefoot stiffness at specific clamp locations with second order linear fit, (d) hindfoot stiffness at specifics slider location with linear fit. All results are shown for the second version of the Footropter; version 1 was similarly characterized and had similar behavior.

As described in the subsequent section, we demonstrated the potential use of this device with two experiments and with two transtibial prosthesis users. Between these two experiments, however, minor hardware modifications were made to the device to facilitate ease of use. For the sake of completeness, the design changes between the first and second prototypes are listed here: 1) the toe portion of the prototype was shortened to better fit within a cosmetic foot shell, 2) the forefoot fiberglass beam width was decreased to counteract the stiffness change associated with the shortening of the toe, and 3) the forefoot clamp was redesigned to include two counterrotating bolts to increase clamping force (bolts were chosen to counterrotated such that tightening the bolts simultaneously did not exert a torque on the user in the transverse plane). With the first version, we found that for heavier participants, the clamp would stick-slip; that is, it would slip slightly under the first step and then remain in a slightly upward curved position for the remainder of the steps taken. This did not impact stiffness but did impact the effective dorsi/plantarflexion alignment. Both versions of the Footropter were characterized in the same way and the characterization results for the second version are shown in Fig. 3. The stiffness values (Fig 3 c, d) were calculated by dividing the force by the displacement of the loading phase curve at 500N. Additionally, testing results from the first subject are shown with the stiffness values appropriate to the iteration of the FootRopter.

Due to the clamp slipping issue under heavy loads from an older version, we also explored how clamping force affects the load at which the forefoot clamp slips. The forefoot clamp was set to the location of 6 cm and loaded to 735 N at an approximate rate of 200 N/s. We used a torque wrench to torque the forefoot clamp to 13.6 Nm, 20.4 Nm, 27.1 Nm, 34 Nm, and 40.7 Nm. Because one of the screws was reverse threaded and our torque wrench worked only in the clockwise (typical tightening) direction, we tightened it second and attempted to match the tightening torque of the first screw. We tested each clamp torque 3 times and quantified the load at which the clamp slipped via visual inspection of the force-deflection plots. The force at which the plates slipped increased linearly with clamp screw torque (Fig. 4).

TABLE I
PARTICIPANT DETAILS

Participants	Sex	Age	Weight	Daily Use Foot
S1	M	65	195 lbs	Fillauer AllPro
S2	F	69	145 lbs	Ossur Talux

IV. Experimental Methods

The Footropter was assessed in a pilot experiment with two participants (Table I): both participants completed Experiment 1 and one completed Experiment 2 (S1). Both participants were unilateral transtibial prosthesis users with a functional level of K3. Ethical approval to perform these experiments was granted by the Northeastern University Institutional Review Board (Protocol #22-03-07), and written informed consent was obtained from the participants prior to the assessment. All experiments were conducted on separate days at the Northeastern University Motion Capture Lab.

A. Experiment 1

The purpose of Experiment 1 was to assess the potential of the Footropter for use in combination with a formal

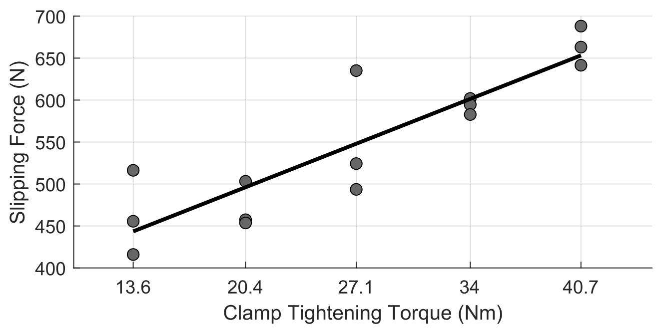

Fig. 4. Each torque tested is plotted at the force in which the forefoot clamp slipped. As the torque increased, the force at which the clamp slipped at increased.

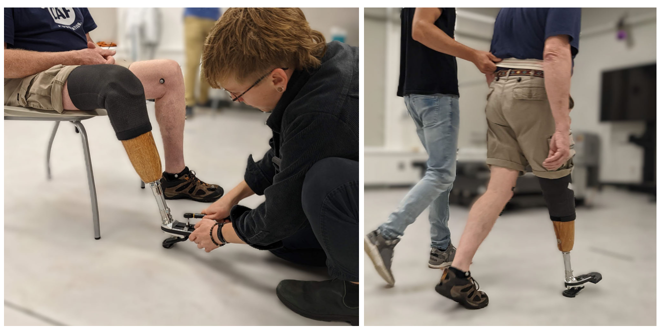

Fig. 5. Left: Certified Prosthetist adjusting the Footropter while the subject is seated. Right: Example of a trial, in which the subject walks overground with the Footropter and provides binary feedback on their preferences.

2D preference optimization. To increase accessibility to the foot and speed of the adjustments, we replaced the foot shell and shoe with 4” 50-55 durometer soling material glued to the plantar surface of the 3D printed nylon and carbon fiber foot plates. For all trials, the participant wore a gait belt and a researcher walked with the participant for safety. The participant’s daily use prosthetic foot was removed from their socket, the Footropter was attached with standard componentry, bench aligned, set to an intermediate stiffness, and then dynamically aligned. A CP on the research team made all adjustments to the prosthesis (Fig. 5). Participants familiarized themselves with the Footropter through overground walking prior to the experiment.

To validate the use of the Footropter as a tool to efficiently determine patient preferences, we performed a two-dimensional preference optimization with forefoot stiffness and hindfoot stiffness. We used Tucker et al. open-source MATLAB toolbox, POLAR (Preference Optimization and Learning Algorithms for Robotics) [33], [34]. Our prior preference optimizations with prosthesis mechanics were all one-dimensional (an overall stiffness or alignment variable), and relied on participants to understand the directionality of their preference (e.g., “I would prefer it to be stiffer”) [18], [23]. Sequential one-dimensional optimizations would fail to find a global optimum if the parameters interact (e.g., if preferred forefoot stiffness depended on the hindfoot stiffness). POLAR uses only pairwise comparisons without a directionality (e.g., “I prefer this more than the last.”) and is able to scale up to higher dimensions [33], though optimization time scales exponentially with dimensionality [35].

The Footropter was set to a random starting forefoot and hindfoot stiffness combination given by the POLAR algorithm. Simulating a common clinical task during prosthesis fittings, the participant walked 10 m to a marked location overground, turned around, and walked back to the start. The participant was occasionally reminded to consciously remember what this felt like. The Footropter’s forefoot and hindfoot stiffnesses were then changed to a new stiffness given by POLAR while the participant sat. The participant walked on these new settings and then made a forced choice of which stiffness combination they preferred. This process continued, having the participant make A-B comparisons, until the algorithm converged on the participant’s preferred forefoot and hindfoot stiffnesses.

The algorithm converged after 24 comparisons and 22 comparisons for S1 and S2, respectively. The optimization was set up with several hyperparameters manually tuned during pilot testing: notably, “length scale” (a notion of preference landscape smoothness) was set to 2, which within our tested range allowed only a few local maxima. “Preference noise” (a unitless parameter describing assumed preferences noisiness) was set to 0.03. Finally, the optimization was run in regret minimization mode, which seeks to maximize the total reward through successful comparisons (as compared to maximizing information gain, the other possible mode). This mode efficiently explores regions around the user preference.

For S2, the time required for the CP on the research team to make the Footropter adjustments was recorded by a 2nd researcher.

B. Experiment 2

The purpose of Experiment 2 was to assess how changing the Footropter forefoot stiffness and hindfoot stiffness could elicit sensations hypothesized or anecdotally assumed to exist when the forefoot or hindfoot stiffness is too stiff or soft, as well as to better understand how stiffness interacts with dorsi/plantarflexion alignment. This experiment took place at a self-selected walking speed on the treadmill (determined in conversation with the subject), with the Footropter inside an off-the-shelf foot shell, and with the participant’s shoes. The participant (S1) completed 27 walking trials, each one in a complete block design for three levels of hindfoot stiffness, forefoot stiffness, and dorsi/plantarflexion alignment. The three settings for the hindfoot were: 54 N/mm, 68 N/mm, and 85 N/mm; the three settings used for the forefoot were: 36 N/mm, 49 N/mm, and 73 N/mm; the three settings used for dorsi/plantarflexion alignment were -4.2° (dorsiflexed), +0, and +4.2° (plantarflexed) from the dynamically aligned neutral angle. The alignment changes were made by the set screws in the pyramid adaptor, and counting half-rotations, with 3 half rotations approximately equaling 4.2° [18]. The participant selected terms from a screen that most accurately described his sensations. These sensations were drawn from a combination of reported sensations in the literature [36], [37], pilot testing, and recommendations from CPs and are listed in Fig. 7. The participant walked for approximately 30 s

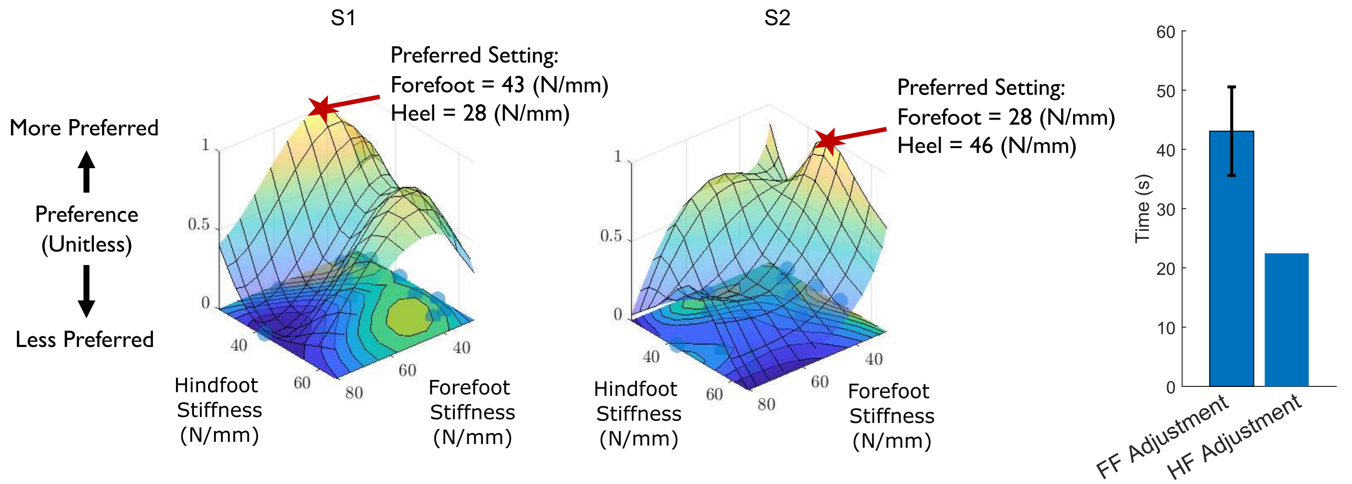

Fig. 6. Experiment 1: Preference Optimization. Left: S1's Preference landscape. Center: S2's Preference landscape. Right: Time required to adjust the forefoot and hindfoot stiffness for S2. Error bars denote standard deviation.

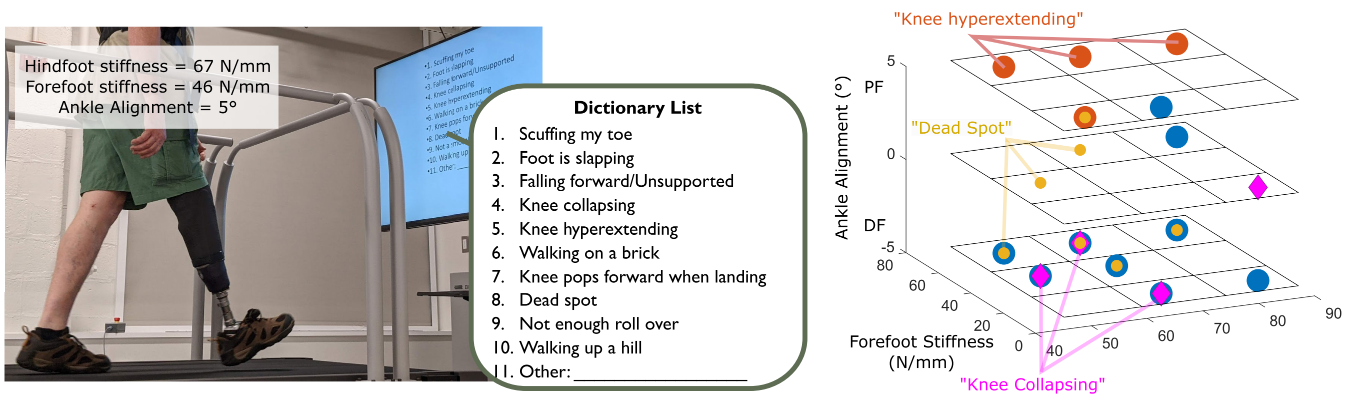

Fig. 7. Experiment 2. Left: Subject walks on the treadmill while looking at a dictionary of possible sensations. Each combination of forefoot stiffness (3 levels), hindfoot stiffness (3 levels), and alignment (3 levels) was tested once, and the subject was asked which phrase best described their sensation. Right: Location of patient sensations relative to forefoot stiffness, hindfoot stiffness, and dorsi/plantarflexion alignment.

in each foot configuration while describing their sensations. The participant was allowed to say one or more of the listed sensations, describe their own sensations, or say nothing.

V. Results

A. Experiment 1

The POLAR algorithm converged on a preferred forefoot and hindfoot stiffness for both participants (Fig. 6), as determined post-hoc by the algorithm-identified optimal stiffness not changing over the last 3 and 4 comparisons, respectively.

Figure 6 shows the preference landscapes for both participants, with a higher posterior mean being a higher preference. Because only pairwise comparisons are used, preference does not have a numerical equivalent, and the landscape can only be interpreted as illustrating relative preferences for different parameter combinations, where a combination of forefoot and hindfoot stiffnesses with a high posterior mean can be interpreted as being likely preferred over a combination with a lower posterior mean.

S1 had more consistent preferences for forefoot stiffness compared to hindfoot stiffness (Fig. 6). The preference landscape had a local maximum at the same forefoot stiffness but a higher hindfoot stiffness than the global maximum. Their overall hindfoot preference appeared to hit the lowest end of the range for the Footropter's hindfoot stiffness.

S2 was less consistent during their experiment, and this is evident in their non-smooth preference landscape (Fig. 6). They generally preferred stiffness to be as low as possible with the forefoot and preferred an intermediate hindfoot stiffness. S1 was limited in hindfoot stiffness range, preferring the lowest possible setting, and S2 was limited in forefoot stiffness range, also preferring the lowest possible setting.

During S2's experiment, the time required to adjust the forefoot was 43 ± 7 s, and 22 ± 10 s for the hindfoot, for an average of 1 min and 6 s of total time required to change both forefoot and hindfoot.

B. Experiment 2

The participant's most commonly reported sensations were "Feels like I'm falling forward," which occurred primarily with a dorsiflexed alignment, "Knee hyperextending" which occurred primarily with the plantarflexed alignment, "Knee collapsing," which occurred at a combination of low hindfoot stiffness and a dorsiflexed alignment, and "Dead spot," which tended to occur at low hindfoot stiffness (Fig. 7). In general, the large changes in dorsi/plantarflexion alignment appeared to dominate the overall experience when far from neutral.

Other sensations from the listed sensations were reported, though sparsely. During some trials, the participant explicitly mentioned they felt the hindfoot was too soft (twice) or stiff (once), or the forefoot was too soft (twice). In all five instances the identified settings were at their extreme.

VI. Discussion

The Footropter successfully accomplished a wide range of forefoot stiffness and hindfoot stiffness levels. It was relatively fast and easy for a CP to make these changes (Fig. 6). Participants were able to develop preferences for a specific hindfoot stiffness and forefoot stiffness combination (Experiment 1) and perceive sensations elicited by changes in foot stiffness and dorsi/plantarflexion alignment (Experiment 2).

We anticipated that the wide range of forefoot stiffness and hindfoot stiffness options would enable the same foot to be used for a wide range of users; however, during our preference optimization experiments, both users hit a limit to the range. S1 preferred the lowest possible hindfoot stiffness, and S2 preferred the lowest possible forefoot setting. It should be noted, however, that the range of FootRopter forefoot and hindfoot stiffness values is consistent with commercially available feet [9]. As seen in Fig. 6, S1 preferred a higher forefoot stiffness than hindfoot stiffness while S2 preferred the opposite. All of the commercial feet characterized in Ruxin et al. exhibit a higher hindfoot stiffness than forefoot stiffness [9]. Interestingly, S1 preferred relative forefoot and hindfoot stiffnesses that oppose what is present in commercial feet. Our previous research combined forefoot and hindfoot stiffness, testing these along a single combined overall-stiffness variable, and found highly consistent preferences, but it is possible that hindfoot preference is not as strong or reliable as forefoot.

Our previous research in preference optimization used quasi-passive devices, which were able to adjust either their stiffness or alignment with a small motor [6], [15], [18]. These adjustments were rapid enough to occur during swing phase of gait, meaning the participant was not required to stop during adjustments (though could stop and lift their foot for the adjustments depending on the protocol.) The Footropter required participants to sit (standing is possible but difficult), and it took a little over a minute to adjust both the forefoot and hindfoot stiffness. The higher amount of downtime between settings likely contributed to the low consistency for preference [20]; anecdotally, it did seem that participants would occasionally become distracted and forget the previous settings. Moreover, patient distraction may be a general limitation to the informal utilization of preferences in a clinical setting, and this may be particularly difficult with passive devices that require longer adjustment times.

Generally, we assume that there are no local preference maxima, and that the local maxima and saddle points evident in the individual preference landscapes are a result of participant inconsistencies in their preferences. The POLAR framework is generally capable of handling noise in the pairwise comparisons if the hyperparameters are appropriately set, and with more trials the landscapes would likely become smoother. Increased inconsistency in preferences for hindfoot stiffness (local maxima along this dimension) may indicate that hindfoot stiffness is less important to users, or that participants prioritized forefoot stiffness in comparisons where one variable was improved and the other variable was made worse. This finding may relate to the wide variation in hindfoot stiffness levels ($\sim$2X) for similar forefoot stiffness levels among different models of prosthetic feet [9].

In our second experiment, there were several consistent phrases from the subject, but not all phrases we proposed in the dictionary list were used. As expected, the subject identified perceptible negative sensations in the knee with parameter combinations that have been known to cause these issues; namely, plantarflexion and high forefoot stiffness caused 3 of the 4 “knee hyperextending” comments, and dorsiflexion (regardless of stiffness) caused 3 of the 4 “knee collapsing” comments. Similarly, dorsiflexion caused 7 of the 9 “falling forward” comments. The comment “dead spot” was most associated with a high forefoot stiffness, a low heel stiffness, and dorsiflexed alignment; this may be explained by the subject quickly gaining foot flat due to the low stiffness and dorsiflexion but then encountering a heavy resistance to rollover in midstance from the high forefoot stiffness. Though not completely consistent, all three tested variables were associated with sensations that are consistent with the biomechanics literature.

The forefoot stiffness adjustment mechanisms had limitations. Notably, the forefoot clamp required substantial tightening torque to prevent slipping (Fig. 4). This issue was largely improved with the design modification in which two screws with opposite thread directions were used to increase the clamping force, however, slipping issues still remain when high forces are applied to the forefoot. As a potential improvement in future iterations, the clamping mechanism could be replaced with shear pins that prevent the relative sliding of plates. Shear pins could be repositioned along the length of the forefoot to vary forefoot stiffness. Because of the potential to slip, we did not load to as high of a force as was done in several characterizations of commercial feet, and, due to the stiffening behavior, the stiffness of our device may actually have higher than commercial feet with similar stiffness values. Access to the Footropter stiffness adjustment mechanisms is also impaired by the foot shell; because of this, and to improve efficiency of the trials, Experiment 1 had the limitation that the Footropter was not used within a foot shell and shoe.

Future work will investigate several factors that may have contributed to inconsistent preferences in the subjects, particularly compared to our past work. In particular, questions remain about how the downtime between settings may have caused subjects to forget the sensations of walking with the previous settings, and similarly how preferences may drift over longer time scales. There may also be differences in the overall perceptibility of forefoot vs hindfoot stiffness, and future perception research could help determine whether both variables should be optimized, or a single overall stiffness parameter would be adequate. Furthermore, future work will be required to translate this or similar devices in a clinical setting. Manufacturers would need to publish the stiffness of their prosthetic foot models or perform their own testing and publish suggested feet for specific Footropter settings.

VII. Conclusion

This paper presents the design and assessment of a novel prosthetic foot emulator called the Footropter. The Footropter is a passive prosthetic foot designed specifically for clinical use and allows for adjustable forefoot and hindfoot stiffness levels spanning the range of stiffnesses observed in commercial prosthetic feet. The forefoot and hindfoot stiffness are easily adjustable by repositioning a forefoot spring clamp and a heel spring support, respectively, using Allen keys. The Footropter was assessed on two unilateral transtibial prosthesis users during two preference and perception studies. The participants were able to perceive changes in forefoot and hindfoot stiffness as well as perceive sensations elicited by changes in foot stiffness and dorsi/plantarflexion alignment. The Footropter demonstrates potential for the use of simple, passive, prosthetic foot emulators in clinical settings to integrate patients' experiential input into the clinical prescription process.

Conflicts of Interest

Authors Bartlett, Lawson, and Shepherd are inventors on a patent describing the Footropter. H. Bartlett and B. Lawson are cofounders and owners of Little Room Innovations LLC which owns the patent.

Author Contributions

Harrison L. Bartlett, Brian E. Lawson, and Max K. Shepherd developed the concept for the Footropter. Brian E. Lawson manufactured the prototypes used. Brittany M. Moores and Max K. Shepherd performed the human subject testing and analyzed the human subjects data. Brittany M. Moores and Harrison L. Bartlett led the initial writing of the manuscript, and Max K. Shepherd and Brian E. Lawson helped edit and revise.

Appendix

A. Derivation of Hindfoot Stiffness Equation

The propped cantilever used in the hindfoot is statically indeterminate to the first degree—that is, it has one extra constraint. It can be solved via superposition by breaking the problem into two simpler cases, first solving for the reaction force at the prop, and then combining the deflections due individually to the reaction force and endpoint load. [29]:

1) Consider the cantilever beam without the prop. The deflection v at point a along a beam of length L due to a load P at the endpoint is:

$$v_a^P = \frac{Pa^2(3L-a)}{6EI} \tag{A1}$$

2) Consider only a reaction load R at the prop location a, without an endpoint load. The deflection at a due to this load is given by:

$$v_a^R = \frac{Ra^3}{3EI} \tag{A2}$$

Because deflection at a must be zero due to the prop, the deflections given by (A1) and (A2) must be equal and opposite:

$$\frac{Pa^3}{3EI} = \frac{Pa^2(3L-a)}{6EI} \tag{A3}$$

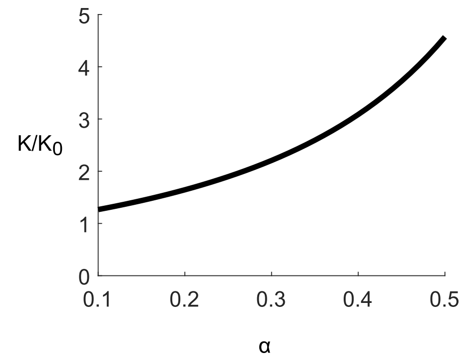

Fig. 8. Stiffness (relative to stiffness without slider) as a function of slider location.

(A3) can then be used to solve for the reaction force, R:

$$R = \frac{P(3L-a)}{2a} \tag{A4}$$

The endpoint deflections due individually to each of these two forces may then be calculated and combined. First, the end deflection due to the load is:

$$v_L^P = \frac{PL^3}{3EI} \tag{A5}$$

Second, the end deflection given by the reaction at the prop is:

$$v_L^R = \frac{Ra^2(3L-a)}{6EI} \tag{A6}$$

Substituting (A4) into (A6) and simplifying:

$$v_L^R = \frac{Pa(3L-a)(3L-a)}{12EI} \tag{A7}$$

Combining the two deflections described by (A5) and (A7) via superposition yields the following total deflection:

$$v_L = \frac{PL^3}{3EI} - \frac{Pa(3L-a)(3L-a)}{12EI} \tag{A8}$$

Factoring out the constant load P and beam characteristics EI, setting the length $L = 1$, and describing a instead as a unitless value between 0 and 1 to represent it as a fraction along the length of the spring, (A8) can be approximated as follows:

$$v_L \sim \frac{1}{3} - \frac{a(3-a)(3-a)}{12} \tag{A9}$$

Since stiffness is inversely proportional to displacement ($K(a) \sim 1/v_L$), (A9) can be used to approximate the beam stiffness:

$$K(a) \sim \frac{-12}{a^3 - 6a^2 + 9a - 4} \tag{A10}$$

Finally, beam stiffness can be described as a function of the lowest stiffness, K_0,which occurs when a = 0:

$$K(a) = K_0 \frac{-4}{a^3 - 6a^2 + 9a - 4} \tag{A11}$$

Within the range of $a = [0.1, 0.5]$, which is the approximate range of slider locations of the hindfoot slider, we get a theoretical range of approximately 3.5X (Fig. 8), though actual range is more limited, likely due to series compliance in the rest of the structure.

References

[1] K. Ziegler-Graham, E. J. MacKenzie, P. L. Ephraim, T. G. Travison, and R. Brookmeyer, "Estimating the prevalence of limb loss in the United States: 2005 to 2050," *Arch. Phys. Med. Rehabil.*, vol. 89, no. 3, pp. 422–429, Mar. 2008, doi: 10.1016/j.apmr.2007.11.005.

[2] T. R. Dillingham, L. E. Pezzin, and A. D. Shore, "Reamputation, mortality, and health care costs among persons with dysvascular lower-limb amputations," *Arch. Phys. Med. Rehabil.*, vol. 86, no. 3, pp. 480–486, Mar. 2005, doi: 10.1016/j.apmr.2004.06.072.

[3] A. C. Donaghy, S. J. Morgan, G. E. Kaufman, and D. C. Morgenroth, "Team approach to prosthetic prescription decision-making," *Current Phys. Med. Rehabil. Rep.*, vol. 8, no. 4, pp. 386–395, Dec. 2020, doi: 10.1007/s40141-020-00289-x.

[4] *Practice Analysis of Certified Practitioners in the Disciplines of Orthotics and Prosthetics*, American Board Certification Orthotics, Prosthetics, & Pedorthics, Alexandria, VA, USA, 2022.

[5] E. G. Halsne, J. M. Czerniecki, J. B. Shofer, and D. C. Morgenroth, "The effect of prosthetic foot stiffness on foot-ankle biomechanics and relative foot stiffness perception in people with transtibial amputation," *Clin. Biomechanics*, vol. 80, Dec. 2020, Art. no. 105141, doi: 10.1016/j.clinbiomech.2020.105141.

[6] M. K. Shepherd, A. F. Azocar, M. J. Major, and E. J. Rouse, "Amputee perception of prosthetic ankle stiffness during locomotion," *J. NeuroEng. Rehabil.*, vol. 15, no. 1, p. 99, Dec. 2018, doi: 10.1186/s12984-018-0432-5.

[7] 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. Movement Sci.*, vol. 54, pp. 154–171, Aug. 2017, doi: 10.1016/j.humov.2017.04.005.

[8] 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. Biomechanical Eng.*, vol. 144, no. 11, Nov. 2022, Art. no. 111009, doi: 10.1115/1.4054834.

[9] T. R. Ruxin et al., "Comparing forefoot and heel stiffnesses across commercial prosthetic feet manufactured for individuals with varying body weights and foot sizes," *Prosthetics Orthotics Int.*, vol. 46, no. 5, pp. 425–431, 2022, doi: 10.1097/pxr.0000000000000131.

[10] E. M. Glanzer and P. G. Adamczyk, "Design and validation of a semi-active variable stiffness foot prosthesis," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 26, no. 12, pp. 2351–2359, Dec. 2018, doi: 10.1109/TNSRE.2018.2877962.

[11] P. M. Stevens, J. Rheinstein, and S. R. Wurdeman, "Prosthetic foot selection for individuals with lower-limb amputation: A clinical practice guideline," *J. Prosthetics Orthotics*, vol. 30, no. 4, pp. 175–180, 2018.

[12] G. Stark, "Perspectives on how and why feet are prescribed," *J. Prosthetics Orthotics*, vol. 17, pp. 18–22, Oct. 2005, doi: 10.1097/00008526-200510001-00007.

[13] C. J. Hofstad, H. van der Linde, J. van Limbeek, and K. Postema, "Prescription of prosthetic ankle-foot mechanisms after lower limb amputation," *Cochrane Database Systematic Rev.*, vol. 2010, no. 1, Jan. 2004, doi: 10.1002/14651858.cd003978.pub2.

[14] S. U. Raschke et al., "Biomechanical characteristics, patient preference and activity level with different prosthetic feet: A randomized double blind trial with laboratory and community testing," *J. Biomechanics*, vol. 48, no. 1, pp. 146–152, Jan. 2015, doi: 10.1016/j.jbiomech.2014.10.002.

[15] M. K. Shepherd and E. J. Rouse, "Comparing preference of ankle–foot stiffness in below-knee amputees and prosthetists," *Sci. Rep.*, vol. 10, no. 1, p. 16067, Sep. 2020, doi: 10.1038/s41598-020-72131-2.

[16] A. T. Turner et al., "Prosthetic forefoot and heel stiffness across consecutive foot stiffness categories and sizes," *PLoS ONE*, vol. 17, no. 5, May 2022, Art. no. e0268136, doi: 10.1371/journal.pone.0268136.

[17] T. R. Clites, M. K. Shepherd, K. A. Ingraham, L. Wontorcik, and E. J. Rouse, "Understanding patient preference in prosthetic ankle stiffness," *J. NeuroEng. Rehabil.*, vol. 18, no. 1, p. 128, Aug. 2021, doi: 10.1186/s12984-021-00916-1.

[18] M. K. Shepherd, A. M. Simon, J. Zisk, and L. J. Hargrove, "Patient-preferred prosthetic ankle-foot alignment for ramps and level-ground walking," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 29, pp. 52–59, 2021, doi: 10.1109/TNSRE.2020.3033711.

[19] T. R. Clites, M. K. Shepherd, K. A. Ingraham, and E. J. Rouse, "Patient preference in the selection of prosthetic joint stiffness," in *Proc. 8th IEEE RAS/EMBS Int. Conf. Biomed. Robot. Biomechatronics (BioRob)*, New York City, NY, USA, Nov. 2020, pp. 1073–1079, doi: 10.1109/BioRob49111.2020.9224405.

[20] D. L. King, F. L. Jones, R. C. Pearlman, A. Tishman, and C. A. Felix, "The length of the retention interval, forgetting, and subjective similarity," *J. Experim. Psychol., Learn., Memory, Cognition*, vol. 28, no. 4, pp. 660–671, Jul. 2002, doi: 10.1037/0278-7393.28.4.660.

[21] S. Chang, C.-Y. Kim, and Y. S. Cho, "Sequential effects in preference decision: Prior preference assimilates current preference," *PLoS ONE*, vol. 12, no. 8, Aug. 2017, Art. no. e0182442, doi: 10.1371/journal.pone.0182442.

[22] A. Hansen, "Effects of alignment on the roll-over shapes of prosthetic feet," *Prosthetics Orthotics Int.*, vol. 32, no. 4, pp. 390–402, 2008, doi: 10.1080/03093640802366158.

[23] M. K. Shepherd and E. J. Rouse, "The VSPA foot: A quasi-passive ankle-foot prosthesis with continuously variable stiffness," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 25, no. 12, pp. 2375–2386, Dec. 2017, doi: 10.1109/TNSRE.2017.2750113.

[24] M. J. Major, M. Twiste, L. P. J. Kenney, and D. Howard, "The effects of prosthetic ankle stiffness on stability of gait in people with transtibial amputation," *J. Rehabil. Res. Develop.*, vol. 53, no. 6, pp. 839–852, 2016, doi: 10.1682/jrrd.2015.08.0148.

[25] E. Rogers-Bradley, S. H. Yeon, C. Landis, and H. M. Herr, "Design and evaluation of a quasi-passive variable stiffness prosthesis for walking speed adaptation in people with transtibial amputation," *IEEE/ASME Trans. Mechatronics*, vol. 29, no. 1, pp. 335–346, Feb. 2024, doi: 10.1109/TMECH.2023.3276710.

[26] J. Tabucol et al., "The MyFlex-ξ foot: A variable stiffness ESR ankle-foot prosthesis," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 33, pp. 653–663, 2025, doi: 10.1109/TNSRE.2025.3534096.

[27] H. L. Bartlett, S. T. King, M. Goldfarb, and B. E. Lawson, "Model based design of a low cost and compliant low profile prosthetic foot," *J. Biomechanical Eng.*, vol. 144, no. 3, Mar. 2022, Art. no. 031009, doi: 10.1115/1.4052369.

[28] C. Lecomte, A. L. Ármannsdóttir, F. Starker, H. Tryggvason, K. Briem, and S. Brynjolfsson, "Variable stiffness foot design and validation," *J. Biomechanics*, vol. 122, Jun. 2021, Art. no. 110440, doi: 10.1016/j.jbiomech.2021.110440.

[29] R. C. Hibbeler, *Mechanics of Materials*, 9th ed., Upper Saddle River, NJ, USA: Prentice-Hall, 2014.

[30] Y. S. Narang, J. J. Vlassak, and R. D. Howe, "Mechanically versatile soft machines through laminar jamming," *Adv. Funct. Mater.*, vol. 28, no. 17, Apr. 2018, Art. no. 1707136, doi: 10.1002/adfm.201707136.

[31] F. Caro and M. G. Carmichael, "A review of mechanisms to vary the stiffness of laminar jamming structures and their applications in robotics," *Actuators*, vol. 13, no. 2, p. 64, Feb. 2024, doi: 10.3390/act13020064.

[32] Y. S. Narang, B. Aktaş, S. Ornellas, J. J. Vlassak, and R. D. Howe, "Lightweight highly tunable jamming-based composites," *Soft Robot.*, vol. 7, no. 6, pp. 724–735, Dec. 2020, doi: 10.1089/soro.2019.0053.

[33] M. Tucker, K. Li, Y. Yue, and A. D. Ames, "POLAR: Preference optimization and learning algorithms for robotics," 2022, *arXiv:2208.04404*.

[34] M. Tucker et al., "Preference-based learning for exoskeleton gait optimization," 2019, *arXiv:1909.12316*.

[35] K. A. Ingraham, M. Tucker, A. D. Ames, E. J. Rouse, and M. K. Shepherd, "Leveraging user preference in the design and evaluation of lower-limb exoskeletons and prostheses," *Current Opinion Biomed. Eng.*, vol. 28, Dec. 2023, Art. no. 100487, doi: 10.1016/j.cobme.2023.100487.

[36] J. H. Bowker, "Critical choices: The art of prosthesis prescription," in *Atlas of Limb Prosthetics*, J. Bowker and J. W. Michael, Eds., 2nd ed., St. Louis, MO, USA: American Academy of Orthopedic Surgeons, 1992, pp. 717–720. [Online]. Available: http://www.oandplibrary.org/alp/chap29-01.asp

[37] S. Kapp and D. Cummings, "Transtibial amputation: Prosthetic management," in *Atlas of Limb Prosthetics*, 2nd ed., St. Louis, MO, USA: American Academy of Orthopedic Surgeons, 2002.