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Informing Ankle-Foot Prosthesis Prescription through Haptic Emulation of Candidate Devices

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Abstract— **Robotic prostheses can improve walking performance for amputees, but prescription of these devices has been hindered by their high cost and uncertainty about the degree to which individuals will benefit. The typical prescription process cannot well predict how an individual will respond to a device they have never used because it bases decisions on subjective assessment of an individual's current activity level. We propose a new approach in which individuals 'test drive' candidate devices using a prosthesis emulator while their walking performance is quantitatively assessed and results are distilled to inform prescription. In this system, prosthesis behavior is controlled by software rather than mechanical implementation, so users can quickly experience a broad range of devices. To test the viability of the approach, we developed a prototype emulator and assessment protocol, leveraging hardware and methods we previously developed for basic science experiments. We demonstrated emulations across the spectrum of commercially available prostheses, including traditional (e.g. SACH), dynamic-elastic (e.g. FlexFoot), and powered robotic (e.g. BiOM® T2) prostheses. Emulations exhibited low error with respect to reference data and provided subjectively convincing representations of each device. We demonstrated an assessment protocol that differentiated device classes for each individual based on quantitative performance metrics, providing feedback that could be used to make objective, personalized device prescriptions.**

I. Introduction

A. Typical Prescription Process

The prescription of ankle-foot prostheses is hindered by uncertainty about which device is most suitable for a given individual [1]. Payers expect justification for prosthesis selection, but without objective data clinicians can only provide their subjective impression, the expressed needs of the individual, and, at best, basic assessment of an individual's pre-prescription mobility [2]. Recent robotic devices have intensified this problem, as they have demonstrated benefits to the user [3, 4], but at a high price (about $80,000 for a BiOM® T2 vs. about $1,000 for a conventional prosthesis). The degree to which individual users will benefit also remains unclear. Given this uncertainty, clinical practice is slow to accommodate disruptive technologies, and is not able to effectively predict a user's activity-level and ability with a device they have never used.

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B. Informing Prescription by Haptic Emulation

We propose a new approach, wherein patients 'test drive' candidate devices, providing hard data on how they perform with each prosthesis. This could be done by buying and trying many different prostheses for each individual, but the process would be laborious and would require expensive inventories of different models of prosthesis (each with variations for different body weights, activity levels, and foot sizes). Instead, clinicians could fit patients with a prosthesis emulator and provide the experience of wearing these different prostheses by simply switching modes in a software interface. Most commercially-available devices can be classified into one of three groups: traditional stiff and dissaptive solid ankle cushioned heel (SACH) prostheses, conventional spring-like dynamic elastic response (DER) prostheses, and actively-controlled robotic prostheses. Emulating these diverse behaviors with a single prosthesis requires versatility beyond the capabilities of currently-available mobile robotic prostheses, which are fine-tuned to exhibit specific behaviors in a convenient autonomous package. To maximize versatility in basic science experiments that do not require autonomy, e.g. [5], we previously developed a robotic prosthesis system in which a powerful off-board motor and controller actuate a lightweight prosthesis end-effector through a flexible Bowden cable transmission [6]. In the present study we test whether such a system can convincingly emulate the behavior of existing off-the-shelf prostheses.

C. Metrics for Evaluating Benefit

To evaluate the benefits each emulation mode provides to an individual, it would be useful to have outcome metrics that capture aspects of performance that are relevant to daily life. The most-cited measure for the efficacy of an assistive device is metabolic rate (the rate at which biochemical energy is used by the body to perform a task). However, in clinical practice, the expensive equipment required to measure metabolic rate is typically not available. Also, energy consumption must be balanced against other factors such as comfort, stability, versatility, and maximal performance. Therefore, it would be useful to have a set of outcomes that can be measured simply and quickly in a clinical setting, and can estimate energy consumption as well as other important outcomes. Heart rate scales roughly with metabolic rate [7] and could be used as a surrogate that is simpler to measure and responds more quickly to the task. Maximum sustainable walking speed (MSWS) also scales with metabolic rate [8], and might include information about perceived stability and comfort. Finally, patient-reported satisfaction scores and comments can include information about perceived effort and stability, comfort, and gait aesthetics.

D. Summary and Hypotheses

The aim of this study was to test the feasibility of a new approach to the prescription of ankle-foot prostheses that includes quantitative measurements of how an individual will perform with a set of candidate devices. We hypothesize that (1) a tethered robotic prosthesis can accurately emulate different classes of commercially-available prostheses and that (2) simple, clinically-relevant performance metrics can provide quantitative data on an individual's performance that differentiate device classes and individuals.

II. Methods

A. Overview of the Ankle-Foot Prosthesis Emulator

We developed a prototype haptic emulator capable of exhibiting the behavior of a wide range of commercially available ankle-foot prostheses. The prosthesis emulator consists of a powerful off-board motor and real-time controller, a flexible tether transmitting sensor signals and mechanical power, and an ankle-foot prosthesis end-effector (Fig. 1, [6]). The user wears the prosthesis as they would a conventional prosthesis, except that they are constrained by the tether to walk on a treadmill.

Device behavior was controlled by matching the ankle torque vs. angle relationships of commercially available prostheses. We also programmed a behavior that is unlike any commercially available device, to demonstrate the system's ability to emulate candidate designs for testing prior to physical implementation. Emulated behavior was switched by buttons in a simple software interface, without mechanically modifying the emulator hardware. Walking performance was measured for each mode using a variety of techniques that could be used to inform device prescription.

B. Experimental Methods

We recruited six subjects with unilateral transtibial amputation to test the efficacy of the prosthesis emulator. Subject parameters are listed in Table I. Subjects wore the prosthesis emulator as they would a standard ankle-foot prosthesis: a pylon, with universal prosthesis adapters at each end, was sized according to each subject's leg length and used to attach the prosthesis emulator to each subject's prescribed socket. Subjects were fitted with the prosthesis emulator by a Certified Prosthetist, who set the alignment of the device, which was then retained throughout the study. Subjects had previous experience with the prosthesis emulator hardware (but not the controller used here) totaling at least four hours of walking. Subjects completed the protocol twice, with data reported for the second repetition. The experimental protocol consisted of two days of walking: one day walking on a level treadmill and the other on an inclined (5°) treadmill. Treadmill speed was set to 1.25 $\text{m}\cdot\text{s}^{-1}$ or each subject's preferred walking speed (measured overground in a 50 m hallway) if it was less than 1.25 $\text{m}\cdot\text{s}^{-1}$. Subjects walked with their prescribed prosthesis (PRES) and with the prosthesis emulator in four modes (see supplementary video): SACH (emulating a Solid Ankle Cushioned Heel foot), DER (emulating a Dynamic Elastic Response foot), BIOM (emulating the BiOM® T2), and HIPOW (a custom mode with high power output). Conditions were presented in random order, and subjects were required to rest for five minutes between conditions.

TABLE I

HUMAN SUBJECT PARAMETERS

#	K-Level	Cause	TSA [yrs]	Age [yrs]	BW [lbs]	Prescribed device
1	K3	Traumatic	9	42	176	Fillauer Wave
2	K3	Traumatic	6	57	183	Ottobock Triton V. S.
3	K3	Traumatic	1	45	180	Össur Vari-Flex
4	K3	Congenital	46	49	165	F. I. Renegade A·T
5	K3	Traumatic	12	48	210	BiOM® T2
6	K3	DVT	18	53	189	Össur Vari-Flex T. S.

We evaluated users' walking performance in each emulator mode using four different metrics: two objective measures of steady-state walking efficiency and two subjective measures indicating user satisfaction and maximal performance. Metabolic energy consumption was estimated using indirect calorimetry [9], performed using gas concentrations and flow rates measured by a commercial respirometry system (Oxycon™ Mobile), averaged over the last three minutes of each trial. Heart rate was measured by the same respirometery system using pulse oximetry, and averaged over the last three minutes of each trial. Net metabolic energy consumption and net heart rate were computed as the average measurement in each condition, minus the average measurement during a quiet standing trial. Percent change in net metabolic energy consumption and percent change in net heart rate were computed relative to the level ground SACH condition, to quantify the marginal benefits of other conditions. Satisfaction was assessed by asking the subjects to rate each of the emulated modes on a Likert Scale [10] which ranged from from -10 to 10, where -10 indicated "walking is impossible", 0 indicated "similar to walking with my prescribed prosthesis", and $+10$ indicated "walking is effortless". Maximum sustainable walking speed was established at the end of each walking trial by progressively increasing the speed of the treadmill in 0.05 $\text{m}\cdot\text{s}^{-1}$ increments every ten seconds until the subject indicated they felt they could no longer sustain walking at the set speed for five more minutes. Measures of ankle torque and angle were calculated using on-board encoders (torque was inferred by measuring the deflection of a series elastic spring).

C. Ankle Joint Torque vs. Angle Control

Prosthetic ankle torque (τ_a) was controlled as a function of ankle angle (θ), with different relationships for the dorsiflexion ($\dot{\theta} < 0$) and plantarflexion ($\dot{\theta} > 0$) phases of stance [6]. Desired ankle torque ($\tau_{a,des}$) was a piecewise linear fit to representative literature data obtained from inverse dynamics measurements made during walking (Fig. 2, data from [11] for SACH, [12] for DER and BIOM). To switch the emulator from one mode to another, the experimenter selected a different ankle torque vs. angle reference.

The motor was controlled as a velocity source (low-level control embedded in the motor driver performed velocity control), which was driven according to simple proportional control on torque error.

$$\dot{\theta}_{motor} = k_p * \tau_{a,err} \quad \tau_{a,err} = \tau_{a,des}(\theta) - \tau_{a,mes} \tag{1}$$

We tuned k_p to best suit the stiffness of each mode's ankle torque vs. angle relationship: when stiffness was high (e.g., SACH or the plantarflexion phase of HIPOW) larger k_p resulted in better tracking; but when stiffness was low (e.g., DER or the dorsiflexion phase of HIPOW) smaller k_p resulted in more stable torque tracking.

Rapidly decreasing torque during the plantarflexion phase of the BiOM® T2 emulation proved challenging for this simple proportional control scheme, so desired ankle torque was adjusted with an iteratively learned torque ($\tau_{a,lrn}$) to compensate for steady-state errors (inspired by [13]).

$$\dot{\theta}_{motor} = k_p * (\tau_{a,des}(\theta) + \tau_{a,lrn}(\theta) - \tau_{a,mes}) \tag{2}$$

The learned torque during step n was a function of torque errors ($\tau_{a,err}$) on previous steps.

$$\tau_{a,lrn}(\theta, n+1) = \tau_{a,lrn}(\theta, n) + k_l * \tau_{a,err} \tag{3}$$

We tuned k_l to minimize steady-state tracking errors quickly but without overshoot, approximately thirty walking strides.

Because ankle torque is minimal during swing, the proportional controller was switched to control ankle angle when $\tau_{a,mes} < 15$ N·m at the end of stance, driving the joint to the initial dorsiflexion angle (θ_{des}) of the reference data.

$$\dot{\theta}_{motor,swing} = k_s * (\theta_{des} - \theta) \tag{4}$$

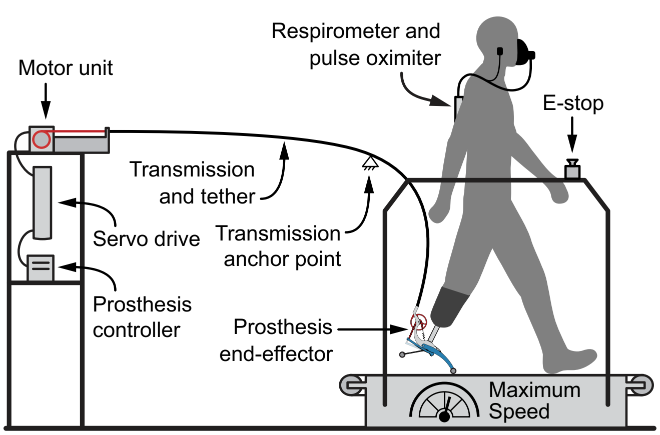

Fig. 1. The ankle-foot prosthesis emulator consists of a lightweight prosthesis worn by the user and actuated through a flexible tether by a powerful motor and control system. By placing actuation and control off-board, the system can emulate an exceptional variety of behaviors at a worn mass comparable to passive mobile prostheses. Adjustments to the device behavior experienced by the user are made in the prosthesis control software rather than by modifying the end-effector. Metabolic rate, heart rate, maximum walking speed, and user preference are measured to assess which behaviors best suit the user.

III. Results

A. Torque vs. Angle Control

Mean desired and measured prosthetic ankle torque trajectories during the stance phase of the prosthetic limb for a representative subject during level ground walking are presented in Fig. 3. Root mean squared (RMS) error is presented to quantify torque tracking errors. Mean RMS errors across all subjects was 7.8±2.4 N·m, 2.6±0.7 N·m, 3.4±0.9 N·m, and 7.9±1.1 N·m for SACH, DER, BIOM, and HIPOW modes, respectively. Mean measured prosthetic ankle torque vs. angle in each emulation mode is presented for a representative subject in Fig. 3, along with the reference data used to design the emulation for comparison.

B. Walking Performance Outcome Metrics

Measurements of walking performance are listed for each subject in Fig. 4. Subject #1, a DER user, used the least metabolic energy and had the lowest heart rate in passive modes (DER and SACH) on level ground, although on inclined ground metabolic energy was minimized in HIPOW. However, this subject always preferred and walked fastest with the robotic modes (BIOM and HIPOW). Heart rate data were inconsistent with these observations, with passive modes (DER and SACH) always exhibiting the lowest heart rate. Subject #2, a DER user, used the least metabolic energy and had the lowest heart rate in the robotic modes, but always preferred DER. This subject walked fastest in BIOM on level ground but walked fastest in DER when walking uphill. Subject #3, a DER user, used the least metabolic energy and had the lowest heart rate in BIOM, but walked fastest in HIPOW. This subject preferred the passive modes on level walking, but preferred HIPOW on inclined ground. For subject #4, a BiOM® T2 user, DER was optimal by all metrics on level ground. This subject used less metabolic energy in BIOM on inclined ground, but still preferred DER. Subject #5, a DER user, used the least metabolic energy and walked fastest in BIOM on level ground but used the least

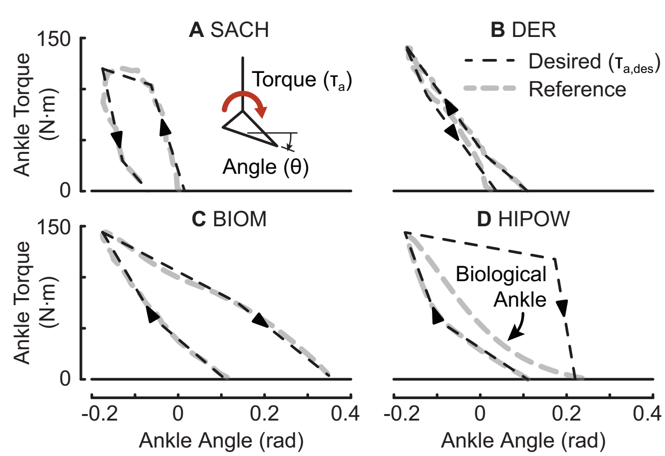

Fig. 2. Emulation was performed by matching the ankle torque vs. angle relationships of commercially-available prostheses. Ankle torque was controlled as a function of ankle angle, with different relationships for the dorsiflexion and plantarflexion phases of stance. The desired torque (dark dashed) was a piecewise linear fit to literature reference data (light dashed).

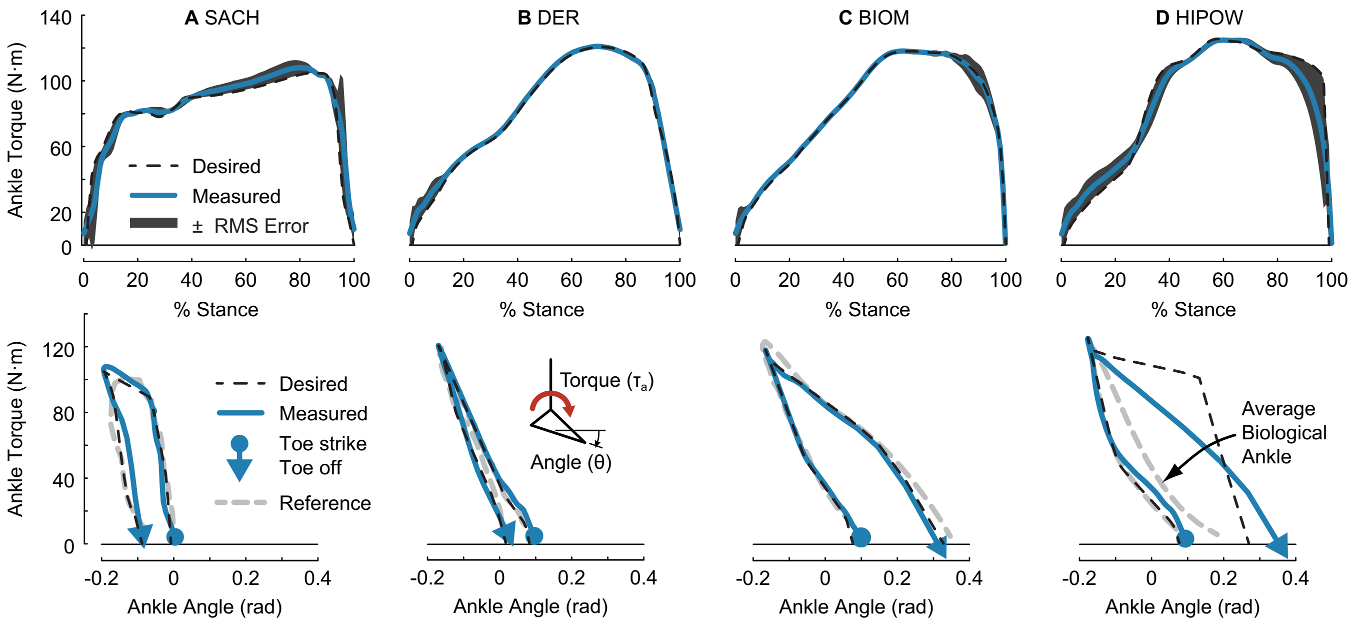

Fig. 3. Emulating ankle torque vs. angle behavior of candidate prostheses. Demonstrated emulations include: **A** solid-ankle cushioned heel (SACH), **B** dynamic-elastic response (DER), **C** an active robotic foot, the BiOM® T2 (BIOM), and **D** a conceptual high-powered robotic foot design (HIPOW) that was designed to maximize torque during plantarflexion, with the expectation that torque would not be tracked precisely. Data in A-D are from a single individual with unilateral transtibial amputation walking at 1.25 m·s^{-1} on level ground over approximately 150 strides. *Top:* Prosthetic ankle torque plotted vs. % stance of the prosthesis-side step. *Bottom:* Prosthetic ankle torque plotted vs. prosthetic ankle angle.

energy in HIPOW on inclined ground. This subject always preferred to walk in BIOM. Heart rate, inclined SACH, and inclined MSWS data were not available due to equipment failure and scheduling difficulties. Subject #6, a DER user, used the least metabolic energy in HIPOW, although heart rate was minimized and walking speed maximized in BIOM. This subject preferred BIOM on inclined ground but preferred DER on level ground.

IV. Discussion

A. Quality of Prosthesis Emulation

We demonstrated a haptic emulator that exhibited high-quality tracking of the ankle torque vs. angle relationships of an array of commercially-available prostheses. The emulator tracked the desired torque vs. angle relationships with average RMS error between 2 and 4% of the maximum ankle torque, depending on the mode (Fig. 3). The largest tracking errors were exhibited early in stance when torque was below 30 N·m and just after the transition from dorsiflexion to plantarflexion. Because of torque sensor noise and nonlinearities, motor position was held constant below a 30 N·m torque threshold, leading to reduced emulation quality in this region. In future versions we will improve sensor linearity and signal-to-noise ratio by, e.g., implementing a digital ankle encoder and reducing backlash in the series elastic actuator, or through the implementation of strain gauge sensing. The state-based torque vs. angle controller requires some plantarflexion velocity to be certain of the state transition from dorsiflexion to plantarflexion. Variability in the timing of this transition led to reduced emulation quality near the transition. In future versions we will eliminate this state distinction, instead emulating the ankle torque as a function of ankle velocity in addition to ankle angle. Iterative learning control improved torque tracking quality in BIOM but also introduced dynamics that are likely not exhibited by the BiOM® T2. Subjectively, we observed increased step-to-step variability and slow changes in device behavior as it adapted to the user's own slow changes. In future versions of the emulator system we will improve feedback control performance, through improved sensing and actuation as well as by implementing a derivative term in the feedback and improving the hardware to mitigate the deleterious effects of transmission friction and compliance.

We demonstrated successful emulation of different classes of device behavior, but it remains to be seen if the system demonstrated here can successfully differentiate subtle variations within device classes. The current emulator prototype can be programmed to exhibit such subtleties, but a controlled test of quality has yet to be performed. Most unilateral transtibial amputees are prescribed DER feet, so it would be useful if the emulator could differentiate brands, models, and configurations of prostheses, including variations in stiffness, damping, geometry, and weight. For robotic feet with programmable behavior, such as the BiOM® T2, device behavior should be optimized to ensure that prescription decisions are made using the best possible configuration of the emulated device for a given user. To this end, we are currently developing methods for automatic configuration of device behavior to maximize user benefit.

Subjects generally reported that the behavior of the emulator was similar to the devices that were being emulated, with some subtle differences that we will address in future versions. Two subjects had experience walking with a SACH foot. One reported: "[SACH mode was] stiff as a board! Felt

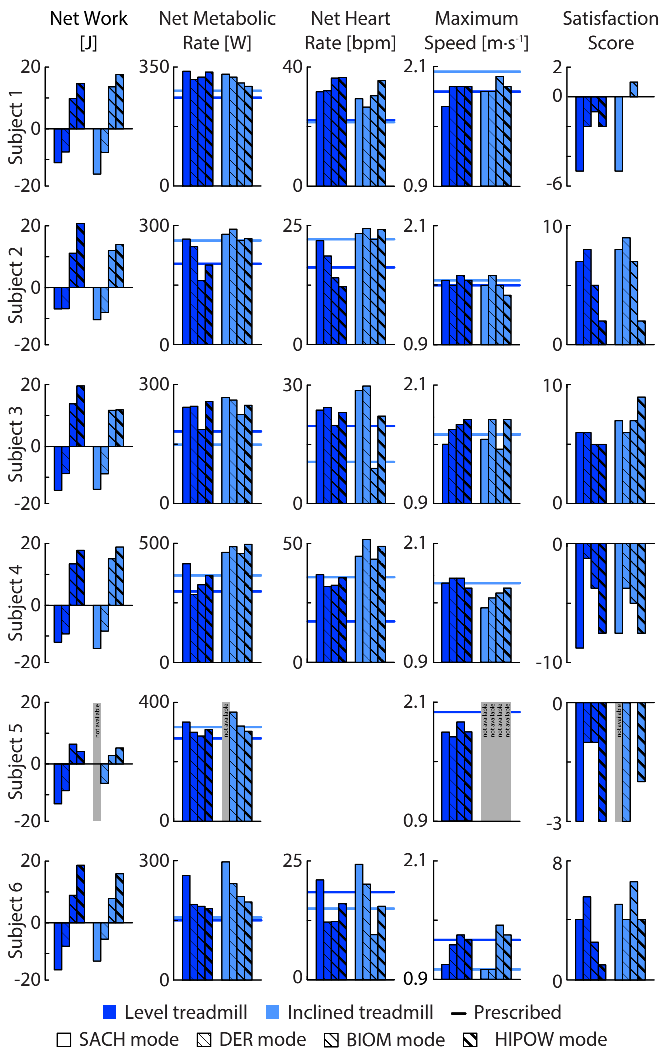

Fig. 4. Walking performance outcome metrics listed for each subject across different emulator modes and two treadmill incline conditions. Hatching indicates different modes, colors indicate different inclines, and horizontal lines indicate reference measurements taken during a condition where the user walked with their prescribed prosthesis.

just like my old leg and made it hard to walk fast." All subjects had extensive experience walking with DER feet, and all DER users reported that DER mode felt similar to their prescribed device. One subject reported: "This [DER mode emulation] is really good, I'll say my prescribed device is more comfortable, but just barely." The BiOM® T2 user reported that the DER mode felt most similar to his prescribed device, possibly because of the device's ability to be reconfigured to suit an individual's needs. This impression suggests that a fixed reference for BIOM emulation may be too simplistic, but also that this user may find a satisfactory balance of cost and performance with a DER prosthesis.

User feedback on the HIPOW mode demonstrated the emulator's utility as a tool for testing design ideas prior to physical implementation. All users found the HIPOW mode to be much too powerful during steady-state walking on a level treadmill, but some commented that the additional power was useful during uphill and/or maximum speed walking. For example, one said "The high push-off is hard to control. The region of good places to put my foot is much smaller. If I put my foot in the wrong place I get a lot of push-off in the wrong direction." But, another comment identified benefits during inclined walking: "Push-off with [HIPOW mode] was way too much on the flat treadmill but just now [on the 5° slope] it felt helpful."

B. Limitations of the Scope of Emulation

Several aspects of prosthesis behavior were not considered in our emulation scheme, which could affect outcomes. We represented different devices by their stance-phase sagittal-plane ankle torque vs. angle relationship as measured in previously published amputee walking experiments. This common model of ankle behavior [14, 15] is limited as it contains only one degree of freedom, ankle plantar/dorsiflexion, and does not consider the swing phase of gait. This model cannot fully predict the six independent components of force and moments that act on the user's residual limb.

We observed three main limitations of considering just saggital ankle angle in our emulation. First, to emulate the effect of varying foot length independent of joint impedance would require an additional degree of freedom to control the reaction forces independent of the reaction moment. Second, as one user reported, "Because [the emulator] is so stiff, I notice whenever I take a slightly off step. My prescribed foot is compliant in every direction so there's more room for error." In future versions of the emulator system we will characterize the complete force/torque-deflection characteristics of the different commercially available prostheses through amputee-independent benchtop tests [16, 17] and through controlled walking trials. It is likely that including passive compliance in the structure of the prosthesis, comparable to what is provided by a DER prosthesis, will improve emulation quality significantly. We are also exploring prosthesis designs with additional controlled degrees of freedom [18] to capture differences across device type. Third, robotic feet with programmable behavior, e.g. the BiOM® T2, need not exhibit the same ankle torque vs. angle behavior from one step to the next; their behavior can be a function of inputs other than ankle angle [19]. To better emulate the behavior of such systems we are developing emulations of device-specific high-level control schemes.

Ankle torque is typically not considered significant during swing, but adding mass to the foot increases metabolic energy consumption by about 9% per added kilogram [20], which suggests inertial and gravitational forces during swing are significant. Given that powered ankle-foot prostheses require extra mass for motors, batteries, and electronics, they tend to be about 1 kg heavier than passive prostheses, which would reduce our expectation for the energetic benefits of powered assistance strategies. Future versions of the prosthesis end-effector will be about 30% lighter, matching the mass of the lightest passive prostheses, and modular weights will be added to emulate candidate device mass.

Dorsiflexion torque at the beginning of stance during heel-only ground contact was provided by a passive heel spring, rather than through active control, in order to simplify the design of the emulator. Peak torque and energy absorption/dissipation are relatively small during this relatively short period of stance, so we believe this behavior to be less important than the primary stance phase behavior. However, future versions of the emulator will include active control of dorsiflexion torque to more completely characterize differences across device types.

C. Utility of Performance Metrics

We demonstrated a protocol for measuring users' walking performance across emulator modes that discerned individuals' needs using simple quantitative measures. All unilateral transtibial amputees we tested appeared to benefit from robotic assistance strategies to some degree but with individual subject differences. The five DER users we tested appeared to have the potential for improved walking performance and satisfaction with a robotic prosthesis, but were never able to explore this option within the conventional prescription process. The BiOM® T2 user showed benefits from robotic assistance, but only when walking uphill, and always preferred walking in the passive modes. Despite having the good fortune of using the most sophisticated technology available, it is possible that the conventional prescription process falsely identified this individual as one who would benefit most from a robotic device. By exploring candidate device behaviors through haptic emulation, prosthesis prescriptions could be objectively justified and ensure that users reach an appropriate balance of cost and benefit.

Users' comments suggest that a variety of factors contribute to overall level of satisfaction with the various modes. In future protocols we will expand the subjective satisfaction assessment to address contributing factors such as perceived effort, stability, and pain.

While our subjects varied greatly in time since amputation and make and model of prescribed device, they were relatively homogeneous in K-Level, cause of amputation, and weight. We expect that users with lower K-Level, dysvascular amputation, and/or significantly higher or lower body weight could have different needs from the subjects tested here. We are currently recruiting individuals with more diverse medical histories and developing hardware to support a broader group of individuals for future tests of the emulator.

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