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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 adoption of these devices has been limited by their relatively high cost and uncertainty about the degree to which individual users will benefit. Prostheses are typically prescribed based on an individual's level of activity and the professional opinion of doctors and prosthetists, assessed by visual observation and patient feedback. We propose a new approach, in which individuals 'test-drive' new technology using a haptic prosthesis emulator while their walking performance is quantitatively assessed and results are distilled to inform device prescription. In this emulator system, prosthesis behavior is controlled by computer software rather than mechanical implementation, so users can experience a broad range of devices in a short period of time. We developed a prototype ankle-foot prosthesis emulator system and assessment protocol to test the viability of such an approach. We demonstrate successful emulations across the spectrum of commercially available prostheses, including traditional (e.g. SACH), dynamic-elastic (e.g. FlexFoot), and robotic (e.g. BiOM® T2 System). Emulations exhibited low error with respect to reference data and provided a subjectively convincing representation of each device. We demonstrate an assessment protocol that differentiates these device classes based on individuals' walking performance, providing informative objective feedback for device prescription.**

I. Introduction

A. Conventional Prescription Process

The prescription of ankle-foot prostheses has been plagued by uncertainty about which device is most suitable for a given individual [1]. Practitioners must balance individuals' needs, subjective measures of activity-level and ability, and insurers' justification requirements, given very little common knowledge to inform their decision. Recent robotic devices have intensified this problem, as they can provide great benefits to the user [2], but at a very high price (about $80,000 for a BiOM® T2 System vs. about $1,000 for a conventional prosthesis). The conventional prescription process is slow to adapt to disruptive technologies and has no means to predict a user's activity-level and ability with a type of device they have never used.

B. Device 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 re-fitting would be time-consuming and would require expensive inventories of different models of prosthesis (each with variations for different body weights, activity levels, and shoe sizes). Instead, using a prosthesis emulator, practitioners could provide patients with a haptic emulation of the available prostheses, making only simple software adjustments to switch devices. Most 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. These devices vary greatly in their behavior, so successful emulation would require an exceptionally versatile robotic prosthesis. Versatility can be maximized by placing mechanical power and control off-board [3], enabling considerably higher control performance than with on-board actuation [4].

C. Metrics for Evaluating Benefit

To evaluate the benefits of each emulation mode to an individual, it would be useful to have outcome metrics that capture aspects of performance that are relevant to utility in daily life. The most-cited measure for the efficacy of an assistive device is metabolic energy consumption rate (the rate at which energy is used by the user to perform a task). However, in clinical practice, the expensive equipment required to measure metabolic rate is typically not available and energy consumption is 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 the clinic, and can estimate energy consumption as well as several other outcomes. Heart rate scales roughly with metabolic rate [5] and could be used as a surrogate for metabolic rate that is simpler to measure and responds more quickly to the task. Maximum sustainable walking speed (MSWS) also scales with metabolic rate [6], and might include information about comfort and perceived stability. Finally, patient-reported satisfaction scores and comments can capture intangible factors like comfort.

D. Summary and Hypotheses

The aim of this study was to test the feasibility of two components of a new approach to ankle-foot prosthesis prescription. We hypothesize that (A) a tethered robotic prosthesis can provide accurate haptic emulation of different classes of commercially available prostheses and that (B)

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simple, clinically-relevant performance metrics can provide quantitative data on an individual's performance that differentiate device classes.

II. Methods

A. Overview of 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, [4]). The user wore the prosthesis as they would a conventional prosthesis, except that they were 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 five 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, featuring universal prosthesis adapters at each end, was cut 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 completed the protocol twice, with data reported for the second day, and subjects all had experience walking with the prosthesis emulator hardware (but not the controller used here) through previous experiments totaling at least 2 hours of walking. 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 $m \cdot s^{-1}$ or each subject's preferred walking speed (measured overground in a 50 m hallway) if it was less than 1.25 $m \cdot s^{-1}$. Subjects walked with their prescribed prosthesis (PRES) and with the prosthesis emulator in four modes: SACH (emulating a Solid Ankle Cushioned Heel foot), DER (emulating a Dynamic Elastic Response foot), BIOM (emulating the BiOM® T2 System), and HIPOW (a custom mode with high power output).

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 comfort and maximal performance. Metabolic

TABLE I

Human subject parameters

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

energy consumption was estimated using indirect calorimetry [7], 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. % change in net metabolic energy consumption and % change in net heart rate were computed relative to the level ground SACH condition, to quantify the marginal benefit of other conditions. User satisfaction was rated using a survey in which subjects reported their level of comfort with each of the emulated modes, considering 0 to be "as comfortable as the prescribed prosthesis", -10 to mean "walking is impossible", and $+10$ to mean "walking requires no effort". Maximum sustainable walking speed was established at the end of each walking trial by progressively increasing the speed of the treadmill in 0.05 $m \cdot s^{-1}$ increments every 10 s until the subject indicated they no longer felt they could sustain walking at the set speed for five more minutes. Measures of ankle torque and angle were calculated using on-board sensors.

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 [4].

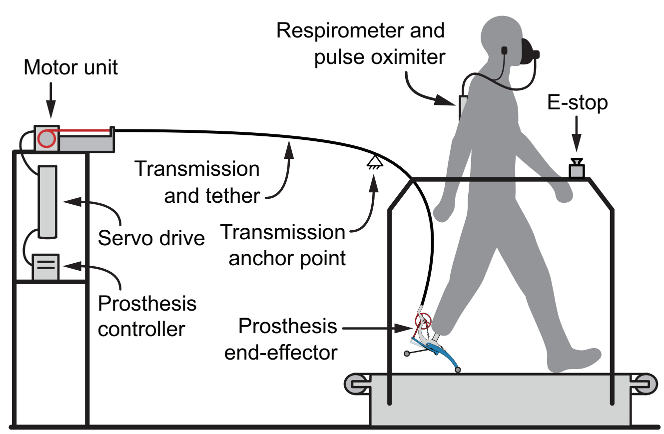

Fig. 1. The ankle-foot prosthesis emulator consists of a lightweight prosthesis worn by the user which is actuated through a flexible tether by a powerful motor and control system. By placing actuation and control off-board, the system can emulate an exceptional variety of behaviors at a worn mass comparable to passive mobile prostheses.

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Desired ankle torque ($\tau_{a,des}(\theta)$) was a piecewise linear fit to representative literature data obtained from inverse dynamics measurements made during walking ([8] for SACH; [9] for DER and BiOM® T2 System; Fig. 2). To switch the emulator from one mode to another, the experimenter simply 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 a simple proportional control on torque error (1).

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

k_p was tuned to best suit each mode's desired ankle impedance: when impedance was high (e.g., SACH mode, plantarflexion phase of HIPOW) larger k_p resulted in better tracking; and when impedance was low (e.g., DER mode, dorsiflexion phase of HIPOW) smaller k_p resulted in more stable torque tracking.

The combination of high torque and velocity during the plantarflexion phase of BiOM® T2 emulation proved challenging for this simple proportional control scheme, so an iterative learning controller was implemented to correct steady-state errors in joint torque (Eq. 2, inspired by [10]).

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

Learned torque on any given step (n) was a function of torque errors on previous steps (3).

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

k_l was tuned to eliminate steady-state tracking errors as quickly as possible without overshoot, roughly 30 strides as implemented here. This control design resulted in a form of integral controller, computing error between the reference and measured torques separately at each time interval, integrating across strides, and increasing the motor command proportionally.

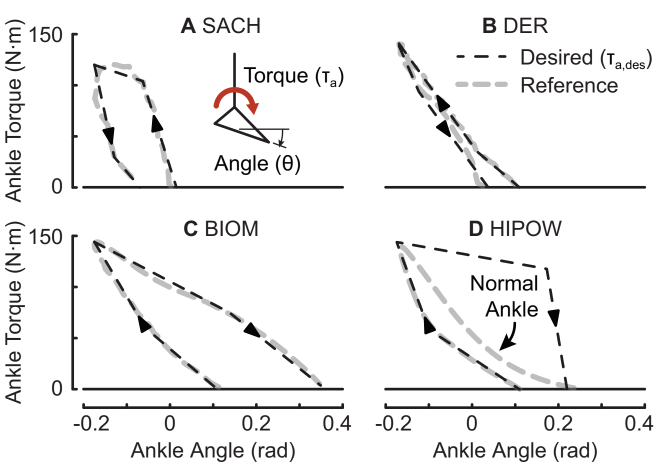

Fig. 2. Emulation was performed by matching the ankle torque vs. angle relationships of commercially-available prostheses. Ankle torque was controlled as a function of ankle angle, with different relationships for the dorsiflexion and plantarflexion phases of stance. The desired torque (dark dashed lines)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

Since ankle torque is very low during swing, the prosthetic ankle was position controlled, driving the ankle to the angle where the reference torque begins to build (θ_{des}, Eq. 4).

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

III. Results

A. Torque vs. Angle Control

Mean desired and measured prosthetic ankle torque trajectories are presented during the stance phase of the prosthetic ankle for a representative subject in Fig. 3. Root mean squared (RMS) error is presented to quantify torque tracking errors. Mean RMS error was 5 N·m, 2 N·m, 3 N·m, and 7 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 Table II. Subject #1, a DER user, always preferred and walked fastest with the robotic modes, despite these modes being metabolically suboptimal in level walking. Heart rate data were inconsistent with these observations, with passive modes always exhibiting the lowest heart rate. Subject #2, a DER user, always used the least energy and had the lowest heart rate in the robotic modes, but always preferred DER mode. The subject walked fastest in BIOM mode on level ground but walked fastest in DER mode when walking uphill. Subject #3, a DER user, always used the least energy and had the lowest heart rate in BIOM mode. The subject preferred the passive modes during level walking, despite walking fastest with BIOM mode, but in uphill walking he preferred and walked fastest in HIPOW mode. For subject #4, a BiOM® T2 user, DER mode was optimal for all metrics on level ground, with energy metrics shifting to BIOM mode for uphill walking, though preference remained at DER mode. Subject #5, a DER user, always preferred, walked fastest, and used the least energy with the robotic modes. Heart rate, uphill SACH, and uphill MSWS data were not available due to equipment failure.

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 prototypes we will improve sensor

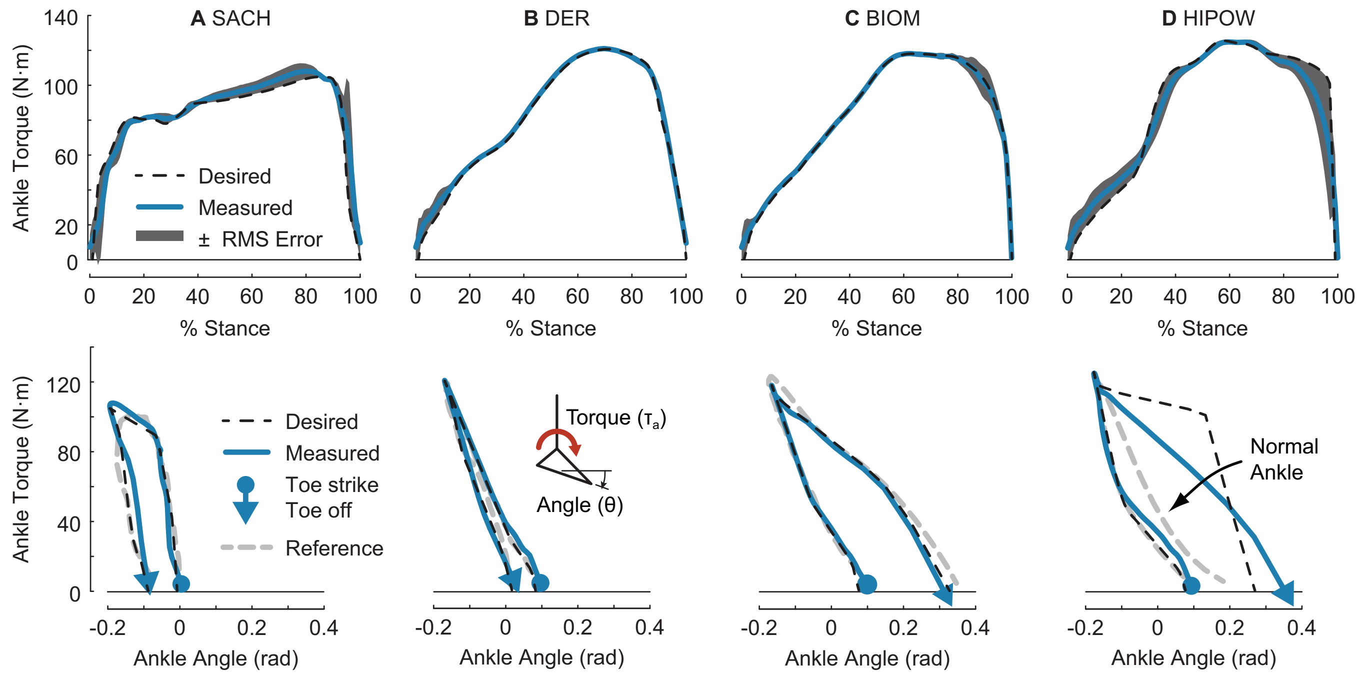

Fig. 3. Emulating commercial and hypothetical prostheses. The system has demonstrated high-quality haptic emulations of: **A** economical, but low-performance, solid-ankle cushioned heel (SACH) feet, which are stiff and dissipate significant energy, **B** mid-range dynamic-elastic response (DER) feet, which are more compliant and spring-like, **C** expensive, active robotic feet (BIOM), which use motors to generate positive work, and **D** conceptual designs, such as this high-powered robotic (HIPOW) foot 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 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.

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 change and variability in the timing of this state change led to reduced emulation quality near the state transition. In future prototypes we will eliminate the dorsi/plantar-flexion 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 for BIOM mode but also introduced dynamics that are likely not exhibited by the BiOM® T2 System. 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 prototypes of the emulator system we will work to improve feedback control torque tracking, through the means mentioned above as well as by implementing a derivative term in the feedback and mitigating deleterious effects of transmission friction and compliance.

We have demonstrated successful emulation of broadly different classes of device behavior, but it remains to be seen if the system demonstrated here can successfully differentiate subtle variations within a class of devices. The current emulator prototype can be programmed to exhibit such subtleties, but a controlled test 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, realistic adjustments in behavior will need to be made to ensure that prescription decisions are made using the best possible configuration for a given user. To this end, we are currently developing novel 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 difference that we will address in future prototypes. Two subjects had experience walking with a SACH foot. One reported: "[SACH mode was] stiff as a board! Felt 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.

Inconsistencies between emulated and real prostheses seem mostly to be related to choices made in the mechanical design of the prosthesis end-effector, rather than the choice of reference data or quality of torque tracking. For instance, one user reported, "Because the robotic foot [the emulator] is so stiff so I notice whenever I take a slightly off step. My prescribed foot is compliant in every direction so theres

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TABLE II
PERFORMANCE METRICS

	Condition		NW	Net Met. Rate		Net Heart Rate		User	MSWS
	Device	Slope	[J]	[W]	Change	[bpm]	Change	Score	[m · s^{-1}]
Subject #1	PRES	0°	n/a	260	-23%	22	-30%	0	1.85
	SACH	0°	-11	338	0%	32	0%	-5	1.70
	DER	0°	-8	314	-7%	32	+1%	-2	1.90
	BIOM	0°	10	320	-5%	36	+15%	-1	1.90
	HIPOW	0°	15	335	-1%	37	+15%	-2	1.90
	PRES	5°	n/a	280	-17%	11	-32%	0	2.05
	SACH	5°	-15	329	-2%	29	-7%	-5	1.85
	DER	5°	-8	320	-5%	27	-16%	0	1.85
	BIOM	5°	14	303	-10%	30	-4%	+1	2.00
	HIPOW	5°	18	293	-13%	25	+12%	0	1.90
Subject #2	PRES	0°	n/a	201	-24%	16	-26%	0	1.50
	SACH	0°	-7	263	0%	21	0%	+7	1.55
	DER	0°	-7	244	-7%	18	-15%	+8	1.50
	BIOM	0°	11	158	-40%	14	-36%	+5	1.60
	HIPOW	0°	21	198	-25%	12	-45%	+2	1.55
	PRES	5°	n/a	260	-1%	22	+3%	0	1.55
	SACH	5°	-10	276	+5%	23	+9%	+8	1.50
	DER	5°	-8	289	+10%	24	+14%	+9	1.60
	BIOM	5°	12	261	-1%	22	+3%	+7	1.50
	HIPOW	5°	14	265	+1%	24	+13%	+2	1.40
Subject #3	PRES	0°	n/a	182	-25%	20	-17%	0	1.60
	SACH	0°	-14	243	0%	24	0%	+6	1.50
	DER	0°	-9	245	+1%	24	+3%	+6	1.65
	BIOM	0°	14	192	-21%	20	-16%	+5	1.70
	HIPOW	0°	20	258	+6%	23	-2%	+5	1.75
	PRES	5°	n/a	149	-39%	11	-56%	0	1.60
	SACH	5°	-14	267	+10%	29	+21%	+7	1.55
	DER	5°	-9	261	+7%	30	+26%	+6	1.75
	BIOM	5°	12	225	-8%	9	-62%	+7	1.40
	HIPOW	5°	12	248	+2%	22	-6%	+9	1.75
Subject #4	PRES	0°	n/a	291	-28%	17	-54%	0	1.70
	SACH	0°	-12	407	0%	37	0%	-7	1.70
	DER	0°	-9	278	-32%	32	-14%	-1	1.75
	BIOM	0°	13	318	-22%	32	-12%	-3	1.75
	HIPOW	0°	18	357	-12%	35	-4%	-6	1.65
	PRES	5°	n/a	363	-11%	36	-2%	0	1.70
	SACH	5°	-14	459	+13%	45	+21%	-6	1.45
	DER	5°	-8	483	+19%	52	+41%	-3	1.55
	BIOM	5°	15	454	+12%	43	+18%	-4	1.60
	HIPOW	5°	19	493	+21%	49	+33%	-6	1.65
Subject #5	PRES	0°	n/a	274	-17%	n/a	n/a	0	2.00
	SACH	0°	-13	329	0%	n/a	n/a	-3	1.80
	DER	0°	-9	295	-11%	n/a	n/a	-1	1.75
	BIOM	0°	7	282	-14%	n/a	n/a	-1	1.90
	HIPOW	0°	4	304	-8%	n/a	n/a	-3	1.80
	PRES	5°	n/a	312	-5%	n/a	n/a	0	n/a
	SACH	5°	n/a	n/a	n/a	n/a	n/a	n/a	n/a
	DER	5°	-6	363	+10%	n/a	n/a	-3	n/a
	BIOM	5°	3	316	-4%	n/a	n/a	0	n/a
	HIPOW	5°	5	298	-9%	n/a	n/a	-2	n/a

more room for error." With just one degree of freedom, this prototype emulator cannot provide controlled compliance in all three linear and three rotational degrees of freedom. 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 implementing a more complete characterization of device behavior [11, 12] to support such a change.

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, though 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."

Several aspects of ankle-foot prosthesis behavior are not considered in our emulation scheme, which could affect outcomes. For simplicity, we opted to use representative stance-phase sagittal-plane ankle torque vs. angle data from previously published walking data. This common model of ankle behavior [13, 14] is limited as it contains only one degree of freedom, ankle plantar/dorsi-flexion, and does not consider the swing phase of gait. It is likely that this one dimensional model is sub-optimal in its prediction of the three forces and three moments that act on the user's residual limb. For instance, to emulate two prostheses with the same ankle impedance but different foot length would require an additional degree of freedom to control the reaction forces independent of the reaction moment. In future prototypes of the emulator system we will characterize the force/torque-deflection characteristics of the different commercially available prostheses through amputee-independent benchtop tests [11, 12] and through controlled walking trials. Also, although ankle joint torques are typically not considered significant during swing, prosthesis inertial and gravitational forces are thought to be. Metabolic energy consumption increases by about 8% per added kilogram at the feet [15], presumably a result of increased inertial forces during swing. Given that powered ankle-foot prostheses require extra mass for motors, batteries, and electronics, they tend to be roughly 1 kg heavier than passive prostheses, so this would reduce our expectation for the energetic benefits of the powered assistance strategies. Future versions of the prosthesis emulator will be designed to the weight of the lightest of ankle-foot prostheses (requiring a reduction in mass of about 30% compared to the current prototype) with dead-weight added to emulate the swing forces of heavier devices.

B. Utility of Performance Metrics

We demonstrated a protocol for measuring users' walking performance across emulator modes that discerned in-

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dividual users' 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. For three subjects, metabolic and heart rate were always minimized by robotic assistance, while for two others metabolic and heart rate were minimized by robotic assistance only when walking uphill. Two subjects always preferred and walked fastest in the robotic device modes, while one subject preferred the robotic modes only when walking uphill, and two others never preferred the robotic modes. Even when the robotic modes were not preferred, subjects tended to exhibit the highest walking speeds with robotic assistance. The four DER users tested all appear to have the potential to improve 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 the robotic assistance, but only when walking uphill and always preferred walking in the passive modes (DER and SACH). Despite being fortunate to have the most sophisticated technology available, it seems possible that the current prescription process falsely identified this user as one who would benefit from robotic assistance. The emulation approach to prescription could ensure that users reach an appropriate balance of cost and benefit.

Finally, 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 considerably different needs from the subjects tested here. We are currently recruiting and developing hardware to support a broader group of individuals for future tests of the emulator system.

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References

[1] C. Hofstad, H. Linde, J. Limbeek, and K. Postema, "Prescription of prosthetic ankle-foot mechanisms after lower limb amputation." *The Cochrane database of systematic reviews*, no. 1, p. CD003978, 2004.

[2] H. M. Herr and A. M. Grabowski, "Bionic ankle-foot prosthesis normalizes walking gait for persons with leg amputation." *Proceedings of the Royal Society of London B*, vol. 279, no. 1728, pp. 457–64, 2012.

[3] W. C. Flowers and R. W. Mann, "An electrohydraulic knee-torque controller for a prosthesis simulator." *Journal of biomechanical engineering*, vol. 99, no. 1, pp. 3–8, 1977.

[4] J. M. Caputo and S. H. Collins, "A Universal AnkleFoot Prosthesis Emulator for Human Locomotion Experiments," *Journal of Biomechanical Engineering*, vol. 136, no. 3, p. 035002, 2014.

[5] G. B. Spurr, A. M. Prentice, P. R. Murgatroyd, G. R. Goldberg, J. C. Reina, and N. T. Christman, "Energy expenditure from minute-by-minute heart-rate recording: comparison with indirect calorimetry." *The American journal of clinical nutrition*, vol. 48, no. 3, pp. 552–9, 1988.

[6] J. J. Genin, G. J. Bastien, B. Franck, C. Detrembleur, and P. A. Willems, "Effect of speed on the energy cost of walking in unilateral traumatic lower limb amputees." *European journal of applied physiology*, vol. 103, no. 6, pp. 655–63, 2008.

[7] J. M. Brockway, "Derivation of formulae used to calculate energy expenditure in man." *Human nutrition, clinical nutrition*, vol. 41, no. 6, pp. 463–71, 1987.

[8] L. Torburn, J. Perry, E. Ayyappa, and S. L. Shanfield, "Below-knee amputee gait with dynamic elastic response prosthetic feet: A pilot study," *The Journal of Rehabilitation Research and Development*, vol. 27, no. 4, p. 369, 1990.

[9] A. E. Ferris, J. M. Aldridge, C. a. Rábago, and J. M. Wilken, "Evaluation of a powered ankle-foot prosthetic system during walking," *Archives of physical medicine and rehabilitation*, vol. 93, no. 11, pp. 1911–8, 2012.

[10] J. Zhang, C. C. Cheah, and S. H. Collins, "A systematic comparison of nine prominent torque control methods in a tethered ankle exoskeleton with series elastic actuation during human walking," *International Journal of Robotics Research*, vol. submitted, pp. 1–16, 2014.

[11] M. J. Major, M. Twiste, L. P. J. Kenney, and D. Howard, "Amputee Independent Prosthesis Properties–a new model for description and measurement." *Journal of biomechanics*, vol. 44, no. 14, pp. 2572–5, 2011.

[12] P. G. Adamczyk, M. Roland, and M. E. Hahn, "Novel method to evaluate angular stiffness of prosthetic feet from linear compression tests." *Journal of biomechanical engineering*, vol. 135, no. 10, pp. 104 502–5, 2013.

[13] A. H. Hansen, D. S. Childress, S. C. Miff, S. A. Gard, and K. P. Mesplay, "The human ankle during walking: implications for design of biomimetic ankle prostheses." *Journal of biomechanics*, vol. 37, no. 10, pp. 1467–74, 2004.

[14] K. Shamaei, G. S. Sawicki, and A. M. Dollar, "Estimation of quasi-stiffness and propulsive work of the human ankle in the stance phase of walking." *PloS one*, vol. 8, no. 3, p. e59935, 2013.

[15] R. C. Browning, J. R. Modica, R. Kram, and A. Goswami, "The effects of adding mass to the legs on the energetics and biomechanics of walking." *Medicine and science in sports and exercise*, vol. 39, no. 3, pp. 515–25, 2007.