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Technical note

A motor-driven adjustable prosthetic socket operated using a mobile phone app: A technical note

Joan E Sanders[a,*], Joseph L Garbini[b], Jake B McLean[a], Paul Hinrichs[a], Travis J Predmore[a], Jacob T Brzostowski[a], Christian B Redd[a], John C Cagle[a]

[a] *Department of Bioengineering, University of Washington, Seattle, WA 98195, United States*
[b] *Department of Mechanical Engineering, University of Washington, Seattle, United States*

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A B S T R A C T

Sockets that allow incremental size adjustment during ambulation may help prosthesis users improve management of their changes in limb volume and the quality of their prosthetic fit. A platform system was developed that allowed people with trans-tibial limb loss to adjust the radial positions of socket panels during ambulation in small increments via a motor mounted beneath the socket. The motor altered the length of a cable running through the socket panels according to commands communicated from a mobile phone. A proportional–integral–derivative controller adjusted the voltage applied to the motor via pulse-width modulation to achieve target settings. Bench test results showed that when the system was subjected to loads comparable to those expected during clinical use, maximum absolute steady state error was 0.036 mm. Treadmill testing on 16 people with trans-tibial limb amputation demonstrated that the range of cable lengths over which participants deemed fit clinically acceptable varied between 24 mm and 114 mm depending on the user. In field testing 11 of 13 participants were comfortable making socket size adjustments while walking. The developed system achieves incremental socket size adjustments appropriate for research and development of ambulatory adjustable sockets.

1. Introduction

The quality of socket fit is commonly acknowledged by both patients and prosthetists as the most important aspect of a prosthesis [1–3]. An important source of socket fit problems is changes in volume of the residual limb that occur over minutes or hours. Volume changes of just 1% are clinically perceptible and may necessitate sock accommodation [4].

Adjustable sockets are a simple means for users to compensate for limb volume changes, making the socket tighter or looser depending on how the socket feels [1,2]. It is possible that incrementally increasing socket size during ambulation will facilitate limb fluid volume recovery and retention, based on results from studies demonstrating that most participants gained limb fluid volume when they walked [5,6]. If increasing socket size during ambulation did facilitate limb fluid volume recovery without causing gait instability then it may be a clinically useful accommodation technique to improve outcomes in active users. To test this prospect, a means for adjusting socket size during ambulation and recording adjustments is needed.

Commercial systems to adjust socket size exist but they are manually controlled thus are not appropriate to adjust during ambulation, and they do not record socket changes. Liquid-filled bladders (Socket Comfort System, Simbex, New Hampshire); adjustable panels held in the socket with a cable (RevoFit, Click Medical, Steamboat Springs, Colorado); and sockets made up of vertical struts with Velcro straps wrapped around the socket perimeter (Lim Innovations, San Francisco, California) have been created.

The purpose of this technical note is to describe a platform system for executing adjustments via a mobile phone app and storing adjustment data. The platform is an extension from existing cabled-panel socket technology. It was bench tested to characterize cable length adjustment error, and then it was tested on participants with transtibial limb amputation to determine the range of cable lengths over which users considered socket fit acceptable. We investigated if users were comfortable making adjustments during walking using the mobile phone app.

* Corresponding author.
E-mail addresses: jsanders@uw.edu (J.E. Sanders), garbini@uw.edu (J.L. Garbini), jakemc@uw.edu (J.B. McLean), hinrichs@uw.edu (P. Hinrichs), tpred@uw.edu (T.J. Predmore), credd@uw.edu (C.B. Redd), jcagle@uw.edu (J.C. Cagle).

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2. Methods

2.1. Mechanical design

The designed system was an extension of a manual cabled-panel adjustable socket system developed previously (RevoFit, Click Medical, Steamboat Springs, Colorado). However, instead of using a manual ratcheting mechanism to change cable length and adjust socket size, the system used a motor assembly positioned beneath the trans-tibial socket that took up or released cable on a spool to adjust socket size (Fig. 1(a) and (b)). The cable was routed through the socket panels and into a custom cable connector laminated into the socket that aligned the cable with a take-up spool of a motor assembly in an enclosure below the socket. The motor assembly included a packaged commercial motor unit – a brushed direct current (DC) motor, encoder, and planetary gear set. A motor unit was used because compared with selecting and assembling individual parts, it was more compact and cost effective, and allowed a simpler controller. The motor unit selected (Table 1) was chosen based on the torque necessary to effect a socket size change on prosthesis users during ambulation while at the same time not back-driving the motor when no power was applied. The motor needed to be small enough and lightweight to fit beneath the socket. Different candidate motors were evaluated during preliminary testing on participants with trans-tibial limb amputation. Those participants were required to be at least 18 months post amputation, have good sensation, be at a K level 3 or 4, and capable of walking for 20 min without walking aides (cane, walker). An institutional review board (IRB) approved protocol was used, and informed consent was obtained before any studies were initiated. Testing was conducted with the participant walking on a treadmill at a self-selected walking speed while wearing an overhead harness for safety (Fig. 1(c)).

A motor unit with a continuous output torque of 1.2 N m, encoder with 64 counts per turn of the motor shaft, and a gearhead ratio of 7078:1 was selected (Table 1). The encoder resolution of 64 counts per turn of the motor shaft combined with the gearhead ratio (7078:1) produced a resolution of 452,992 counts per revolution of the output shaft. Using a 6.05 mm diameter spool, the cable length take-up was 38.01 mm per revolution. The total cable length for a trans-tibial amputee socket is typically about 90–130 cm. The take-up spool and gear motor were connected together using a custom steel shaft coupler. To protect the motor from radial loads, the spool was isolated with bearings at each end. The motor unit, coupler, spool, and bearings were assembled within a custom frame held within an aluminum box (14.7 cm × 6.3 cm × 3.9 cm). The box had a standard 5.1 cm bolt circle for connection to the user's normal prosthetic ankle/foot componentry.

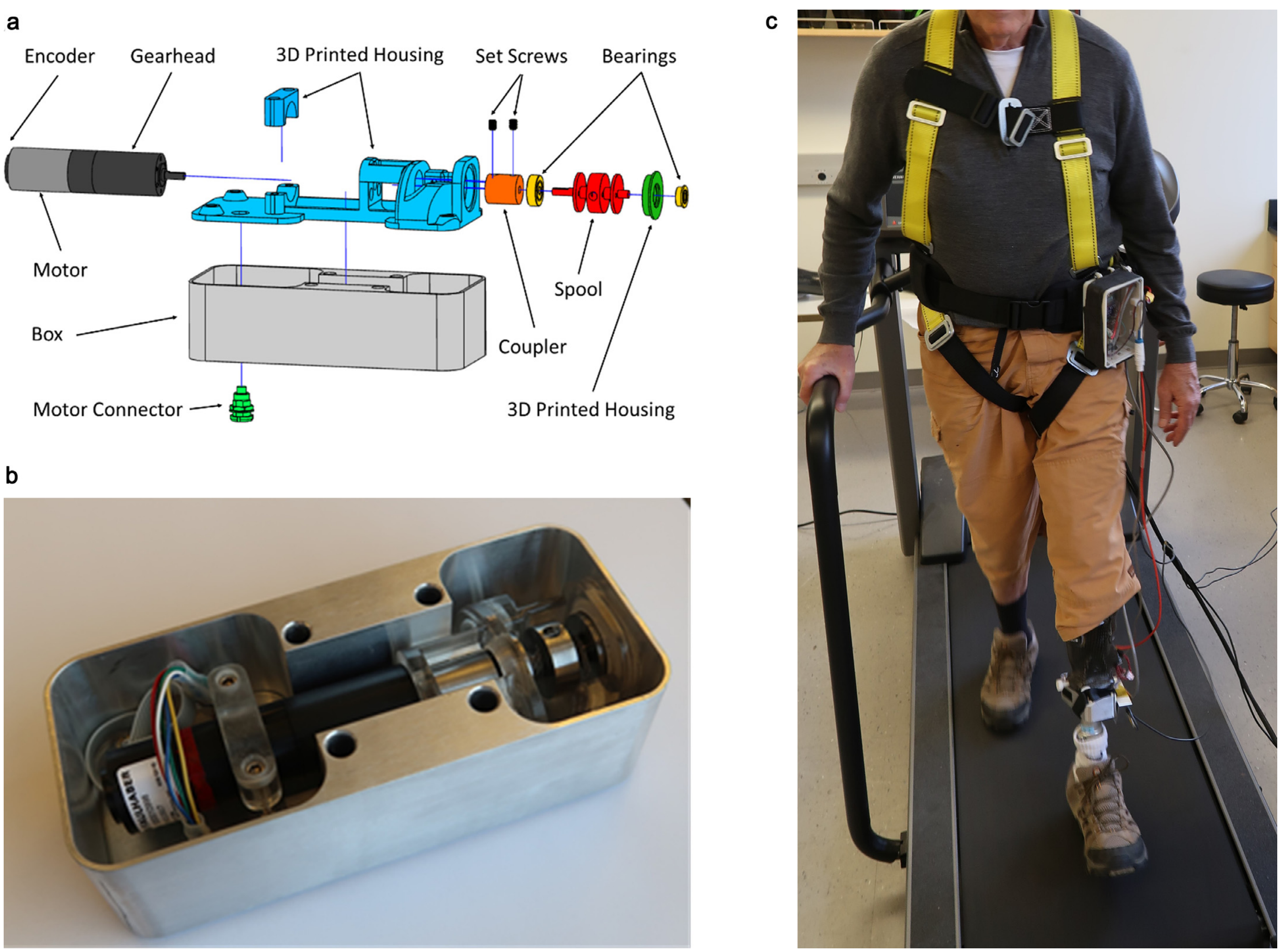

Fig. 1. System motor assembly. (a) Schematic of system components. (b) Assembled system. (c) Participant using the motorized adjustable socket in laboratory testing wearing a harness for safety.

Table 1
List of primary components.

Component	Part	Manufacturer
Brushed DC motor, encoder, and planetary gear set	1000009583	Micromo, Clearwater, Florida
Cable	Braided Vectran 400	Twinline LLC, Castle Rock, Colorado
Bearings to isolate spool	7804K103 and 7804K129	McMaster-Carr, Elmhurst, Illinois
Coupler/spool/box		Custom design
Socket distal end plate	VMP-001	Ossur, Reykjavik, Iceland
Magnetoresistive memory	MR25H256 256-kilobit	Everspin Technologies, Chandler, Arizona
Bluetooth module	BLE113-A-V1	Silicon Labs, Austin, Texas
Data acquisition/mobile phone communication ARM Cortex-M0+ microcontroller	LPC11U67JBD48E	NXP Semiconductors, Eindhoven, Netherlands
Control system microcontroller	Arduino Due	Arduino LLC, Somerville, Massachusetts
μSD card	MB-ME32DA/AM	Samsung, Seoul, South Korea

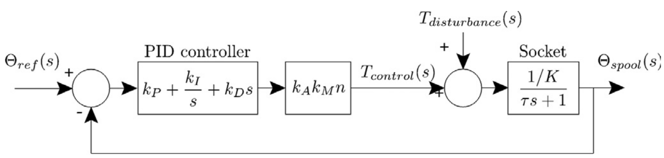

Fig. 2. Diagram of spool angle control loop.

2.2. Spool control

As a starting point during initial testing, the spool rotation was set to induce step displacements of the panels that were approximately equivalent to the thickness of adding a 1-ply cotton sock to the insides of the panels (0.45 mm) [7]. This distance required a cable length change of approximately 4.75 mm [8], thus a spool axis rotation of 45°. The step increment could later be modified as needed based on test results from amputee participants; this setting was considered preliminary.

A digital implementation of a continuous Proportional–Integral–Derivative (PID) feedback controller was used to govern spool motion. A PID controller was selected because the stiffness and damping properties of the socket panels varied during use. In addition, panel positions needed to be maintained regardless of external torque disturbances such as friction and cyclical loading by the residual limb during walking. Finally, useful operation required tracking the geometric state of the socket, so that data could be presented to the user and could be restored after power-down and re-donning. A continuous time block diagram of the system is shown in Figure 2.

The reference angle $\theta_{ref}(t)$ was set by the person wearing the prosthesis when they selected a desired socket size using the mobile phone app. Operating on the error signal $\theta_{ref} - \theta_{spool}$, the PID controller acted continuously to maintain the output spool angle, while rejecting the torque disturbances. In the center block, the control torque $T_{control}$ was determined as the product of the PID output, the transconductance gain of the motor driver k_A, the motor torque constant k_M, and the gear ratio n. The torque acting on the cable spool was the sum of the controller torque $T_{control}$ and the disturbance torque $T_{disturbance}$. For purposes of design, the socket was modeled as a first-order spring/damper system. The time constant was $\tau = B/K$, where B and K were the effective torsional damping and stiffness of the socket, as measured at the spool. Inertial effects were negligible.

Practical implementation of the control scheme required that it differ significantly from the idealization of Figure 2 in two ways: First, during large commanded changes in position (e.g. 45°), the Integrator term k_I/s of the PID block would anomalously accumulate a large value. Uncorrected, this well-known phenomenon of "integrator wind-up" lead to excessive overshoot of the spool final position. An anti-windup algorithm overcame this limitation by effectively disabling the Integrator term until the actuator fell out of saturation near the final value. Second, power consumption by the motor ultimately limited operating time of the system before recharging was necessary. To conserve power, current to the motor was stopped when the absolute value of the error signal was less than a prescribed stopping value. Because of the very high gear ratio, disturbance torques caused by donning, walking, etc. could not back-drive the spool. Therefore, the panels retained their final positions without power to the motor. This control scheme and the user interface were implemented in discrete time using the microcontroller. The PID controller received spool position feedback estimated from the motor encoder.

In the specific implementation, the PID controller design (k_P, k_I, k_D) was based on an empirical measurement of the socket time constant τ, and the performance requirements of the 45-degree step. In particular, the design target was that the small-step rise time ($T_r = 1$ s) should be a fraction of the 45-degree step slew time (5 s). A moderate overshoot (∼20%) was allowed to improve step efficiency. In its discrete form, the controller sample rate was 32 Hz (>50 times the closed-loop bandwidth); and the anti-integrator windup algorithm was set to enable the PID function within 0.40° of the final spool value. Finally, the stopping value was set to 0.20°, well below the smallest detectable change in the spool position.

A non-volatile system memory was used to retain motor position during power-down. A magnetoresistive memory was selected because it was low power, had fast write cycles, and had unlimited read and write endurance, allowing for constant write-back of system state without the additional complexity of wear-levelling algorithms. The memory selected (Table 1) used the industry-standard serial peripheral interface (SPI). Motor position was continuously written into this memory, and upon restart read back from this memory.

2.3. Mobile phone interface

Bluetooth Low Energy (BLE) was implemented to communicate commands wirelessly from a mobile phone to the micro-controller, and to communicate back to the user the current size of the prosthetic socket. BLE supports the Generic Attribute Profile (GATT) which provided a framework for data exchange between the devices. On the microcontroller side a Bluetooth module was interfaced to an ARM Cortex-M0+ microcontroller, which was part of a data logging board developed previously that aggregated data from on-board and external sensors and recorded it to text files on an attached microSD card [9]. The board included a Bluetooth Low Energy (BLE) communications module, allowing for wireless interaction and control via a mobile phone app. API-based (application programming interface), high level control of the Bluetooth link was hosted on the ARM microcontroller and communicated to the Bluetooth module via UART serial communication (Fig. 3(a)).

An Android application was created for the mobile phone to exchange data with the wireless module. A flowchart for operation is

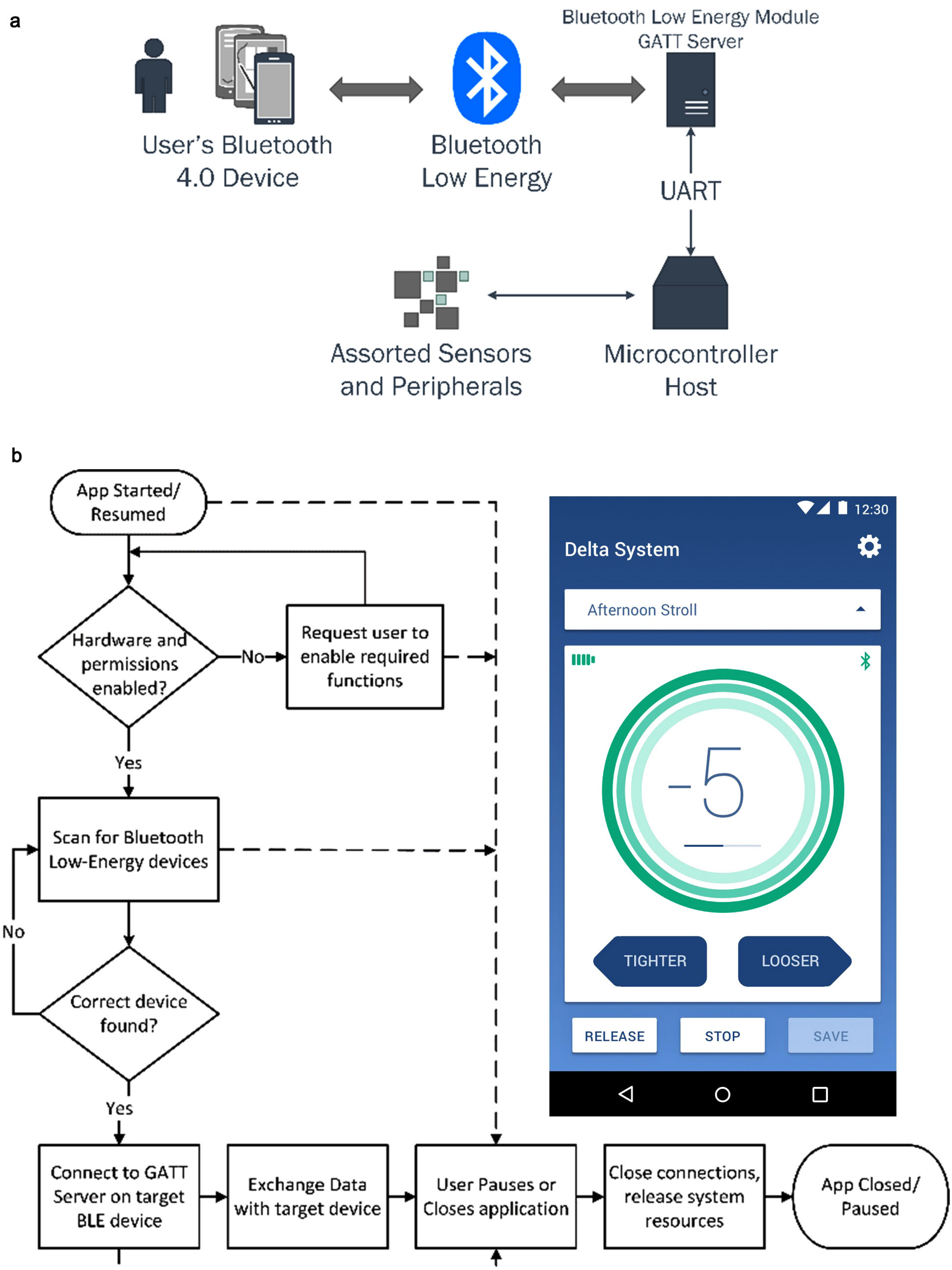

Fig. 3. System communication components and operation. (a) Block diagram of wireless communication. (b) Flowchart for mobile app, and mobile phone user interface.

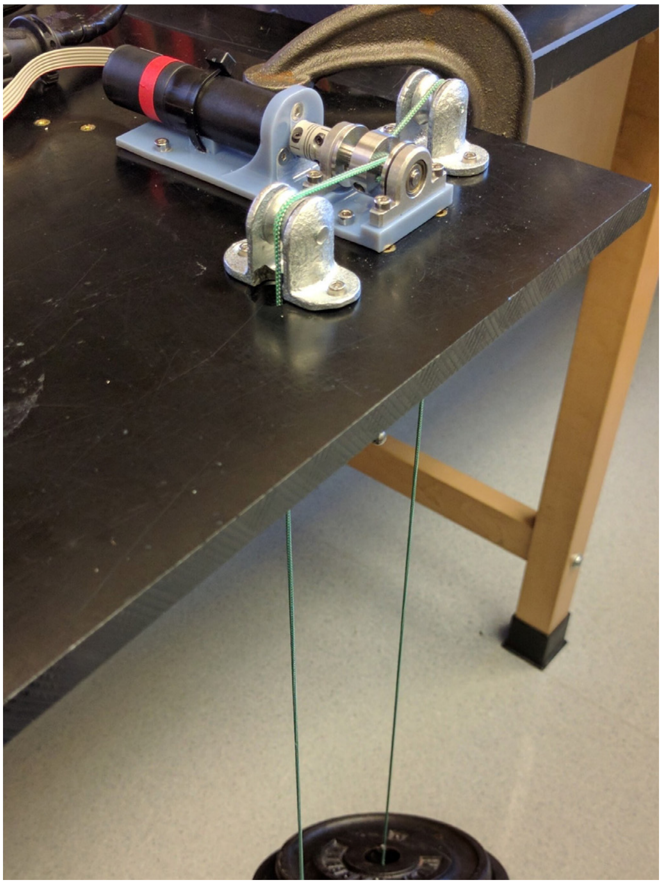

Fig. 4. Bench test apparatus.

shown in Figure 3(b). When the user pushed the button to command "TIGHTER," the cable drew in 1 unit (45-degree spool rotation) and displayed to the user a 1-increment adjustment. The user could also stack commands, such as pressing "TIGHTER" three times, then the spool would move 135°. Cancelling any command by pressing "STOP" stopped at the current position even if it was between steps. A subsequent tighten or loosen command moved to the nearest 45-degree step. When the user closed the application, the program disconnected from the Bluetooth module and released system resources to conserve battery life.

2.4. Evaluation

Bench testing was conducted to characterize the sensitivity of cable length adjustment to variables expected to affect performance – applied load and direction of adjustment. The motor, gearhead, coupler, and spool were mounted to a tabletop so that weights could be applied to the cable to simulate loads from the residual limb on the panels (Fig. 4). Motor encoder positon was recorded continuously while weights of 0.0, 88.3, and 178.5 N, corresponding to torques of 0.00, 0.27, and 0.54 N m, were applied. These magnitudes were 0, 45, and 89 percent, respectively, of the maximum allowable continuous torque specified by the manufacturer of the motor assembly. The highest value (0.54 N m) was selected to test the upper limit of the developed system.

Two series of bench tests were conducted. First, adjustments at the smallest possible step size, a 1-degree spool adjustment, were made to characterize the highest possible resolution the system could achieve. Then adjustments at the step size used for participant testing, a 45-degree spool adjustment, were evaluated. In each test at each applied torque, twenty adjustments were made, ten winding the spool simulating tightening the panels and ten unwinding the spool simulating loosening the panels. Thus a total of 60 conditions were tested for each of the two step sizes. Each adjustment was executed after the previous adjustment had reached a steady state position. Absolute steady state error was calculated as the root-mean-square error from each test. A two-way analysis of variance was used to assess the interaction between torque and direction (tighten or loosen). The dependent variable, absolute steady state error, was tested against independent variables torque with three levels (0.0, 0.27, 0.54 N m) and direction with two levels (tighten, loosen).

Testing on people with limb loss was conducted to determine the ranges of cable lengths over which users considered socket fit acceptable, and if users were comfortable making adjustments during walking using the mobile phone app. Sockets were fabricated with three socket panels, located anterior medial, anterior lateral, and posterior. Panel dimensions were specified by the research prosthetist. Sockets were shaped the same as participant normal sockets. A coordinate measurement machine (FARO Platinum Arm, Faro Technologies, Lake Mary, Florida) was used to measure participants' normal socket shape, then a 3-ply carbon-fiber socket was made over a computer-manufactured positive of the shape (C5 carver, Provel, Cle Elum, Washington). A low deformation cable (3.8% elongation at break, tensile strength of 3.2 GPa) was used to minimize stretch elongation and radial compression during operation (Table 1). Low radial compression was relevant because when socket adjustments were made that required more than approximately 900° of spool rotation, the cable began wrapping on top of itself, changing the length of cable drawn per 45-degree increment from that recorded by the encoder. Low radial compression helped ensure the magnitude of change was not dependent upon the applied load.

Once the test socket was completed, shank/foot componentry similar to that normally used by the participant was added. Care was taken to ensure prosthetic alignment was maintained. Participants did not use prosthetic socks when wearing the test sockets. The neutral panel position was defined as cable length with the panels in the plane of the surrounding socket.

Wearing a harness for safety (Fig. 1(c)), participants walked on a treadmill at self-selected walking speeds and verbalized to the researcher appropriate socket size adjustments ("tighter" or "looser") so as to optimize socket comfort/fit. The researcher executed one single step adjustment (one 4.75 mm cable length change) at a time. Once a user-selected optimum was reached, the researcher then slowly increased socket size until the participant deemed fit unacceptable. The motor took approximately 8.5 s to effect one step (4.75 mm) of cable length adjustment. The researcher then reduced socket size until the participant deemed the socket too small and fit unacceptable. The researcher then returned the socket size back to the user-specified optimal setting, and adjustments within the user-defined socket size range were made so that the participant gained experience wearing the system. Collected data were processed to characterize the cable length range of acceptable socket sizes.

On a separate test day participants operated the adjustable socket using the mobile phone app while outside the research building in pedestrian designated areas. They made adjustments as they saw fit during two 20-minute sections of ambulation separated by a 5-min sit.

3. Results

The analysis of variance on bench test data demonstrated significant interaction between torque and direction ($p = 0.013$) but the absolute differences were small (Table 2). The maximum absolute steady state error for the 1-degree spool adjustment was 0.19°, cor-

Table 2
Absolute steady state spool angle error from bench testing.

Spool angle step size (degrees)	Applied torque (N m)	Absolute stead state error (degrees)	
		Tightening (mean (s.d.), median)	Loosening (mean (s.d.), median)
1.0	0.00	0.12 (0.02), 0.13	0.16 (0.01), 0.15
1.0	0.27	0.04 (0.02), 0.03	0.18 (0.02), 0.19
1.0	0.54	0.18 (0.02), 0.18	0.19 (0.01), 0.19
45.0	0.00	0.13 (0.02), 0.13	0.12 (0.02), 0.12
45.0	0.27	0.12 (0.05), 0.13	0.10 (0.02), 0.10
45.0	0.54	0.30 (0.04), 0.31	0.34 (0.03), 0.35

Table 3
Participant socket data.

Partic. #	Socket volume (mL)	Panel dimensions: posterior, ant medial, ant lateral (cm)	User-selected cable draw range (mm)	# steps in range	Phone use OK while walking outside?	Open-ended question response
1	1797	13.0 × 5.2, 15.5×4.5, 10.5 × 4.0	104.50	22	Y	Requested smaller step size
2	1464	9.7 × 5.0, 9.5 × 4.5, 7.5 × 3.7	95.00	20	Y	
3	1653	7.2 × 5.0, 10.0 × 5.0, 7.8 × 3.6	99.75	21	Y	
4	1282	9.0 × 3.9, 11.5 × 4.0, 7.8 × 3.6	95.00	20	Y	
5	1273	7.5 × 4.4, 9.5 × 4.5, 7.0 × 3.5	61.75	13	Y	
6	1481	8.0 × 6.0, 9.0 × 4.7, 6.6 × 3.6	76.00	16	Y	
7	1184	7.5 × 4.7, 10.0 × 4.1, 6.5 × 4.6	76.00	16	Y	
8	1345	9.2 × 4.4, 10.0 × 4.5, 8.6 × 2.9	114.00	24	Y	Requested larger step size
9	1162	9.0 × 4.4, 9.3 × 3.8, 7.4 × 3.5	90.25	19	Y	
10	1483	8.2 × 6.0, 9.5 × 5.5, 7.0 × 3.9	42.75	9	Y	Requested smaller step size
11	2591	12.3 × 6.1, 14.5 × 5.4, 12.0 × 6.0	23.75	5	Y	Requested smaller step size
12	1224	8.0 × 4.8, 9.3 × 4.1, 6.3 × 4.0	42.75	9	Y	
13	1277	7.5 × 5.6, 9.0 × 5.0, 9.0 × 3.1	71.25	15	N	Did not want panels moving while walking
14	1409	6.7 × 5.6, 9.2 × 6.0, 6.8 × 3.2	80.75	17	N	Even with own phone, would not use mobile phone while walking
15	881	6.0 × 6.5, 7.5 × 4.5, 7.0 × 3.5	Not tested	Not tested	Y	
16	949	4.4 × 6.0, 8.3 × 5.5, 8.0 × 3.0	66.50	14	Y	Requested smaller step size

responding to a cable length error of 0.020 mm. For the 45-degree spool adjustment the maximum absolute steady state error was 0.34°, corresponding to a cable length error of 0.036 mm. Both of these cases were from the 0.54 N m applied torque condition for the loosening direction.

Sixteen participants tested the adjustable socket system in the lab. Socket volumes, panel dimensions, user-selected cable draw ranges, and the number of adjustments within user-selected ranges are summarized in Table 3. Ranges of cable length adjustments over which participants considered socket fit acceptable varied between 24 mm and 114 mm (mean 76 (s.d. 25) mm, median 76 mm) depending on the participant. It took approximately 8.5 s to execute one step adjustment (4.75 mm), thus total time to run the panels across the entire user-selected displacement range ranged from approximately 42 s to 204 s. Example motor position data from a test in the lab on the treadmill are shown in Figure 5.

The panels were at their neutral positions at a spool rotation of approximately 720°. The cable started wrapping onto itself at a spool rotation of approximately 900°, which was at adjustments tighter than 4 steps inward from the neutral position. In laboratory testing, 11 participants selected minimum socket sizes, i.e. the tightest they considered acceptable, that were more than 4 steps inward from the neutral position, Using the cable thickness (1.04 mm) and assuming the worst case that cable stacked directly on top of itself, cable lengths were greater than the encoder measured value at users' smallest socket size by an average of 4.14 (s.d. 2.47) mm.

In testing outside of the lab in pedestrian-designated areas, 11 of the 13 participants were comfortable operating the phone app while walking (Table 3). One of the participants was uncomfortable walking while looking down at any phone. The other participant was uncomfortable with the panels moving while she was walking. Three of the four participants who noted that they would prefer a shorter step cable length adjustment had relatively few steps in their cable length range (5, 9, 14), and one participant who indicated that he would prefer a larger step cable adjustment had a relatively large number of steps in his cable length range (24).

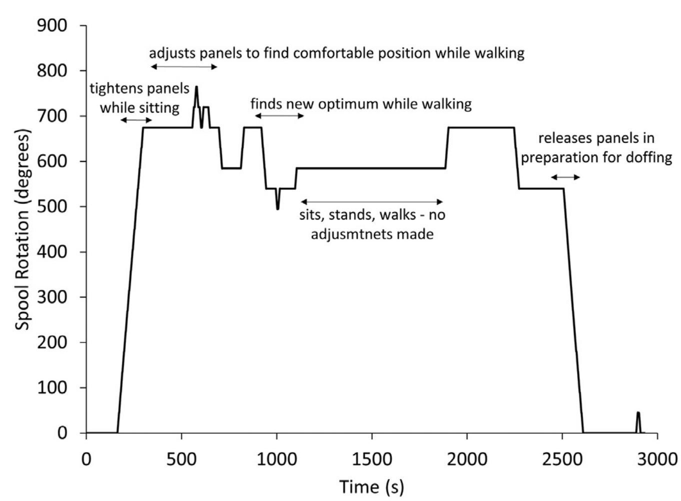

Fig. 5. Example motor position results from a test session.

Monitoring of electrical current from the motor during bench testing demonstrated the system used 1.03 W during steady state, and 1.35 W when an adjustment was being made. Using a 23.68 W h battery, the system had sufficient power to operate continuously for at least 16.5 h, sufficient for daylong testing.

4. Discussion

By implementing small reductions and enlargements using cabled-panel adjustable sockets, the developed platform system allows clinical investigation to determine if incremental adjustment during ambulation improves prosthetic fit and user management of limb fluid volume.

Absolute steady state errors measured during bench testing were consistent with parameter settings implemented in the controller. Steady state errors for 1-degree step sizes were within the specified stopping value of 0.20°. For the 45-degree step size, errors were within the stopping value for the 0.00 and 0.27 N m applied torques. Results for the 0.54 N m torque were slightly outside the 0.20-degree range because this torque approached the current limit of the amplifier. The 0.34-degree error at 0.54 N m applied torque corresponded to a 0.036 mm cable length error, which was expected clinically irrelevant. Since the encoder measurement was absolute, this error did not accumulate upon execution of additional steps or of multiple steps in one continuous adjustment. During ambulation, a constant applied torque would be unlikely because of reduced limb-socket interface loading during swing phase, thus the bench tests were considered a worst case scenario.

For the participant sockets tested in this research, cable overlapping on the spool at small socket sizes caused an overestimate of cable length in the controller when socket sizes tighter than approximately 4 steps from neutral were selected. The error accumulated as more tightening steps were executed. Depending on the clinical relevance of the error, efforts may be needed to either reduce overlap, possibly through mechanical design modification, for example enlarging the diameter of the spool or using a longer spool, or through calibration and then subtraction of the error from encoder data during operation.

Results from laboratory testing using a 45-degree step size demonstrating that 4 of 16 participants preferred smaller step sizes than the system provided indicates that adjustment smaller than radial displacements equivalent to a 1-ply sock (0.45 mm) are desired by some users. Given that the absolute steady state error of the system under the smallest spool degree increment (1°) was 0.19° (0.020 mm cable length adjustment), smaller step sizes should be within the capabilities of this platform system. Since different users preferred different step sizes, adjustable step size may be an important feature to implement in a clinical adjustable socket system.

Ranges of socket adjustment varied considerably among test participants (24 mm to 114 mm). Physical characteristics of the residual limb such as size and stiffness are likely important to this range. Based from experience in this research, we expect that balance and confidence of the user are also key factors. For some users, particularly people with limited protective sensation, cable length adjustment ranges may improve safety and prevent overtightening that may cause injury, via settings in the mobile phone app, for example.

Clinical studies using the developed platform system may provide useful insight towards automated control of a number of powered adjustable-size sockets under development, including systems with liquid-filled bladders and coiled tubes [10–12], and air-filled bladders [13–15]. As these technologies advance and their capabilities to improve socket fit are tested, consideration should also be given to the impact of adjustment on tissues within the residual limb. Internal tissue loads were demonstrated highly sensitive to limb anatomy, limb tissue characteristics, and socket design [16–18] but the impact of socket size adjustments executed by people with limb amputation in their daily lives is unknown.

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Competing interests

None declared.

Ethical approval

Approval was given by a University of Washington Institutional Review Board (IRB), reference #49624.

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