Effect of a Swing-Assist Knee Prosthesis on Stair Ambulation

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***Abstract*—This paper describes stair ambulation control and functionality of a semi-powered knee prosthesis that supplements nominally passive prosthesis behavior with swing-phase assistance. A set of stair ascent and descent controllers are described. The controllers were implemented in a semi-powered prosthesis prototype, and the prospective benefits of swing assist in stair ambulation were assessed on a group of three participants with unilateral, transfemoral amputation, relative to their respective daily-use prostheses. Results indicate that ambulation with the semi-powered knee resulted in improved stair ascent gait symmetry when compared to the participants' passive daily-use devices, and increased similitude to healthy stair ascent movement.**

***Index Terms*—Transfemoral, prosthesis, power-asymmetric, stair, powered, biomechanics, amputation.**

I. Introduction

STAIR ascent and descent are known to be challenging for persons with transfemoral amputation. Stairs are unavoidable, however, for many with lower limb amputation, since they exist in nearly half of domestic settings [1], many occupational buildings, and in instances of emergency egress.

People with transfemoral amputation typically use energetically passive knee prostheses. The more advanced of these passive prostheses, microprocessor-controlled knees (MPKs), employ a microcontroller and some form of modulated dissipator to provide variable levels of damping throughout gait. In this way, they are able to produce the high resistance against flexion needed for stance, and a low resistance to movement for swing. This combination of behaviors works well for stair descent, although it is more compromised during stair ascent (as discussed subsequently).

When ascending stairs, a prosthesis user can employ one of the following approaches: 1) a single-step, step-to ascent; 2) a double-step, step-to ascent; or 3) a step-over ascent. In a step-to approach, the swing phase of the prosthetic side typically

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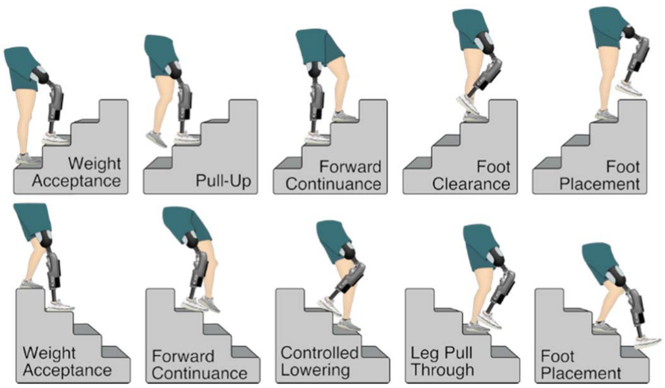

Fig. 1. The phases of step-over stair ambulation, as described in [5].

involves little to no knee movement. As a result, minimal stance knee extension is required. However, the user must often employ compensatory actions using some combination of hip circumduction, ankle vaulting, and/or tilting of the pelvis to bring the affected foot to the stance stair [2].

Some users are also capable of performing a step-over, stair ascent gait with a passive prosthesis. When doing so, users produce appropriate stance knee extension (i.e., weight acceptance through forward continuance, as seen in Fig. 1) by exerting hip torque, supplemented by handrail assistance, to extend the stance knee (i.e., the knee is extended indirectly through hip extension). Specifically, as discussed in [3], due to the frictional constraint between the stance foot and ground, the stance leg is essentially a kinematically closed chain, and therefore a user has control of knee joint movement during the stance phase of gait via his or her hip movement. The user, however, has less control of knee angle during swing, since knee motion in swing can only be generated by the inertial coupling between the thigh and shank (i.e., prosthesis). This inertial control of knee angle works well during level ground walking, but generally does not provide a suitable swing knee trajectory during stair ascent. Specifically, the hip motion associated with swing phase in stair ascent does not generate a knee motion that allows the foot to clear the subsequent step without compensatory behaviors such as hip circumduction, pelvic tilt, and/or ankle vaulting. It should be noted that one MPK, the Ottobock Genium knee, provides an alternative means of obtaining swing phase motion for stair ascent, in which the user accelerates his or her thigh in a posterior direction prior to subsequently accelerating it in an anterior direction [4]. Although this enables sufficient knee flexion, it requires hip movement that deviates substantially from healthy movement.

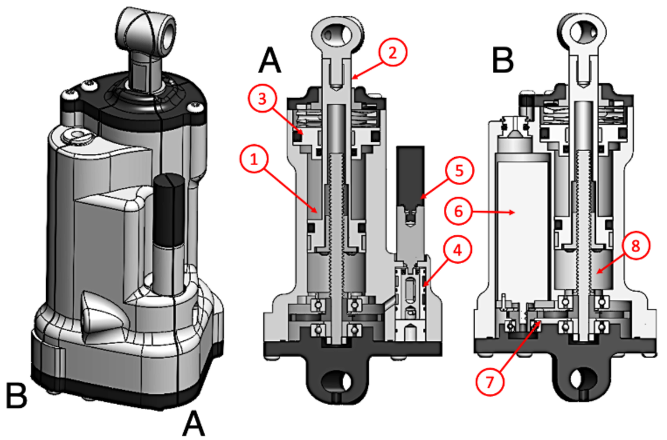

Fig. 2. Solid model of hybrid actuator implementation. Section A shows the passive hydraulic unit, indicating: (1) the piston, (2) piston rod, (3) accumulator, (4) servovalve, and (5) servovalve motor. A check valve in the knee extension flow path is not visible, since it is out of plane with the valve and drive systems. Section B shows the active drive unit, indicating: (6) the drive motor, (7) gearset, and (8) leadscrew, which drives the piston rod (2).

Therefore, one of the most salient deficiencies in the functionality of energetically passive prostheses is the inability of such devices to provide proper stair ascent swing phase motion. More specifically, a swing-phase motion that will enable a user to easily and naturally clear the next stair when performing step-over stair ascent, and to a lesser extent, when performing a step-to stair ascent, without an undue amount of hip circumduction, or other compensatory movement. Note that compensatory movements, aside from appearing asymmetric, have been shown to be related to a number of other health issues [6].

In order to improve functionality during stair ascent (and across other activities of locomotion), a number of researchers have developed various powered prosthesis prototypes capable of providing knee extension assistance during stance, and also appropriate knee motion during swing (e.g., [7]–[16]). Such devices have demonstrated efficacy in providing step-over stair ascent and descent functionality [17], [18]. A study of one such powered knee and ankle prosthesis demonstrated a decreased metabolic cost of transport during step-over stair ascent with the powered prosthesis, relative to passive daily-use devices [19], presumably due to the ability of the powered prosthesis to provide powered extension during stance, offloading the hip power required by the user.

The knee prosthesis described in this paper takes an alternative approach to addressing deficiencies in stair ascent. Rather than employ a high-power, high-torque drive system, capable of providing fully powered knee extension during stance, this device supplements the functionality of a passive stance-controlled MPK with a small motor that offers non-inertial control of knee motion during swing phase (i.e., a stance controlled, swing assist or SCSA device). Since the prosthesis employs a low-torque drive system, it maintains the essential behavior of an MPK, including the ability to provide the wide dynamic range of output impedance necessary to passively provide appropriate stance and ballistic swing functionality [20]. Compared to a fully powered prosthesis in stair ascent, the approach presumably sacrifices some portion of the prospective metabolic benefit resulting from fully powered stance knee extension. In exchange, however, the user retains a greater degree of agency in movement, while still enabling stance knee extension in step-to and step-over stair ascent (albeit via hip effort).

This paper examines the functionality provided by an SCSA knee device when performing both step-to and step-over stair ambulation. The primary supplemental behavior, relative to a standard MPK, is the ability of the device to provide a non-inertial swing-phase motion to help clear stairs during swing phase of step-to and step-over stair ascent, which presumably decreases the need to perform compensatory actions. This paper describes the controllers that enable both stair ascent and descent and describes the implementation of these controllers in a small study involving three knee prosthesis users.

II. Semi-Powered Prosthesis Prototype and Control

A. Prosthesis Design

The prosthesis prototype used in this study employs a hydraulic dissipator with resistance modulated by a motor-controlled valve to provide nominal MPK behaviors, in combination with a low-torque, highly backdrivable actuator system to provide supplemental power, primarily during swing phase. A solid model of the hybrid active/passive actuation unit is shown in Fig. 2, where section A shows (most of) the modulated dissipator subsystem, and where section B shows the active drive system. The hydraulic system employs a spring-loaded accumulator, which acts to accommodate variable fluid volume, and also provides a light extension aid to facilitate knee extension. The hydraulic system also includes a check valve in parallel with the servo valve, such that the variable resistance imposed by the controllable valve acts unidirectionally against knee flexion, while hydraulic resistance against knee extension remains low. The resistance can be bidirectionally supplemented by the drive motor if desired – either actively by driving torque via the batteries, or passively by shorting the motor's leads together for some portion of the PWM period.

In addition to the drive unit shown in Fig. 2, the prosthesis prototype includes an absolute encoder for knee angle measurement; a 6-axis inertial measurement unit used to estimate the movement of the shank; and a load cell that measures axial load in the shank. Finally, the prototype includes an on-board embedded system and battery. The design, design rationale, functional characteristics, and nominal walking controller of the semi-powered knee prototype used in this paper were described in a recent paper [20].

B. Stair Ascent Controller

The finite state machine for stair ascent (Fig 3, A) is comprised of three states: stance, swing flexion, and swing extension. The same state machine was used for both step-to and step-over gait, although with differing switching thresholds and commanded output angles. A summary of switching conditions is included in Table I.

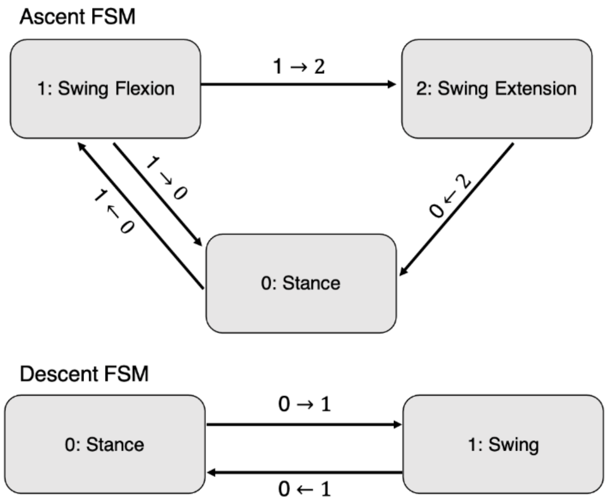

Fig. 3. FSMs for stair ascent and stair descent.

TABLE I
STATE TRANSITION CONDITIONS FOR STAIR ASCENT CONTROLLER

Transition	Condition
$0 \rightarrow 1$	Thigh Angle < Threshold && Load < Threshold
$1 \rightarrow 2$	Thigh Velocity < Threshold && Thigh Angle < Threshold
$2 \rightarrow 0$	Load > Threshold || Knee Angle Trajectory == complete
$1 \rightarrow 0$	Load > Threshold

During stance phase of stair ascent (State 0), the variable resistance valve is moved to a fully closed position, such that the knee is locked unidirectionally against flexion; however, due to the check valve in the hydraulic system, the knee remains unimpeded in extension. This unidirectional behavior allows the user to load and extend the knee as they progress from the weight acceptance phase through the forward continuance phase of stair ascent without fear of having the device buckle when loaded.

Assuming orientation guard conditions are met, when the knee is loaded and still flexed beyond 30 deg (i.e., from the weight acceptance phase to the beginning of the pull up phase), light extension assistance of approximately 7.5 Nm is provided via the drive system. While this is a fraction of the peak torque required to lift the user's body weight, the supplement torque helps to initiate extension movement, essentially acting like a strong extension aid. When the knee is extended to within 15 deg of full extension (i.e., the end of the pull up phase through the beginning of forward continuance), the motor is engaged as a passive damper in order to smoothly approach the hard stop. The valve remains fully locked against flexion throughout this stage, until the conditions to transition into swing flexion are met.

Once the user has moved into an appropriate orientation and the leg is unloaded, the device transitions into swing flexion (State 1). Upon transitioning, the resistance valve (servovalve in Fig. 2) is opened, and the knee angle and thigh orientation with respect to gravity are recorded. The knee angle in the swing flexion state is controlled to be algebraically related to the thigh angle (relative to the gravity vector), as measured by the IMU and knee angle sensor, according to:

$$\theta_{knee} = K\left(\theta_{thigh} - \theta_{0_{thigh}}\right) + \theta_{0_{knee}} + C \qquad (1)$$

where θ_{thigh} is the thigh angle relative to the gravity vector; θ_{knee} is the knee angle; $\theta_{0_{thigh}}$ and $\theta_{0_{knee}}$ are the thigh and knee angles at the instant the controller switches into the swing-flexion state; K is a gain between the commanded knee and thigh angles (1.75 in the implementation here); and C is a constant that provides a desired knee motion relative to thigh motion for stair ascent, set here to 25 deg for early flexion or step-to gait, and 35 deg for step-over gait. While slightly unintuitive, this control methodology aims to essentially control the orientation of the shank (knee angle less thigh angle) to maintain some offset from its initial position with respect to gravity. The essence of the control is that the knee movement is commanded by the thigh movement, in a manner that fairly naturally generates a desired stair ascent leg kinematics, and naturally accommodates varying cadence. This angle tracking is performed until the switching conditions are met which detect either the user's intent to extend the leg in a step-to fashion or that the user has completed the flexion phase of a step-over motion and is ready for the leg to extend prior to foot placement.

Upon meeting the switching conditions, the controller transitions into swing extension (State 2). The knee position control gains are decreased, and the resistance valve is shut to resist any further flexion. The knee angle is extended, either to full extension for step-to gait, or to a flexed position appropriate for foot placement in step-over gait, where the angle is tuned to be a nominal comfortable angle for a given individual, depending primarily as a function of user height. When the user loads the leg, or the knee trajectory has been completed, assuming other guard conditions are met, the device then transitions back into State 0.

C. Stair Descent Controller

The stair descent controller is divided into two states: stance and swing. The same controller is used for both step-to and step-over gait, though during step-to descent the necessary triggers may not be met to enter a true swing phase, due to absence in required knee flexion. Note that failure to enter swing state in step-to descent is not a concern, since step-to descent does not entail knee motion during swing (i.e., a knee prosthesis remains in constant extension in step-to descent). Switching triggers for the finite state machine are shown in Table II.

Upon entry to the stance phase of stair descent (State 0), the resistance valve is commanded to a medium-high damping configuration. Once the knee is adequately flexed and unloaded, the device transitions into the swing state (State 1). Upon entering swing, the valve motor is rotated into a low damping state, while the drive system follows a simple knee angle trajectory, initially flexing the knee slightly in order to help clear the stair, before extending in preparation for the next step. Once the knee trajectory reaches its maximum flexion, just before extension begins, the valve is rotated back into the medium-high flexion damping position in preparation for

TABLE II
STATE TRANSITION CONDITIONS FOR STAIR DESCENT CONTROLLER

Transition	Condition
$0 \rightarrow 1$	Knee Angle > Threshold && Load < Threshold
$1 \rightarrow 0$	Load > Threshold || Knee Angle Trajectory == complete

the beginning of stance. As the knee nears completion of its extension (once it reaches a flexion angle below 5 degrees), the passive motor damping is reintroduced to slow the leg and mitigate terminal impact. When the knee reaches full extension, or is loaded, it transitions back into state 0.

D. End of Extension Behavior

Although not a separate state, in both the ascent and descent tasks, the prosthesis includes a supplemental control behavior to prevent terminal impact near the end of each swing phase and at the end of the user's stance knee extension during ascent. Here, the motor is employed in a strictly passive, dissipative mode in order to slow the extension of the knee smoothly. To achieve this, the motor controller employs PWM switching to short the motor leads together via the low side MOSFETs of the H-bridge, thus modulating passive current buildup in the motor windings, which acts as a controllable damper to slow extension and avoid terminal impact.

E. Controller Setup and Tuning

Controller parameters, gains, and switching conditions were tuned a priori on a healthy subject using an able-bodied adaptor, and subsequently on one individual with transfemoral amputation, until the control system provided comfortable stair ascent and descent functionality. All control gains, parameters, and switching conditions subsequently remained unchanged for the two subsequent experimental participants, with the exception of the following two controller parameters, which were tuned to each participant: 1) the final extension knee angle for step-over stair ascent; and 2) the knee damping constant for step-over stair descent. As mentioned previously, the former has a dependence on the height of the participant, since a shorter participant is better accommodated with more knee flexion for a given stair height. The latter (stair descent damping constant) is a control setting similarly tuned for passive prostheses. These gains were tuned during training, while subjects ascended and descended stairs, based on participant comfort. On average, tuning these two participant-specific control parameters required approximately 10-15 min per participant.

III. EXPERIMENTAL ASSESSMENT

In order to test the efficacy of the approach, a study with three participants with unilateral, transfemoral amputation was performed. In this study, each participant ascended and subsequently descended an 8-step flight of stairs while donning 40 lower body motion capture markers, as shown in Fig. 4. Each step was 6.5 in (0.17 m) high. Lower limb kinematics were recorded via an infrared motion capture system (Vicon, Oxford, GBR). Each person completed this task 5 times

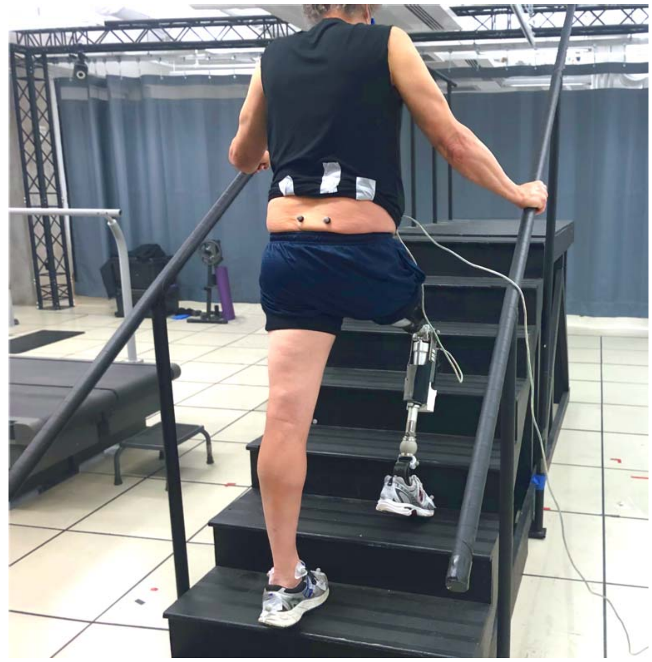

Fig. 4. Study participant wearing motion capture markers and ascending experimental stair set up.

performing both ascent and descent in a step-to fashion, and then repeated the same number of trials using a step-over gait. These activities were performed first on their daily use device, and subsequently on the Vanderbilt SCSA knee. All participants were allowed to rest as desired and were given a minimum 5-minute break between trial sets. Prior to recording data, each participant was allowed approximately 30 min of training time using the SCSA device on stairs, which encompassed the controller parameter tuning previously mentioned. This study was approved by the Vanderbilt IRB. A video is included in the supplemental material that shows representative motion capture playback corresponding to each set of data presented here, and also shows representative video of both step-to and step-over ascent with the SCSA and daily-use prostheses.

The study participants, two male and one female, ranged in age from 28 to 64. Two of the three used an Ottobock C-leg (hydraulic MPK) as their daily use knee (Participants 1 and 3), while Participant 2 used an Ottobock 3R80 (rotary hydraulic non-MPK). Note that the experimental SCSA prosthesis was fit to each subject's daily-use socket in a manner that would not alter the alignment of their daily-use prosthesis. For two of the three subjects, doing so resulted in greater stance-knee hyperextension in the SCSA prosthesis than ideal; this was determined to most likely disadvantage the experimental prototype, and so was deemed experimentally acceptable.

The finite state machine control architecture was implemented on a laptop using MATLAB Simulink Realtime. A CAN message containing the commanded knee position, resistance valve position, and when applicable, passive motor damping was sent from the computer to the prosthesis. Once received by the device, closed loop position and current control was performed on the device's embedded system. Controller parameters for the SCSA knee were tuned to each participant to accommodate variations in weight, height, and preference.

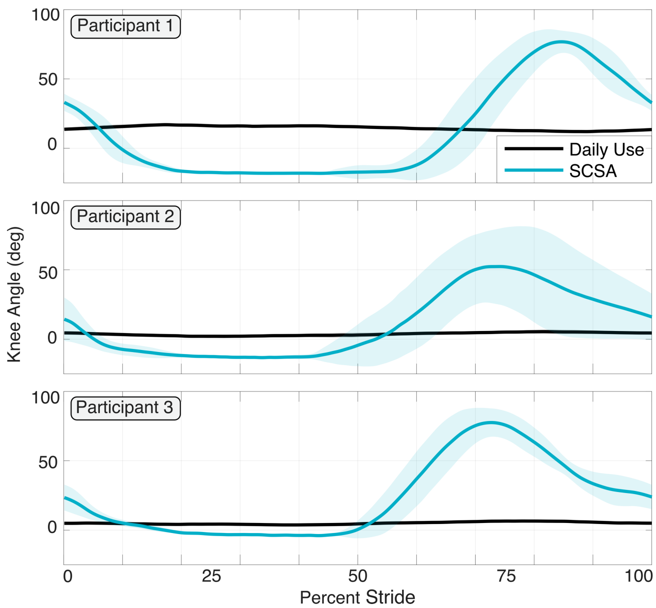

Fig. 5. Knee angle versus stride for step-to stair ascent.

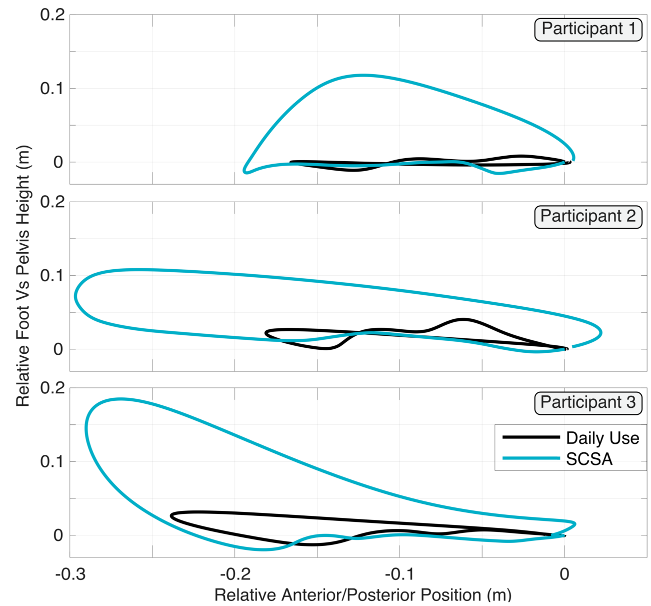

Fig. 6. Affected side foot path, relative to pelvic center when performing step-to stair ascent.

For all presented trials, volitional transitions between tasks (i.e., automatically changing from the walking state machine to the stair ascent state machine) were disabled. These transitions were instead manually controlled by the authors via the computer, as transitions are not directly associated with the objective of the movement analysis presented here.

IV. Results

Figures 5 through 10 show comparisons of step-to and step-over stair ascent with each participant's daily-use prosthesis and the SCSA prosthesis prototype. Note that data is presented in the figures for each individual subject (rather than in aggregate), since each of the three study participants exhibited somewhat different ascent kinematics and corresponding compensatory movements. Aggregate results are presented throughout the text to characterize the generalized relative behaviors provided by the SCSA and daily-use knees.

All data shown were parsed from foot strike to foot strike before being normalized to percent stride, where each foot strike was identified via the kinematic data. All affected-side data was parsed relative to the affected-side foot strike, and all contralateral data to that of the contralateral foot. For all trials, the first and final strides were ignored, leaving approximately 20 strides for each participant for each experimental case (i.e., for each prosthesis, and for each of the step-to and step-over ascent and descent trials).

For the data shown, plots in black are of the daily-use device, while blue (lighter grey, if viewing in greyscale) are of the SCSA prototype. When present, shaded areas indicate the interquartile range of the data, while the bolded line is the mean of the interquartile range. The interquartile range and its mean were used instead of standard deviation and mean, respectively, as phase discrepancies with respect to foot strike, particularly when using the daily-use device, resulted in non-normally distributed data as a function of percent stride. For tasks where it was available, healthy datasets from [21] are included as reference in yellow.

An analysis of symmetry between the affected and contralateral side was conducted for step-over ascent, and also to assess similarity to healthy movement. This analysis was not conducted for step-to data, as step-to gaits are inherently asymmetric. The symmetry and similarity analyses employed calculation of two coefficients: correlation coefficient (CC) to indicate similarity of movement, and percent difference (PD) to indicate similarity in magnitude of movement. The correlation coefficient (CC) between two sets of data was constructed as in [22]. For this analysis, the datasets were unaltered in time, to capture discrepancies in phase. The primary shortcoming of the CC is that it is necessarily magnitude independent. Therefore, it can indicate the extent to which two curves match in shape, but not similarity of amplitude. For this reason, a second metric – the percent difference (PD) in range between the contralateral and affected side –was also used—where percent difference is defined as

$$PD = \frac{Range_{Affected} - Range_{Contra}}{Range_{Contra}}. \qquad (2)$$

A. Step-to Stair Ascent

Fig. 5 shows the knee angle as a function of stride for step-to stair ascent for the respective daily-use prostheses and for the SCSA knee prototype. When using the SCSA knee, the study participants achieved an average maximum knee flexion of 68.9 deg with the SCSA knee, relative to 9.6 deg with their respective daily-use prostheses.

The effect of the knee motion on foot motion provided by the SCSA knee is illustrated in Fig. 6, which shows the movement of the foot, relative to the pelvis center, for both the SCSA knee and the daily-use prostheses. When using the daily-use prosthesis, the prosthesis provided a net height

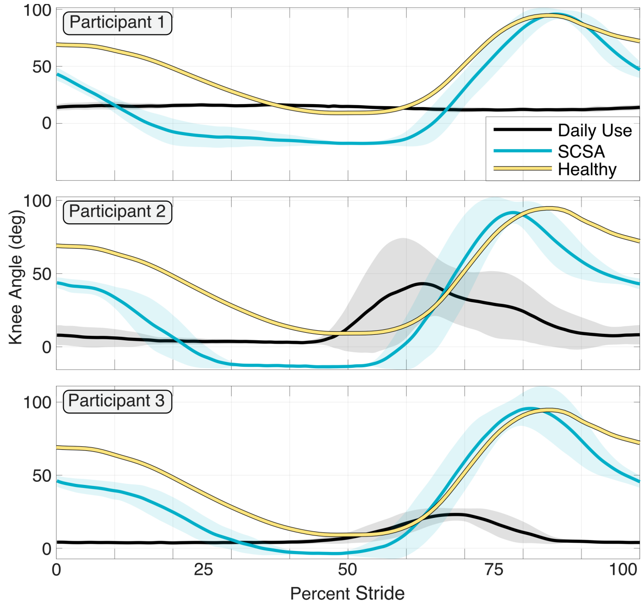

Fig. 7. Knee angle versus stride for step-over stair ascent.

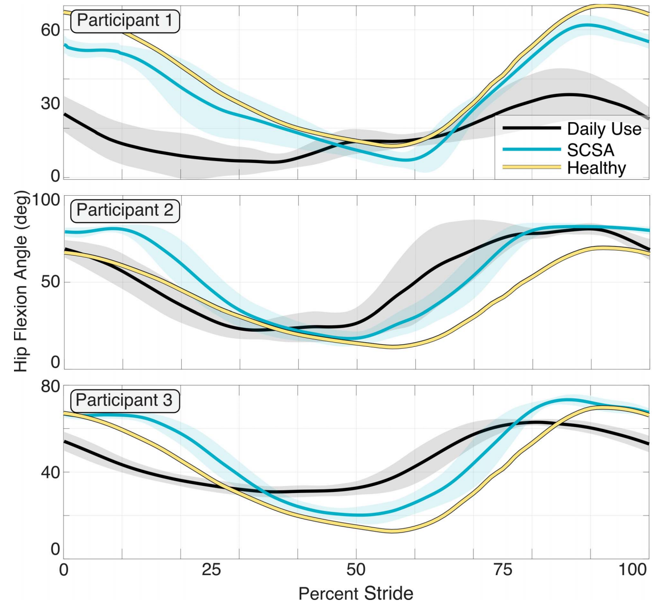

Fig. 8. Hip angle versus stride during step-over stair ascent.

change (relative to the pelvis) across subjects of 2.6 cm, while the SCSA prosthesis provided a net height change of approximately 13.6 cm (where net height change is the maximum height, as compared to the height at foot strike). Recall for reference that the stair height was 17 cm. The difference between the displacements in Fig. 6 and the stair height is due primarily to unaffected ankle plantarflexion and bilateral hip abduction, as evidenced in the accompanying video.

B. Step-Over Stair Ascent

Fig. 7 shows the knee angle as a function of stride for step-over stair ascent for the respective daily-use prostheses, for the SCSA knee prototype, and for healthy subjects. As in step-to ascent, significantly more knee flexion was observed when using the SCSA device (average peak knee flexion of 94.4 deg) when compared to the users' daily use prostheses (average peak knee flexion of 27.5 deg). Additionally, the knee trajectory when using the SCSA device was much more closely correlated to healthy knee trajectory (correlation coefficient of 0.93 on the SCSA, as compared to −0.23 on daily use) and had a total knee angle range much more closely resembling that of the data recorded from healthy subjects (SCSA having 126% of healthy range, daily use having 27.3%).

Fig. 8 shows the hip angle as a function of stride for step-over stair ascent. Users' hip motion was more representative of healthy hip movement when using the SCSA knee (correlation coefficient of 0.97, range percent difference of +1.4%) relative to their daily use devices (correlation coefficient of 0.62, range percent difference of −28.4%). Additionally, as shown in Fig. 9, which displays the vertical foot motion as a function of stride for both prostheses, and for both the sound and affected sides, the foot achieved a greater vertical clearance during swing with the SCSA knee, and demonstrated a more symmetric movement, relative to the daily-use device.

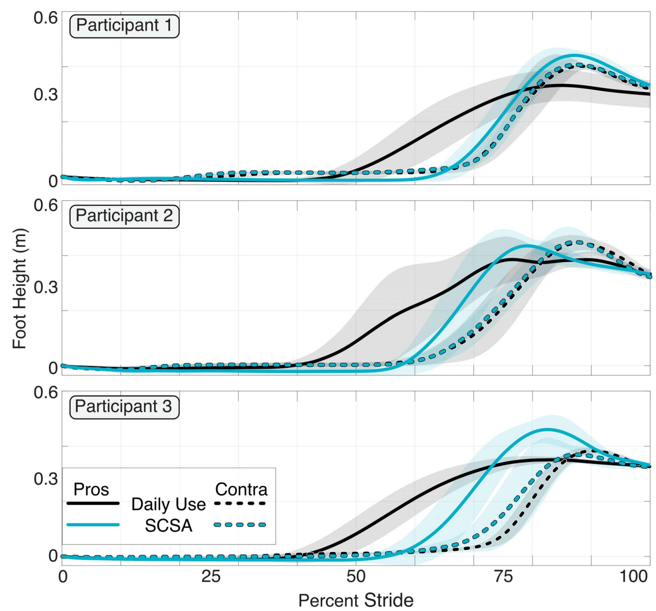

Fig. 9. Participant foot (center) height versus stride. For both the SCSA and the daily use device, the affected side is shown in solid line, while the contralateral side is shown dashed. The interquartile range for each is shaded.

Figure 10 provides a qualitative representation of each participant's lower-body kinematic configuration (as measured by the motion capture system) at randomly selected (but also representative) prosthetic foot strikes. The difference in sagittal-plane kinematic configuration is clear, most notably the increased knee flexion with the SCSA device. Frontal plane differences are most notable at foot strike (i.e., in the figure) in Participant 3, although 2 of the 3 participants substantially reduced circumduction with the SCSA knee (i.e., Participants 1 and 3 exhibited a 25% and 33% reduction in maximum frontal-plane foot path width). Participant 2 increased frontal-plane foot path width slightly with the SCSA knee. On average, there was an 18% reduction in frontal-plane

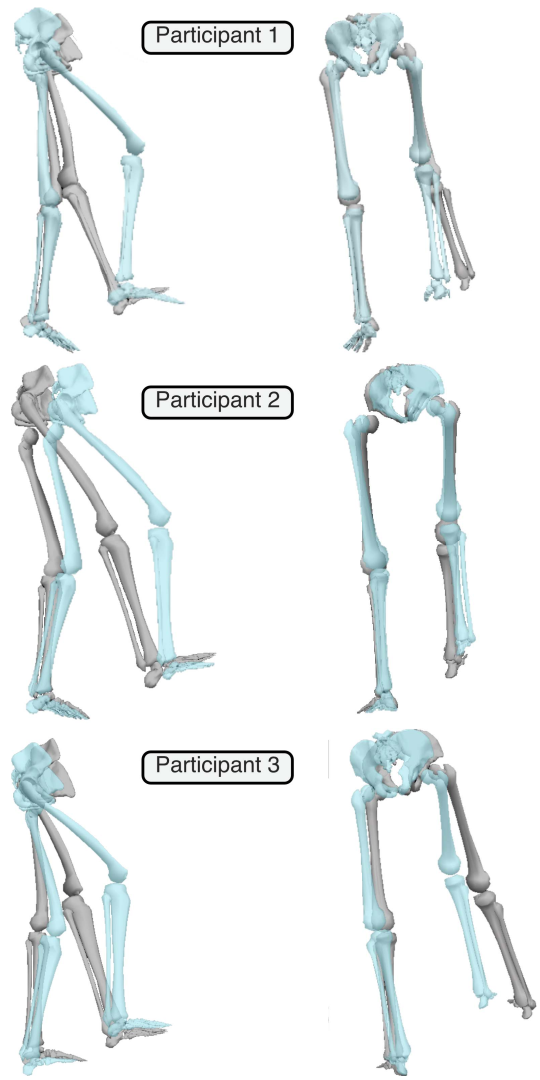

Fig. 10. Kinematic orientation of the lower body for each of the three subjects at randomly selected foot strikes. SCSA orientation is shown in blue (lighter grey, if viewed in greyscale), while daily use is shown in dark grey.

foot path width when using the SCSA knee, relative to the daily-use knees.

C. Stair Descent

The participants also performed stair descent in both step-to and step-over fashion for the SCSA knee and daily use devices. During step-to stair descent, the knee joint is largely static, and thus no data from such trials is shown here. The knee joint kinematics for step-over stair descent are shown in Fig. 11. The behavior of the SCSA device in descent is (by design) patterned after energetically-passive MPK stair descent behavior, and as such ambulation was very similar. For the participant who used the 3R80 knee (Participant 2), stair descent behavior was slightly different from the SCSA knee trials, with the SCSA having less inter-stride variability, slightly less flexion, and achieving full extension more consistently at foot strike. Broadly speaking, stair descent is a task well performed by state-of-the-art MPK devices. As such, the primary goal in the collection of this data was to ensure that the SCSA device and control matched the behavior of an MPK. Small differences in flexion and extension rate between the SCSA and daily use devices in participants 1 and 3 are a direct result of tuning damping and extension gains to preference. This could be modified to better match the kinematics of their daily use devices, but the authors opted to tune to user preference.

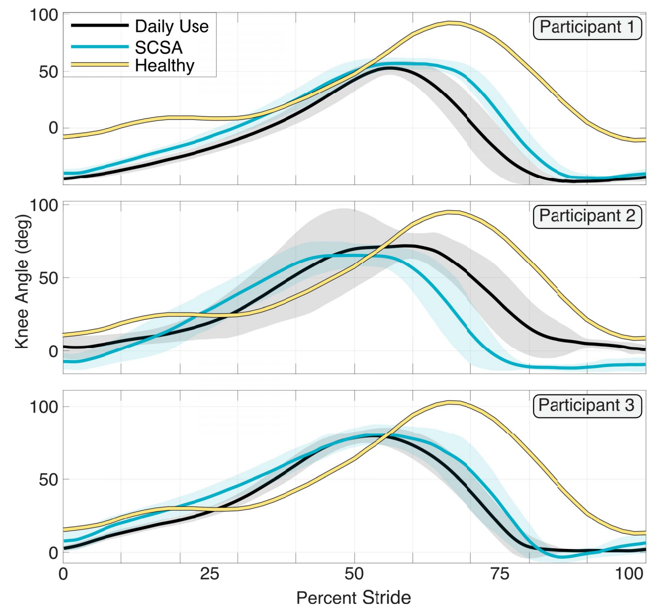

Fig. 11. Knee angle versus stride for step-over stair descent.

D. Symmetry and Similarity Analyses

As previously stated, the CC and PD were used for analysis of symmetry between the affected and contralateral sides, and also for similarity between the affected-side movement and healthy data. These two metrics were calculated for knee trajectory, hip trajectory, and foot height with respect to pelvis for symmetry. Foot height was analyzed with respect to the pelvic center to isolate the effect of the prosthesis from compensatory movements, some of which are entrained in the user. Note that foot height was not used in the comparisons to healthy movement, since that data for healthy subjects was not available. The resulting values for each metric for each component of data are reported in Table III. All data shown in Table III are calculated based on the mean of the interquartile range for each percent stride.

As the data show, in aggregate, both symmetry between sides and similarity to healthy movement was improved substantially in correlation and range for all metrics. Note that (although not shown in the table), each participant improved in all metrics.

V. Discussion

A. Prospective Value of Swing-Assist in Stair Locomotion

Energetically passive knee prostheses are characterized by a low output impedance at the knee joint, particularly in swing phase. This low impedance makes these prostheses highly receptive to physical input from the user, and as a result enables the user to retain substantial agency of movement, and

TABLE III
SYMMETRY AND SIMILARITY TO HEALTHY DATA

	Device	Affected/Contralateral CC	Affected/Contralateral PD	Affected/Healthy CC	Affected/Healthy PD
Knee Angle	Daily Use	-0.34	-84.8	-0.23	-72.7
	SCSA	0.90	7.3	0.93	26
Hip Angle	Daily Use	0.54	-18.1	0.62	-28.4
	SCSA	0.93	-8.2	0.97	1.4
Foot Z wrt Pelvis	Daily Use	0.67	-60.2	Healthy Data Unavailable	
	SCSA	0.95	-19.3	Healthy Data Unavailable	

also facilitates movement coordination between the user and prosthesis. Passive knee prostheses, however, rely on inertial coupling between the thigh and shank to provide appropriate swing-phase motion, which works well for most walking activities, but generally does not provide appropriate swing phase knee motion for stair ascent. The swing-assist approach assessed here employs a modulated hydraulic dissipator, similar to a standard passive knee, in parallel with a small motor with a low-torque drive system that can actively supplement movement. The relatively low-power motor and drive system facilitate implementation of a compact, lightweight, quiet, and low-impedance device, relative to a fully-powered device capable of stance-phase torques. The approach intends to maintain passive inertially-coupled motion during walking, but enable non-inertial swing movement when appropriate, such as during stair ascent.

As such, the objective of this paper was to assess the prospective value added of this approach in stair locomotion, relative to an energetically passive knee (i.e., relative to the current standard of care). For step-to ascent, which is probably the most used form of ascent among knee prosthesis users, the swing-assist approach offers active knee flexion (Fig. 5), which results in substantially increased raising of the foot relative to the pelvis (on average approximately 14 cm rather than 3 cm, Fig. 6), thus reducing the need for compensatory movement needed to clear the stair, relative to energetically passive knees. For step-over ascent, the flexion provided by the swing-assist knee (Fig. 7) results in affected-side hip and knee motion more representative of healthy motion (Figs. 7 and 8), more symmetrical gait (Fig. 9 and Table III), and decreased compensatory movement (Fig. 10). Note that step-to gait was not compared to healthy, nor assessed for symmetry, since it is neither a healthy norm, nor a symmetric gait.

Figure 11 confirms that swing-assist knee neither improves nor diminishes the stair descent functionality of a hydraulic passive knee, which is generally dissipative in stance and swing, and therefore is unlikely to benefit from assistive power.

B. Swing-Assist Relative to Fully Powered Knees

The swing-assist approach presented here does not provide significant stance-knee extension torque or power, relative to those exhibited by the healthy knee during stair ascent. This stance knee assistance presumably facilitates step-over stair ascent for prosthetic knee users, and has been shown to decrease the metabolic cost of step-over stair ascent relative to passive knees [19]. As evidenced in this paper, however, prosthetic knee users do not require powered knee extension to perform effective step-over stair ascent (i.e., as asserted in the introduction, the hip and knee effectively generate torque in parallel in the stance leg). Therefore, the relative merit of a fully powered versus a swing-assist approach, for purposes of stair ascent, is primarily a trade-off between the relative value of an inertially-coupled swing phase (during walking activities), relative to added stance-knee extension assistance in stair ascent, for a given individual. In addition to enabling inertially-coupled swing during walking, a swing-assist approach is also likely to reduce the size, weight, and/or audible noise as well, relative to a fully powered approach.

C. Limitations of This Assessment

Among the most salient limitations of this work, experiments were conducted on three participants, who between them used two types of daily-use knee prosthesis, which is a relatively small sample size. More subjects and greater variety of daily-use knee prostheses would provide greater confidence in the outcomes reported here. Despite the small sample size, stair ascent and descent kinematics, like kinematics of human movement during most locomotion activities, have been shown to be reasonably uniform between people (e.g., [4], [5]). As such, although more subjects would increase confidence in the data, one would not expect inclusion of more subjects to substantially change in the trends reported here.

VI. CONCLUSION

The authors describe a stance-controlled swing-assist (SCSA) knee prosthesis based on supplementing a low-impedance energetically-passive knee with a (relatively) low-torque motor and drive system, which allows the knee to maintain a strictly inertially-coupled swing phase motion during walking, while enabling a non-inertially-coupled motion during stair ascent.

This paper describes a small study with three participants with transfemoral amputation intended to examine the prospective benefits for stair locomotion of the SCSA knee, relative to passive knee prostheses. The study considered step-to stair ascent; step-over stair ascent; and step-over stair descent. For step-to stair ascent, the SCSA knee was shown to raise the foot relative to the pelvis a substantially greater amount (relative to energetically passive knees), thus decreasing the need for compensatory movement. For step-over ascent, the SCSA knee resulted in affected-side hip and knee motion more representative of healthy motion, more symmetrical gait, and decreased compensatory movement. Finally, the SCSA knee was shown to provide stair descent movement similar to a commercially-available hydraulic passive knees.

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