

Swing-Assist for Enhancing Stair Ambulation in a Primarily-Passive Knee Prosthesis

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Abstract—This paper presents the design and implementation of a controller for stair ascent and descent in a primarily-passive stance-controlled swing-assist (SCSA) prosthesis. The prosthesis and controller enable users to perform both step-over and step-to stair ascent and descent. The efficacy of the controller and SCSA prosthesis prototype in providing improved stair ambulation was tested on a unilateral transfemoral amputee in experiments that employed motion capture apparatus to compare joint kinematics with the SCSA prosthesis, relative to performing the same activity with a microprocessor-controlled daily-use passive prosthesis. Results suggest that the SCSA knee significantly decreases compensatory motion during stair activity when compared to the passive prosthesis.

I. INTRODUCTION

Many tasks of daily living are often more difficult for persons with transfemoral amputation than for their age-matched non-amputee counterparts [1, 2]. One task frequently performed either in the home or in daily life which has been shown to be exceptionally difficult for persons with amputation is traversing stairs [3].

Perhaps the most common type of knee prosthesis which provides stair ambulation capability is a microprocessor-controlled knee (MPK). MPKs incorporate stance control via modulated dampers that dissipate energy and thus provide controlled stance-knee yielding during stair descent. Such devices are capable of providing good step-to or step-over stair descent functionality.

Although MPKs can provide adequate stair descent functionality, they are more limited in their ability to provide stair ascent functionality, due to two essential limitations: lack of active knee extension during stance phase, and lack of swing control during swing phase. The first assists during the stance phase of stair ascent in which the body center-of-mass is lifted, while the second guides the swing foot over the subsequent stair riser and onto the tread during the swing phase. The first of these limitations is less fundamental than the second. Specifically, in stance phase, an amputee can provide stance knee extension via hip extension. In swing phase, however, an amputee cannot directly provide a desired knee motion via hip effort. Specifically, a passive transfemoral prosthesis (e.g., an MPK) can be thought of as a double link pendulum with an actuated revolute joint at the hip and unactuated revolute joint at the knee. As shown in Fig.1, when in stance phase, the foot of the prosthesis is grounded, and as such the leg is fully controllable by the hip (i.e., the actuated joints are greater than or equal to the degrees of freedom). During the swing phase of gait, when the foot is

in the air, the leg system is under-actuated [4], and therefore the hip cannot directly control the knee joint. Thus, swing is the only phase of normal gait in which the device-user combination is an underactuated system.

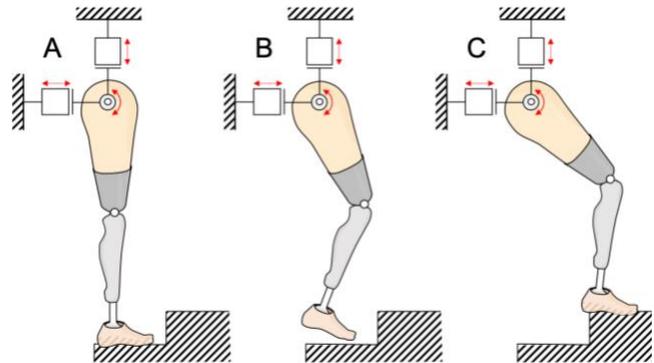


Fig. 1. The leg system is fully actuatable in states A and C when the foot is on the ground, meaning that the user can fully dictate knee motion via hip actuation. However, during the swing phase of stair ascent (B) the leg has too many degrees of freedom and is thus an underactuated system. During this time, the user may have difficulty achieving adequate flexion

The swing phase of healthy level ground walking gait has been described as ballistic motion, meaning that healthy knee trajectory is driven primarily through the hip motion and can be achieved without powered actuation of the knee joint [5, 6]. In a passive device such as an MPK, the combination of shank inertia, joint resistance, and thigh angular trajectory, allows the thigh and shank to be inertially coupled – meaning that the desired swing-phase knee trajectory can be achieved via natural swing-phase movement of the hip. However, during activities which involve slower movement, or movement in which the relative movement between hip and knee is not achieved by the natural dynamics of a double link pendulum, this “ballistic” inertial coupling is not achieved, and thus the hip may not provide a desired knee movement. One such activity is stair ascent. Whether performing a step-to, or a step-over stair ascent, the natural movement of the hip joint during stair ascent, when acting through the passive double-pendulum dynamics of the leg, does not result in a knee motion suitable for stair ascent. As such, the user is forced to clear the subsequent stair by employing exaggerated hip circumduction, typically with a fully, or near fully extended knee. Doing so also requires a far wider stance, relative to non-amputee stair ascent. One potential alternative to this approach is that taken by the Ottobock Genium knee, in which the user can achieve improved knee flexion during

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stair ascent by “whipping” the thigh back, immediately before quickly extending it forward. This approach provides sufficient knee movement but requires visibly unnatural hip movement.

As an alternative to MPKs, some powered prostheses have been described which can provide step-to and step-over stair ascent without requiring awkward compensatory movement [7–12]. In addition to providing swing knee flexion, these devices produce knee extension torque during the stance phase of stair ascent. However, they require a large motor and transmission ratio to do so. The combination of large motor and transmission generates increased audible noise and high output impedance relative to MPKs. The high output impedance, in turn, reduces the user’s ability to initiate gait activities, and also reduces the natural coordination between user and device.

In order to enable improved stair ascent movement, while maintaining the low output impedance and audible noise of an MPK prosthesis, the authors propose here to employ a knee prosthesis similar to a stance-controlled MPK, but with a low-torque swing-assist feature. Since the swing-assist feature is low torque, the swing-assist motor does not require a high transmission ratio, and therefore is characterized by a low audible noise and output impedance. The resulting prosthesis is called a stance-controlled swing-assist (SCSA) prosthesis. With this approach, the prosthesis provides stair descent in the same manner as a standard MPK, but additionally enables both step-to and step-over stair ascent with improved kinematics and less compensatory movement, relative to a standard MPK. This paper describes briefly the device design; describes a stair controller; and presents experimental results indicating improved stair ambulation relative to a standard MPK.

II. DEVICE DESCRIPTION

The device used during this study, known as an SCSA prosthesis [13] aims to improve upon the proven benefits of MPKs while maintaining their low audible noise and output impedance. It does so by coupling a small motor and transmission system to the knee joint in parallel with a modulated damper. The low-torque motor is used during swing-phase to add controllability to this otherwise underactuated phase of motion. Because the motor system is designed for smaller output torques, the system is easily backdriven, and the user remains fundamentally in control of initiating device movement, as with an MPK.

The coupled modulated damper and active motor drive are shown in schematic in Fig. 2 and shown in solid model implementation in Fig. 3. The composite linear actuator is employed in the knee prosthesis in a typical slider-crank configuration. The modulated damper component is based on a hydraulic damper, where the level of damping is modulated via a pressure-balanced rotary spool valve, which is actuated by a 12 mm diameter brushless gearmotor (Faulhaber 1226 with 64:1 gearhead). Since the high damping required for stance phase are exclusively used to resist flexion, a check valve is employed in parallel with the rotary spool valve so

that damping resistance to knee extension always remains low, even in cases that resistance to knee flexion is high. The main hydraulic cylinder is a single-rod-type, which includes a spring-loaded accumulator to account for rod volume. The accumulator spring is a wave spring with a spring constant of 300 N/cm (166 lb/in), which at full flexion imposes a force of 160 N (35 lbs) on the accumulator piston, applying approximately 0.5 Nm of extension bias torque at the knee. The hydraulic cylinder employs a one-inch diameter bore, which results in an estimated maximum pressure of 11.4 MPa (1660 psi) for an individual weighing 100 kg in stair descent, using healthy data [14] as an approximation of resistive torque. In order to minimize friction, the cylinder employs wear rings and cup seals along sliding surfaces, and O-rings or face seals at static interfaces.

In order to provide swing assist, a 90 W brushless motor was selected (Maxon EC22 90 W). The motor is rated at a maximum continuous torque of 50 mNm, and with proper heat-sinking, can be reasonably operated at a lower duty cycle to 150 mNm. As shown in Figs. 2 and 3, the motor is coupled to the hydraulic rod through a low-ratio and low-torque transmission, which consists of a lead screw located within the piston rod, driven by the motor through a pair of spur gears.

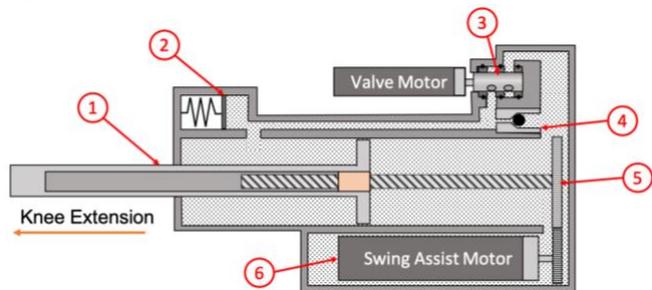


Fig. 2. Schematic of hybrid actuator, including: (1) a single rod cylinder with lead screw contained within rod; (2) spring-loaded accumulator biased toward extension; (3) rotary spool valve; (4) check valve allows for low-resistance extension; (5) lead screw

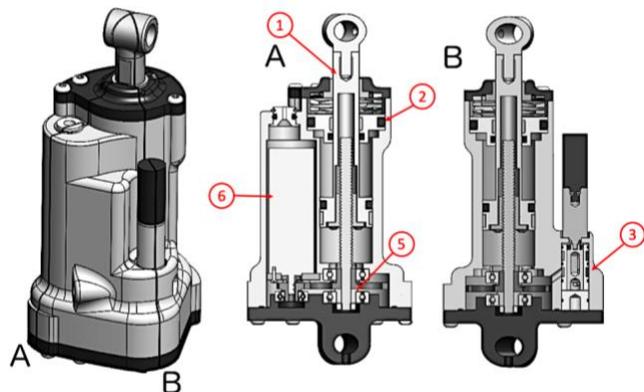


Fig. 3. Solid model of hybrid actuator implementation. Pointers correspond to items in Fig. 2. The check valve is not visible, since it is out of plane with the valve and drive systems.

The rotary-to-linear transmission ratio is approximately 3050 rad/m, which, in combination with the previously described slider-crank configuration, provides an average rotational transmission ratio of approximately 75:1 between the electric motor and knee. Given this transmission, the motor will provide approximately 9 Nm of knee torque (under ideal circumstances). Alternatively, assuming an 85% transmission ratio efficiency, the motor will provide 7.5 Nm of active torque and approximately 75 W of active power at the knee. Note that in the same way as the actuator designed in [15], the entire system is submerged in oil, which is intended to reduce friction in the gears and lead screw and therefore minimize backdrive friction (i.e., minimize output impedance). As indicated in Figs. 2 and 3, the motor is also flooded in the same oil.

As previously mentioned, the composite actuator was incorporated into a knee housing within a slider-crank configuration, as shown in Fig. 4. The housing was designed in two halves with the main components – the hybrid actuator and a load cell – located between them. Fig. 4 shows the prosthesis with one housing side removed, showing the position of the actuator and load cell within the housing. Fig. 4 also shows the fully assembled knee housing, including the location of an embedded system and battery pack. Contact between the housing and rotational assembly (i.e., crank) provides hard stops in both extension and flexion. The total range of motion of the knee is 130 deg (5 deg hyperextension to 125 deg flexion). Note that the actuator reaches a singularity at 110 deg flexion, such that movement between 110 deg and 125 deg flexion is non-actuated. The total mass of the assembled prosthesis, filled with hydraulic fluid, is 2180 g.

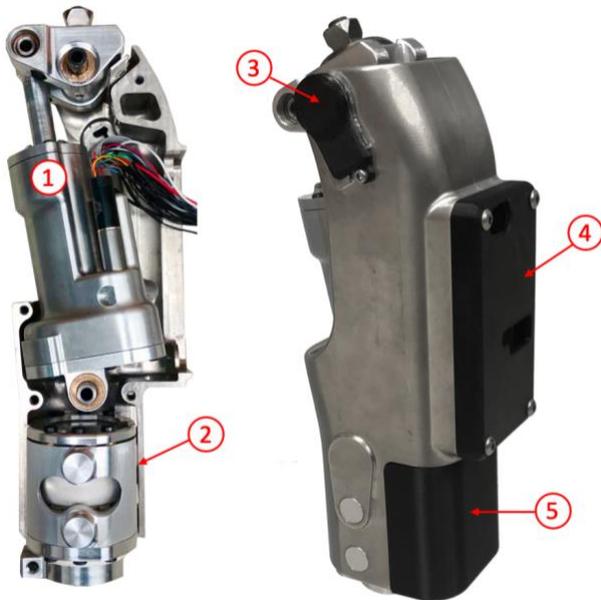


Fig. 4. Fully assembled and open view of the SCSA knee. Image shows (1) hybrid actuator, (2) load cell, (3) covered absolute encoder, (4) covered PCB, and (5) enclosed battery pack

Prosthesis sensing includes a load cell (see Fig. 4) that measures axial load in the shank; an absolute encoder that measures knee angle; and a 6-axis inertial measurement unit (IMU) that measures inertial movement of the shank. The actuator has two controlled outputs: swing-assist motor torque, and valve motor torque. Both the swing assist and valve motors are controlled by PI current control loops which are nested inside of PD position control algorithms. The outputs of the high-level control addressed in this paper correspond to the combination of trajectory signals sent to the valve and swing assist motors, as well as state-dependent modifications to the gains of the PD control loop.

III. CONTROLLER

The device was controlled for the experiments detailed in this paper via a stair ascent controller and a stair descent controller. The stair ascent controller was further subdivided into two sub-tasks: one for step-over and the other for step-to stair ascent. The stair descent controller can perform both step-over or step-to descent with the same controller parameters. These controllers were implemented as finite state machines via Matlab Simulink Realtime on a laptop computer. The output from each controller is comprised of three primary commands: 1) a commanded valve angle; 2) a set of swing-assist control gains; and 3) a desired knee trajectory angle, for cases in which the control gains are non-zero. This information was then transmitted to the device's embedded system via a CAN bus. On board the embedded system, position control loops of the form

$$\tau_r = K_p(\theta_r - \theta) + K_d(\dot{\theta}_r - \dot{\theta})$$

were run around both valve and knee angle, where θ was the measured angle in question, θ_r was the reference to which that angle was being controlled, and τ_r was the reference torque passed along to the nested current control loop.

Stair Ascent

The stair ascent finite-state machine (FSM) consists of three states: stance, swing flexion, and swing extension, as shown in Fig. 5. These states are largely similar for both step-to and step-over gait, with different switching conditions and swing extension trajectory.

During stance phase (State 0), the valve is commanded to a closed position, such that the hydraulic fluid locks the knee against flexion but remains in a low-resistance state in extension due to the check valve. The device's ability to produce such conditions is particularly useful during stair ascent, as it offers the user support against knee buckling and allows for their free extension of the knee, without having to change the valve position. During this state, the swing-assist motor is left unpowered, as knee motion is driven from the hip. Once the leg is unloaded, as measured by the load cell, and assuming other switching gate conditions are met, the leg then transitions into swing flexion.

Upon entry to swing flexion phase (State 1), the valve motor is commanded to open the valve, enabling the leg to be easily flexed. Simultaneously, the angle which the shank makes with the gravity vector is recorded. The reference angle to which the knee is controlled is then modulated to drive the

shank angle to this initial angle plus a predetermined offset (40 deg). This approach to reference angle generation allowed for smooth transitions between step-to and step-over strategies during stair ascent. Finally, at this transition, the control gains are varied and subsequently filtered in a first order manner to provide trajectory tracking while avoiding discontinuities in torque. If at any point during this phase the leg is reloaded (as detected by the load cell), the device transitions back to stance, locking the valve and re-zeroing swing-assist control gains.

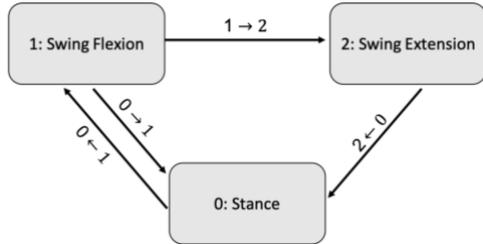


Fig. 5. FSM for stair ascent. switch conditions detailed below in Table 1.

Once the user’s thigh angle has moved through the desired trajectory associated with stair knee flexion, as measured by thigh angle and its first and second derivatives, the device transitions to the swing knee extension phase (State 2). Upon entering this state, the valve is commanded back to a locked position. The extension check valve allows the knee to extend freely, while the rotary valve is ready to fully support the user upon heel strike. The active control gains are then decayed at a first order rate to roughly 10% of the values used in State 1, as the device is now working with gravity to extend the knee. A sinusoidal trajectory is generated to smoothly extend the user’s knee so that it can be placed on the subsequent stair tread. During step-to gait this trajectory extends to a straight knee position, while during step-over gait, it extends to minimize the angle between the shank and gravity. All transition conditions are summarized below in Table 1.

TABLE I. State transition conditions for stair ascent

Transition	Condition
0 → 1	Load ≤ 5lbs && Thigh Angle ≤ 2
1 → 0	Load ≥ 5 lbs
1 → 2	Thigh Angle has cleared 10 degrees && Thigh Jerk ≤ 0 deg/s ² && If Step-to: Thigh Velocity < 100 deg/s If Step-over: Thigh Velocity < 60 deg/s
2 → 0	Load ≥ 5 Leg fully extended for 0.15 s

Stair Descent

The stair descent FSM for this device consists of two states: stance and swing, as shown in Fig. 6. The same FSM and parameters are used for both step-to and step-over gait.

During the stance phase of gait (State 0), swing-assist control gains are set to zero and the device’s hydraulic damping valve angle is commanded to a position corresponding to medium-high damping. The specific damping value used was tuned to the subject’s preference, as it would be in an MPK, such that the user could descend the stair in a controlled fashion and at a self-selected rate.

Because of the nonlinear transmission of the slider crank mechanism, the damping valve angle is varied throughout knee flexion in order to maintain the subject’s preferred level of damping. This allows the user to feel fully supported through the entirety of their stair descent trajectory.

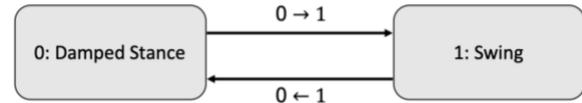


Fig. 6. FSM for stair descent. Switching conditions shown below in Table 2.

Once the knee angle is sufficiently flexed and the leg is unloaded, the knee then transitions into the swing phase (State 1). The valve motor is commanded to open the valve, the swing-assist controller gains are increased and passed through a first order filter to a moderate proportional and damping gains, and a sinusoidal swing trajectory is generated. This trajectory initially flexes the knee in order to allow toe clearance over the edge of the step, then extends the knee to enable foot placement on the subsequent stair tread. After the knee angle trajectory has cleared its maximum flexion, the valve is commanded to an angle corresponding to stance damping in order to support the user on the next step. At foot strike, the device transitions back into stance phase.

TABLE II. State transition conditions for stair descent

Transition	Condition
0 → 1	Load ≤ 5 lbs && Knee Angle ≥ 30 deg
1 → 0	Load ≥ 5 lbs Knee Angle ≤ 5 deg

IV. EXPERIMENTAL ASSESSMENT

In order to test the efficacy of the device and controller suite described above for stair ambulation, the system was implemented on a single subject and compared to stair ambulation with his daily-use prosthesis. The experimental protocol was approved by the Vanderbilt University Institutional Review Board. The test subject was capable of step-to and step-over stair ascent and descent (with use of a hand rail). At the time of the experiment, the subject measured 1.85 m 98.4 kg (6’1”, 217 lb), and used the Ossur Rheo Knee as his daily-use MPK. The subject expressed prior preference for step-to gait for both stair ascent and descent. The subject mentioned that, due to circumduction of his hip during ascent, he was unable to perform step-over stair ascent in narrow stair wells. He was brought into the lab setting for two training sessions on the SCSA knee, during which time the down stairs damping settings were tuned to his preference.

During the experiment, the subject donned Xsens motion-capture devices on his pelvis, upper thighs, lower leg, and feet. He then ascended two floors of stairs (52 steps total), rested for a self-selected amount of time, and then descended back down the same flights of stairs. This was repeated twice for step-over gait, and twice for step-to gait. This was first done on his daily-use prosthesis, and then repeated on the SCSA device. Fig. 7 shows the subject ascending the steps with the SCSA prosthesis (left column), and with the daily-use prosthesis (right column). A video is also included with the supplemental material for this paper, showing the subject

performing ascent and descent with both devices. Following the experiment, the data was parsed on a stride-by-stride basis, excluding the first step of each flight of stairs.

V. RESULTS AND DISCUSSION

Figs. 8 through 10 show data corresponding to stair ascent and descent with both the daily-use and SCSA knee prostheses. Specifically, the figures show for approximately 50 strides, the mean knee flexion angle, hip flexion angle, and hip ab/adduction angle, as a function of stride, for ambulation with the daily-use MPK (black line), and for ambulation with the SCSA prosthesis (red line). Plus and minus one standard deviation across the 50 strides is also shown for each prosthesis via the shaded bands around the means. For the step-over stair gaits there is a direct non-amputee analog corresponding healthy data [13] shown in the plots with a dashed line.

As shown in Fig. 7 and in the supplementary video, differences in behavior was easily observable in the experimemnts. Using the SCSA results in a notable increase in knee flexion, as well as a clear difference in the amount the user has to elevate his hip in order to clear the stair.

A. Step-over Stair Ascent

The subject was able to ascend stairs (with use of the handrail) in a step-over fashion for both prostheses. As shown in Fig. 8, when using his daily-use prosthesis, knee angle achieved during step-over gait was substantially less than when using the SCSA knee. When using his daily-use prosthesis, average peak knee angle achieved was 13.5° , as compared to the average peak knee flexion of 87.5° achieved by the SCSA. Note that healthy knee flexion reaches approximately 90° .



Fig. 7. Subject ascending stairs in a step-over fashion. The left column shows the subject's gait using the SCSA device, the right showing his daily use device. A video corresponding to these experiments is provided in the supplementary materials.

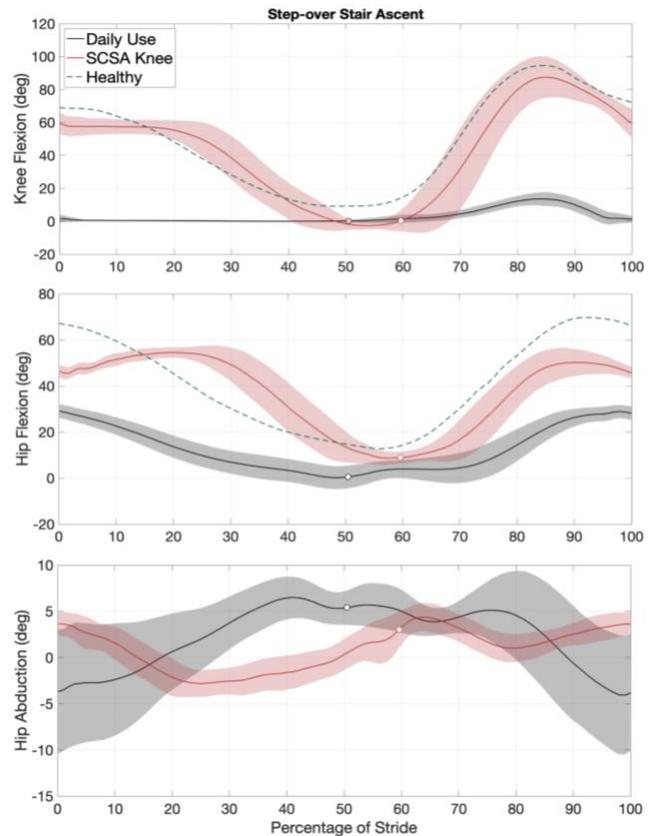


Fig. 8. Mean step-over stair ascent data for 50 steps of SCSA prosthesis (red) and daily-use prosthesis (black). Shaded bands are \pm one standard deviation. White dots indicate mean timing of toe-off for each respective device. Dashed lines show corresponding data for healthy subject.

With the lower knee flexion on his daily-use device, additional compensatory hip abduction/adduction was observed during these tasks. The RMS value of the mean hip abduction on his daily-use device was 4.1° , as compared to 2.4° on the SCSA. Additionally, the hip abduction/adduction was observed to be much less uniform during the step over task on his daily-use device when compared to the SCSA knee. Finally, with knee flexion which more closely resembled healthy knee flexion, the hip flexion also trended toward healthy gait. On his daily-use device, the maximum and minimum hip flexion angles were 29.0° and 0.2° , respectively; as compared to healthy data of 69.7° and 12.7° . On the SCSA these mean hip flexion angles were 54.4° and 8.6° , much more representative of the healthy trajectory.

B. Step-to Stair Ascent

Fig. 9 shows knee and hip data for step-to stair ascent. As seen in the figure, negligible knee flexion was observed when using the daily-use MPK. In order to compensate, the subject employed various compensatory actions with his body, including substantial hip ab/adduction. The SCSA device had a mean peak knee flexion of 47.8° during step-to gait, as compared to the daily-use device's 2.5° . The enhanced knee flexion substantially decreased the peak abduction values (shown as negative on the plot). Anecdotally, the subject remarked that when climbing stairs using the SCSA in a step-to fashion, he felt a noticeable decrease in strain of his low-back.

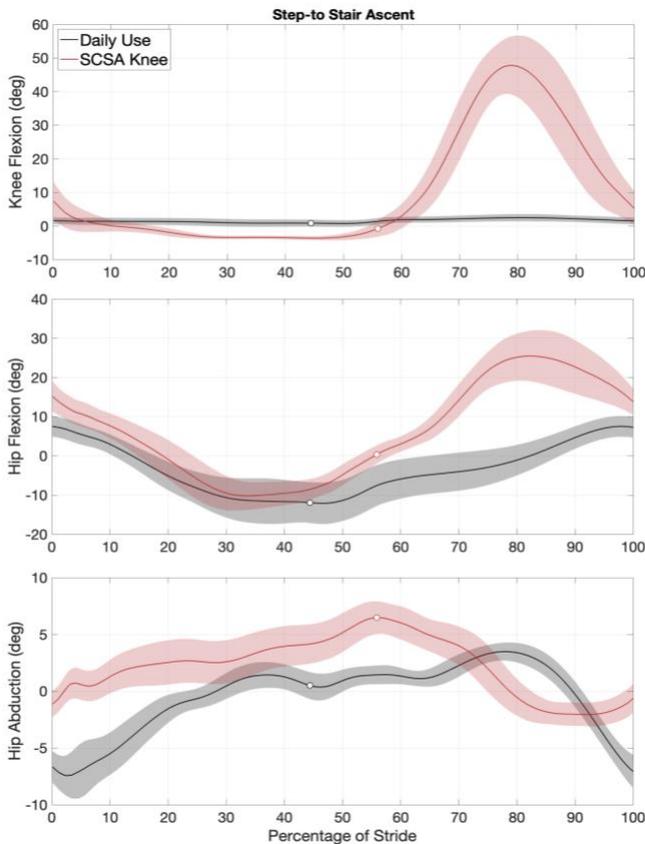


Fig. 9. Mean step-to stair ascent data for 50 steps of SCSA prosthesis (red) and daily-use prosthesis (black). Shaded bands are \pm one standard deviation. White dots indicate mean timing of toe-off for each respective device.

C. Stair Descent

When descending stairs, the subject was able to perform both step-to and step-over gaits with both prostheses with the aid of a handrail. The data from the step-over trial are displayed in Fig. 10. No data is shown for the step-to task, since the knee in both cases is held straight, so little information is provided by it. In the step-over stair descent data (Fig. 10), the behavior of the two devices is also not notably different from each other, as it is a primarily dissipative task, one which is well suited to an MPK. As observed in the figure, the SCSA provided a more consistent rate of knee flexion during stance-knee yielding. This behavior was provided intentionally in response to the subject, who expressed dissatisfaction with the rapid “drop-off” experienced on his daily-use device. In addition to the more consistent rate of flexion, the trajectory to full extension following flexion was notably quicker on the SCSA relative to the daily-use device. This is due to the swing-assist feature, which appeared to enhance the ability of the subject to place his foot on the stair tread.

The stair descent stance capabilities of this device and control regime were shown to maintain and/or improve upon those offered by an MPK. At the same time, the aided extension during stair descent swing enables quicker extension and easy foot placement.

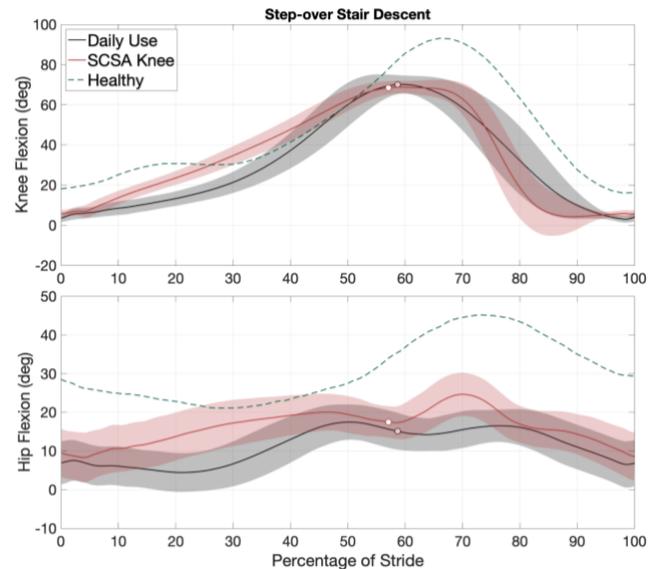


Fig. 10. Mean step-over stair descent data for 50 steps of SCSA prosthesis (red) and daily-use prosthesis (black). Shaded bands are \pm one standard deviation. White dots indicate mean timing of toe-off for each respective device. Dashed lines show corresponding data for healthy subject.

VI. CONCLUSION

This paper describes the design and preliminary testing of a stance-controlled swing-assisted knee prosthesis intended to provide enhanced functionality relative to a conventional MPK prosthesis. This paper specifically describes prospective enhancements for purposes of stair ascent and descent, during which the SCSA prosthesis is hypothesized to provide improved swing-phase kinematics relative to the MPK, particularly during stair ascent, and therefore reduce compensatory movements such as hip elevation and circumduction. Experimental testing with a single transfemoral amputee subject indicates corresponding substantial improvement in swing-phase knee flexion, relative to an MPK, during both step-to and step-over stair ascent. Due to this increased knee flexion, compensatory hip circumduction was reduced in the subject’s movement, and the corresponding stance phase was further normalized, as a result of a corresponding narrower stance. Hip and knee motion during both stance and swing when using the prosthesis was shown to better match the motion of a healthy individual when climbing stairs, relative to a standard MPK.

Transitions in and out of each task have been explored, but are beyond the scope of this paper, and will be further discussed in later work. Additionally, future work will explore to what extent these controllers need to be manually tuned between subjects, though preliminary testing indicates this will be minimal, if at all necessary.

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