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Powered Knee and Ankle Prosthesis with Adaptive Control Enables Climbing Stairs with Different Stair Heights, Cadences, and Gait Patterns

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Abstract—Powered prostheses can enable individuals with above-knee amputations to ascend stairs step-over-step. To accomplish this task, available stair ascent controllers impose a pre-defined joint impedance behavior or follow a pre-programmed position trajectory. These control approaches have proved successful in the laboratory. However, they are not robust to changes in stair height or cadence, which is essential for real-world ambulation. Here we present an adaptive stair ascent controller that enables individuals with above-knee amputations to climb stairs of varying stair heights at their preferred cadence and with their preferred gait pattern. We found that modulating the prosthesis knee and ankle position as a function of the user's thigh in swing provides toe clearance for varying stair heights. In stance, modulating the torque-angle relationship as a function of the prosthesis knee position at foot contact provides sufficient torque assistance for climbing stairs of different heights. Furthermore, the proposed controller enables individuals to climb stairs at their preferred cadence and gait pattern, such as step-by-step, step-over-step, and two-step, similar to able-bodied individuals. We anticipate the proposed control strategy will improve the robustness of powered prostheses to environmental and human variance without the need for expert tuning, machine learning, or direct subject intervention, which may enable powered prostheses to more easily move from the lab to the real-world.

Index Terms—Robotic Leg, Prosthesis, Powered Prosthetics, Amputation, Gait, Biomechanics

I. INTRODUCTION

CONVENTIONAL knee and ankle prostheses cannot provide net-positive energy [1], which is necessary to propel the body forward and upward during ambulation [2]. Additionally, they cannot generate torque, which is critical for active control of the joint movements to achieve toe clearance in swing [2]. While walking, individuals with an above-knee amputation make up for the deficiencies in their passive prostheses by performing undesirable compensatory movements with their residual limb, intact leg, and upper body [3]–[5]. Unfortunately, these compensatory movements are

insufficient for most users to ascend stairs step-over-step [4]–[6]. As a result, individuals with a conventional passive prosthesis commonly ascend stairs using a slower and less efficient step-by-step gait pattern, leading each step with their intact leg [4]–[6]. With this step-by-step pattern, the intact leg and upper body perform all the effort required to climb the step, which requires significant strength and endurance [3], [7]. Moreover, the residual limb hip joint needs to extend and circumduct unnaturally for the passive prosthesis to clear the step during swing as the prosthetic knee joint cannot flex as the biological leg would [8]. This residual limb extension is often difficult due to muscle contractures [9] further challenging the user's balance [10]. Improved prosthesis technologies are needed to enable individuals with above-knee amputations to ascend stairs more naturally.

Powered prostheses [11] have the potential to imitate the biological leg biomechanics during stair ascent [12], [13]. A powered prosthesis can propel the body upward by injecting positive energy when the prosthetic foot is in contact with the step (i.e., stance phase). Also, a powered prosthesis can ensure adequate clearance with the step and correctly place the prosthetic foot in preparation for the next step to be climbed by actively controlling the joint movements when the prosthetic foot is off the ground (i.e., swing phase). Because climbing taller stairs requires larger net-positive energy and higher joint torque than climbing smaller stairs [12], the torque generated by the prosthesis in stance phase must be adapted to the stair height. In addition, different stair heights or variations in gait patterns require the prosthesis to change the swing movement trajectory so that proper clearance and foot placement on the step can be achieved. Thus, powered prosthesis controllers must be robust to variability in stair geometry, gait pattern, and gait cadence to enable ascending stairs in the real world.

Experimental studies have shown that powered prostheses can enable above-knee amputee subjects to ascend stairs step-over-step [14]–[22] resulting in improved stair ambulation speed and reduced metabolic effort compared to using a conventional passive prosthesis [19]. However, available stair ascent controllers are designed to produce a predefined, fixed action of the powered prosthesis [15], [17], which must be

manually tuned for each subject and staircase [23]. Thus, if a user attempts to climb a taller step than the one the stair controller was tuned for, the prosthesis may not provide enough clearance, the prosthetic foot may hit the step riser, and the user may trip and fall. Even if the step is cleared by the user through hip circumduction and sound-side vaulting, the prosthetic foot may not lay flat on the step. So, when the prosthetic knee starts generating torque as necessary to climb the step, the subject may be pushed backward rather than upward, potentially causing the user to fall. Similarly, the amount of energy injected by the prosthesis when climbing the step may be too large or too small, if a shorter or taller step is taken compared to the tuning step. We need powered prosthesis controllers that automatically adapt to the variability of stair height encountered in the real world.

In this paper, we propose an alternative control strategy for a powered knee and ankle prosthesis to ascend stairs with varying stair heights, cadences, and gait patterns. Specifically, we propose modulating the prosthesis knee and ankle position in swing as a continuous function of the user's thigh position, thigh velocity, and vertical acceleration of the prosthesis. In addition, we propose modulating energy injection in stance using a continuous adaption of the knee joint torque-angle relationship as a function of the prosthesis knee position when the prosthetic foot contacts the step. Previous studies have proposed using the residual limb orientation as a proxy for the gait phase to produce a fixed prosthesis trajectory in level-ground walking [24], [25]. In contrast, we propose changing the swing trajectory online during stair ambulation. We hypothesize that such a continuous modulation of stance energy and swing trajectory will enable individuals with above-knee amputations to climb stairs of different heights at different cadences and to seamlessly transition between different stair climbing strategies, such as step-by-step, step-over-step, and two-step. We will test this hypothesis by asking an individual with a unilateral above-knee amputation to ascend stairs of different heights using different gait patterns at their self-selected speed. We expect a stair controller with these characteristics to facilitate translating powered prostheses to real-world use.

II. METHODS

A. Adaptive Stair Controller

The proposed controller uses a finite-state machine with two states *Standing* and *Lifting*. When the ground reaction force (GRF) is lower than a predefined threshold ( ) the stair controller is in *Lifting*. In *Lifting*, the desired angles of the knee joint ( ) and ankle joint ( ) are continuously adapted based on the movements of the user's thigh (i.e., residual limb). To this end, the desired angle of the knee joint ( ) is defined as the sum of three terms ( ) determined by (1) to (3) (Fig. 1(a-d)). The first term ( ) is proportional to the orientation of the user's thigh with respect to gravity ( ) provided a predefined certain threshold ( ) is exceeded as defined in (1) and shown in Fig. 1(a).

$$\begin{aligned} &= ( ) \\ &= 0 \end{aligned} \quad (1)$$

Fig. 1. *Lifting Controller*. (a-d) Express the relationship between the thigh position (a), thigh velocity (b), and thigh vertical acceleration (c,d) for the powered prosthesis knee joint desired position. (e-g) Express the relationship between the thigh position (e), and thigh vertical acceleration (f,g) for the powered prosthesis ankle joint desired position.

The second term ( ) is proportional to the positive angular velocity of the user's thigh as defined in (2) and shown Fig. 1(b).

$$\begin{aligned} &= 0 \\ &= 0 \end{aligned} \quad (2)$$

The third term ( ) depends on the vertical acceleration of the user's thigh with respect to gravity ( ). As defined in (3) and shown in Fig. 1(c) a first factor ( ) is subtracted to the thigh acceleration ( ) before calculating the double integral. The result of the double integration is then multiplied by a second non-constant factor ( ). This multiplication factor changes as a function of the thigh orientation ( ) as shown in Fig. 1(d). Specifically, the multiplication factor is kept constant ( ) until the thigh orientation ( ) exceeds a certain threshold ( ). Above said threshold, the multiplication factor decreases linearly, reaching zero when the thigh orientation equals the thigh threshold plus 5° (+ 5°), Fig. 1(d).

$$\begin{aligned} &= ( ) \\ &< \\ &= ' \\ &= \\ &= , \end{aligned} \quad (3)$$

So, k_4 works as a linear gain that decreases as the thigh orientation angle increases after a certain threshold has been achieved. With the proposed algorithm, the knee flexion

additional stability in *Standing*. With the proposed adaptive controller, larger knee extension torque is produced when the powered prosthesis transitions between *Lifting* and *Standing* at a larger knee flexion angle, ultimately injecting higher mechanical energy into the stair-climbing cycle. Moreover, if the powered prosthesis transitions between *Lifting* and *Standing* with the knee fully extended, for example when the user shuffles around with no intention to climb a step, the desired torque is defined solely by the impedance component, which stabilizes the knee joint and prevents it to collapse under the user's body weight. Thus, the proposed *Standing* controller adapts the desired knee torque and energy injection with the stair height while leaving the user the freedom to take the step at their preferred cadence.

The ankle behavior during *Standing* is defined using an impedance controller with an adaptive virtual equilibrium angle (θ_{eq}). Due to the adaptive nature of the *Lifting* controller, the angle of the powered ankle joint at the transition between *Lifting* and *Standing* is not fixed, but changes as a function of the user's residual limb orientation and acceleration as defined by (4) to (5). Thus, at the transition between *Lifting* and *Standing*, the equilibrium angle of the ankle (θ_{eq}) is set to the measured ankle angle. Then, the equilibrium angle of the ankle changes linearly with the knee position as defined by (6).

$$\theta_{eq} = \frac{0}{0} \left(\theta_{knee} - \theta_{knee}^0 \right) \quad (6)$$

With the proposed *Standing* controller, the powered ankle joint moves from whatever its initial angle is when the prosthetic foot contacts the step to a neutral position (i.e., 0°) when the powered knee joint is fully extended. Thus, the powered ankle can contribute positive power to the *Standing* movement. At the same time, if the subject shuffles around without taking a step, the ankle stays in a neutral position while providing compliant support to help the user balance while standing.

The desired torque or angle defined by the *Standing* and *Lifting* controllers is enforced by a dedicated low-level controller using a hybrid feedforward/feedback approach. During *Lifting*, closed-loop position controllers (Fig. 3(a)) are used to impose the desired joint angles at the ankle and knee joints. For each powered joint, the closed-loop position controller takes as input the desired angle (θ_d) and compares it to the measured angle (θ_m), which is estimated using a complementary filter. The angle error (e_θ) is fed to a PID controller that determines the desired torque command (τ_d). In *Standing*, the ankle joint uses a virtual impedance controller after [11], [18] (Fig. 3(c)) with predefined stiffness and damping parameters (k , b) to define the desired torque command (τ_d). The desired torque command is then fed to a low-level torque controller (Fig. 3(c)). The low-level torque controller comprises a feedforward command based on the position-dependent transmission ratio (r). In addition, two compensators are used to reduce the apparent impedance (i.e., viscosity and inertia) of the transmission system to improve the fidelity of the virtual impedance controller [21]. The first one (τ_{vis} , τ_{in}) takes as input the motor position and generates an online estimate of the viscous torque due to the linear actuator. The second one takes as input the motor position and

computes a scaled and low-pass filtered estimate of the transmission inertia similar to [28] (i.e., $\hat{I}$).

B. Human Testing

One subject with an above-knee amputation was recruited to participate in the study. Before the study took place, the subject provided written informed consent to participate, including written consent to use photos and videos of the experiment for dissemination purposes. The subject is 27 years old, weighs 65 kg, is a 1.7 m tall male, and has had an above-knee amputation for 6 years. The subject had prior experience with this powered prosthesis through participation in other studies. However, he did not have prior experience with the proposed controller. All study protocols were approved by the University of Utah Institutional Review Board.

The experiment preparation took place before data collection. The subject donned the Utah Lightweight Leg [17], [29], which is a fully powered, modular, and self-contained knee and ankle prosthesis. A certified prosthetist adjusted the build height of the prosthesis using a standard pylon and ensured proper alignment of the knee and ankle joints. After the prosthesis fitting was completed, the subject donned an IMU-based motion capture system [30] (MTw Awinda, Xsens). Eight sensors were placed on the subject. Two sensors were placed on the top of each foot, two on each shank just below the knee joint, two on the outside of each thigh, one in the center of the lower back, and one sensor on the sternum. Then, the motion capture system was calibrated to the subject. The calibration protocol consisted of having the subject stand still for 5 seconds, take 3 strides forward, turn around, take another 3 strides, and return the original standing position [30]. After the system calibration, the subject practiced climbing stairs with the proposed controller for about 15 minutes. During practice, the controller parameters were fine-tuned by the experimenter

Fig. 4. The subject climbs stairs with the Utah Lightweight Leg. 4 in stairs on the left, 7 in stairs on the right. Step-by-step and step-over-step gait pattern shown on top, two-steps gait pattern shown on bottom.

based on the subject's preference. The experiment preparation lasted about 30 minutes.

After the experiment preparation, the subject performed the experimental protocol for data collection. The subject was asked to ascend two staircases of 4 steps, each with 3 different gait patterns. The first staircase is the maximum ADA compliant step height of 7 inches (18 cm), the second staircase is the minimum ADA compliant step height of 4 inches (10 cm) [31]. First, the subject used the *step-over-step* gait pattern, which is the most common way to climb stairs for non-amputee individuals. When climbing stairs *step-over-step*, each foot is placed one step above the other foot. Then, the subject used a *step-by-step* gait pattern, which is the most common stair ascent method for above-knee amputees using conventional prostheses. When climbing stairs *step-by-step*, the leading foot is placed one step above, and the following foot is brought to match on the step of the leading foot. Finally, the subject used a *two-steps* gait pattern, which is less common and mostly used when in a hurry. When climbing stairs with the *two-steps* gait pattern, the leading leg is taking two-steps at a time and the following leg is brought to match that step. The subject performed 5 ascents for each gait pattern and staircase. The subject was asked to climb the staircase at the preferred cadence. Thus, the protocol tested all possible combinations of gait patterns and stair heights, while leaving the gait cadence up to the user's preference.

Data acquired from the motion capture systems and the sensors embedded in the powered prosthesis were processed offline. The motion capture system provides the kinematics of the ankle, knee, and hip joint, the orientation of the leg segments, and the Cartesian-space position of the toe, ankle, knee, and hip joints for both the prosthesis side and the sound side. The powered prosthesis provides the kinetics and kinematics of the prosthetic ankle and knee joints. Data recorded from the motion capture system and the powered prosthesis were synchronized online through Wi-Fi. The

synchronized raw data was filtered offline using a zero-lag low-pass Butterworth filter with a cutoff frequency of 8 Hz. Joint angular velocities, accelerations, and power were calculated post-filtering. Segmentation indexes for stance and swing phase during stair ascent were determined using the gait state parameters defined online by the powered prosthesis controller. Full strides started and ended at toe-off on the prosthesis side. After segmentation, each stride was resampled to 1000 samples, and the time was normalized as percent of stride accomplishment. Energy per stride was calculated as the integral of the joint torque-angle curve during stance phase only. Averages and standard deviations of each condition were calculated across strides for all the variables of interest. A two-factor ANOVA was performed on all variables against gait pattern and stair height conditions. Post-hoc analysis was performed using Tukey's Honestly Significant Difference (HSD) for statistical analysis between the different gait pattern conditions. The offline data analysis provides both quantitative and qualitative measures to assess the performance of the proposed controller.

C. Powered Knee and Ankle Prosthesis

For this experiment, we used the Utah Lightweight Leg (Fig. 4). This powered knee and ankle prosthesis can generate biologically appropriate torque and power during ambulation while fitting within the 50th percentile male leg profile [17]. The Utah Lightweight Leg weighs ~2.7 kg including battery and protective covers—approximately half of similar powered prostheses [21]. The ankle-foot module uses a compact, lightweight powered polycentric design, which is contained within a commercially-available foot shell [29][32]. The powered polycentric mechanism is connected to custom carbon-fiber feet of different sizes to accommodate different subjects. The knee module uses an actively variable transmission [16], [33] to optimize the effective transmission ratio and leg dynamics for different locomotion tasks. In

TABLE 1

STATISTICAL ANALYSIS OF THE POWERED PROSTHESIS STAIR ASCENT GAIT PATTERN AND STAIR HEIGHT COMPARISON

Variable		Step-By-Step	Step-Over-Step	Two-Steps	ANOVA		
					Gait Pattern	Tukey HSD	Stair Height
Max Knee Flexion During Swing	4-in	74.9±2.8°	73.6±1.8°	95.9±1.5°	F(2,20)=31.57 p=6.49x10 ⁻⁷	Step-By-Step & Two-Steps Step-Over-Step & Two-Steps	F(1,20)=29.00 p=2.85x10 ⁻⁵
	7-in	88.5±2.9°	88.3±2.6°	96.0±1.9°			
Thigh Position Start of Stance	4-in	-15.5±2.0°	-30.7±6.2°	-43.8±2.1°	F(2,20)=45.65 p=3.51x10 ⁻⁸	Step-By-Step & Two-Steps Step-Over-Step & Two-Steps	F(1,20)=56.39 p=3.06x10 ⁻⁷
	7-in	-40.6±3.1°	-40.6±3.5°	-57.9±3.1°			
Knee Position Start of Stance	4-in	50.0±6.5°	49.9±3.4°	68.8±3.6°	F(2,20)=4.50 p=4.19x10 ⁻⁶	Step-By-Step & Two-Steps Step-Over-Step & Two-Steps	F(1,20)=130.06 p=3.32x10 ⁻¹⁰
	7-in	75.3±1.0°	74.5±1.9°	84.4±5.3°			
Peak Knee Torque Stance	4-in	0.40±0.15 Nm/kg	0.50±0.10 Nm/kg	1.20±0.08 Nm/kg	F(2,20)=46.98 p=2.78x10 ⁻⁸	Step-By-Step & Two-Steps Step-Over-Step & Two-Steps	F(1,20)=68.77 P=6.66x10 ⁻⁸
	7-in	1.06±0.06 Nm/kg	1.03±0.04 Nm/kg	1.44±0.15 Nm/kg			
Peak Knee Power Stance	4-in	1.52±0.63 W/kg	1.90±0.43 W/kg	3.60±0.68 W/kg	F(2,20)=29.51 p=1.08x10 ⁻⁶	Step-By-Step & Two-Steps Step-Over-Step & Two-Steps	F(1,20)=70.10 P=5.73x10 ⁻⁸
	7-in	3.37±0.25 W/kg	3.39±0.27 W/kg	4.71±0.20 W/kg			
Knee Energy Injection	4-in	0.17±0.08 J/kg	0.20±0.04 J/kg	0.64±0.08 J/kg	F(2,20)=65.63 p=1.63x10 ⁻⁹	Step-By-Step & Two-Steps Step-Over-Step & Two-Steps	F(1,20)=122.26 p=5.69x10 ⁻¹⁰
	7-in	0.60±0.03 J/kg	0.59±0.03 J/kg	0.95±0.14 J/kg			
Peak Thigh Position Stride	4-in	-40.9±1.8°	-41.0±3.3°	-55.5±2.3°	F(2,20)=26.25 p=2.55x10 ⁻⁶	Step-By-Step & Two-Steps Step-Over-Step & Two-Steps	F(1,20)=70.85 p=5.27x10 ⁻⁸
	7-in	-55.7±2.0°	-56.4±0.8°	-61.6±3.9°			
Swing Duration	4-in	1.06±0.08 s	0.85±0.09 s	1.52±0.09 s	F(2,20)=63.23 p=2.26x10 ⁻⁹	Step-By-Step & Two-Steps Step-Over-Step & Two-Steps	F(1,20)=25.44 p=6.22x10 ⁻⁵
	7-in	1.20±0.05 s	1.18±0.13 s	1.67±0.11 s			

addition, the knee module contains the battery, control electronics, and motor drivers for both the knee and ankle joints [17]. The knee and ankle modules connect with a standard 30-mm pylon allowing for height and intra-extra rotation adjustments. A custom instrumented pyramid adapter is located at the top of the ankle modules to estimate the ground reaction force and torque [34]. The Utah Lightweight Leg is equipped with position sensors at the knee and ankle joints as well as at the motor shafts. Three 9-DOF IMUs (MPU9250, Invsense) sense the movements and the orientation of the foot, shank, and thigh segments in space. The 3-D printed protective covers host the Li-Ion battery (2500 mAh, 6S) and the onboard system-on-module (SOM, MyRio 1900, National Instruments). The SOM runs all custom control algorithms in real-time interfacing with the sensors and servo drivers for the knee and ankle modules. The SOM can be connected through wi-fi to a host computer or smartphone for data monitoring and controller tuning.

III. RESULTS

The swing trajectories for different stair heights and gait patterns were visibly different (Fig. 5). The maximum knee angle during swing was $88.5 \pm 2.9^\circ$, $88.3 \pm 2.6^\circ$, and $96.0 \pm 1.9^\circ$ for step-by-step, step-over-step, and two-steps gait patterns for the 7-in stairs, respectively (Fig. 7). The maximum knee angle during swing was $74.9 \pm 2.8^\circ$, $73.6 \pm 1.8^\circ$, and $95.9 \pm 1.5^\circ$ for step-by-step, step-over-step, and two-steps gait patterns for the 4-in stairs, respectively (Fig. 7). There were significant differences between the maximum knee flexion angles achieved with different gait patterns ($F(2,20) = 31.57$, $p=6.49 \times 10^{-7}$) and on different stair heights ($F(1,20) = 29.00$, $p=2.85 \times 10^{-5}$), TABLE 1. Post-hoc analysis revealed that there were significant differences in the maximum knee flexion angle between the step-by-step and step-over-step gait patterns compared to the two-steps gait pattern. However, the maximum knee flexion angle was not statistically significant between the step-by-step and step-over-step gait patterns. There was a statistically significant difference found between stair heights, with an 18% and 20% increase in knee flexion between the 7-in and 4-in stairs for the step-by-step and step-over-step gait patterns, respectively. No significant differences were found for the ankle angle at the end of swing and maximum plantarflexion angle with respect to gait pattern or stair height ($p>0.3$, TABLE 1). The average foot orientation at the end of swing was $3.0 \pm 1.5^\circ$ for all stair heights and gait patterns. The swing duration was significantly different between gait patterns ($F(2,20) = 63.23$, $p=2.26 \times 10^{-9}$) and stair heights ($F(1,20) = 25.44$, $p=6.22 \times 10^{-5}$), TABLE 1. The swing duration ranged from 0.76 s for the 4-in stairs with step-over-step gait pattern and 1.80 s for the 7-in stairs with the two-steps gait pattern. The step-over-step gait pattern on the 7-in stairs had the highest deviation in swing duration, with a minimum of 1.1 seconds and a maximum of 1.4 seconds, Fig. 5. The proposed controller enabled the powered prosthesis to provide adequate foot clearance and proper foot placement for all gait patterns and stair heights, while the subject was in control of the cadence.

There was a significant difference in the knee angle at the start of stance (i.e., 0% Stance, Fig. 6) between the different stair heights ($F(1,20) = 130.06$, $p=3.32 \times 10^{-10}$) and gait patterns ($F(2,20)=24.49$, $p=4.19 \times 10^{-6}$), TABLE 1. The knee angle at the

Fig. 5. Analysis of the powered prosthesis during swing for different stair heights (7-in and 4-in, left to right, respectively) and gait patterns (step-by-step, step-over-step, two-steps, top to bottom, respectively). The top plots show the swing trajectory of the powered prosthesis from the cartesian space, and the solid black line highlights the position of the toe throughout the swing movement. The bottom plot shows the time duration of swing for each of the different conditions.

start of stance was $75.3 \pm 1.0^\circ$, $74.5 \pm 1.9^\circ$, and $84.4 \pm 5.3^\circ$ for step-by-step, step-over-step, and two-steps gait patterns for the 7-in stairs, respectively. The knee angle at the start of stance was $50.0 \pm 6.5^\circ$, $49.9 \pm 3.4^\circ$, and $68.8 \pm 3.6^\circ$ for step-by-step, step-over-step, and two-steps gait patterns for the 4-in stairs, respectively, TABLE 1. Thus, the knee angle at the start of stance was 51%, 49%, and 23% larger on the 7-in stairs compared to the 4-in stairs for the step-by-step, step-over-step, and two-steps gait pattern, respectively. Also, the knee angle was 13% and 38% larger for the two-steps gait pattern compared to the single-step gait patterns for the 7-in and 4-in stair heights, respectively. The proposed controller changed the prosthesis knee angle at the start of stance based on the gait pattern and stair height.

There was a significant difference in the peak knee torque in stance (Fig. 6) between gait pattern ($F(2,20) = 68.77$, $p = 6.66 \times 10^{-8}$) and stair height ($F(1,20) = 68.77$, $p = 6.66 \times 10^{-8}$), TABLE 1. The peak knee torque was 1.06 ± 0.06 Nm/kg, 1.03 ± 0.04 Nm/kg, and 1.44 ± 0.15 Nm/kg for step-by-step, step-over-step, and two-steps gait patterns for the 7-in stairs, respectively, Fig. 6. The peak knee torque was 0.40 ± 0.15 Nm/kg, 0.50 ± 0.10 Nm/kg, and 1.20 ± 0.08 Nm/kg for step-by-step, step-over-step, and two-steps gait patterns for the 4-in stairs, respectively, Fig. 6. Thus, the peak knee torque increased by 38% and 166% for the two-steps gait pattern compared to the single-step gait patterns for the 7-in and 4-in stairs, respectively. The peak knee torque measured for the 7-in stairs was 164%, 106%, and 20% larger compared to the 4-in stairs for the step-by-step, step-over-step, and two-steps gait pattern, respectively. The peak knee torque was provided at a knee angle of 29.9° , 33.6° , and 52.7° for the step-by-step, step-over-step, and two-steps gait pattern for the 4-in stairs and 53.6° , 52.6° , and 64.2° for the step-by-step, step-over-step, and two-steps gait pattern for the 7-in stairs. Thus, the knee angle at peak torque was 30% smaller than the knee angle at the start of stance. The proposed controller changed the torque based on the gait pattern and stair height.

There was a significant difference in the peak knee power in stance (Fig. 6) between gait patterns ($F(2,20) = 68.77$, $p = 6.66 \times 10^{-8}$) and stair heights ($F(1,20) = 68.77$, $p = 6.66 \times 10^{-8}$), TABLE 1. The peak knee power was 3.37 ± 0.25 W/kg, 3.39 ± 0.27 W/kg, and 4.71 ± 0.20 W/kg for the step-by-step, step-

Fig. 6. Kinematic analysis of the thigh segment, knee joint, and ankle joint for the step-by-step, step-over-step, and two-steps gait patterns from left to right. The prosthesis side average and standard deviation is in blue for the 7-in stairs and orange for the 4-in stairs.

Fig. 7. Kinetic analysis of the stance phase. The step-by-step, step-over-step, and two-steps gait patterns are displayed from top to bottom, respectively. The first two columns show the sound and the prosthesis side orientation in the cartesian space at the start of stance (e.g. 0% stance) for the 7-in stairs on the left and 4-in stairs on the right. The knee torque against knee position plots, knee torque against % stride and knee power against % stride are shown in columns 3 to 5 for all conditions. The bar plots show the energy injection for each condition along with the respective average and standard deviations.

Fig. 8. Kinematic analysis of the thigh segment, knee joint, and ankle joint for the step-over-step gait pattern. The left column is the 7-in stairs and the right column is the 4-in stairs. The sound side average and standard deviation is in black for both stair heights. The prosthesis side average and standard deviation is in blue for the 7-in stairs and orange for the 4-in stairs.

over-step, and two-steps gait patterns for the 7-in stairs, respectively, Fig. 6. The peak knee power was 1.52 ± 0.63 W/kg, 1.90 ± 0.43 W/kg, and 3.60 ± 0.68 W/kg for the step-by-step, step-over-step, and two-steps gait patterns for the 4-in stairs, respectively, Fig. 6. The energy injected in stance was 0.60 ± 0.03 J/kg, 0.59 ± 0.03 J/kg, and 0.95 ± 0.14 J/kg for the step-by-step, step-over-step, and two-step gait patterns for the 7-in stairs, respectively, Fig. 6. The energy during stance was 0.17 ± 0.08 J/kg, 0.20 ± 0.04 J/kg, and 0.64 ± 0.08 J/kg for the step-by-step, step-over-step, and two-steps gait patterns for the 4-in stairs, respectively, Fig. 6. The two-steps gait pattern injected 60% and 246% more energy compared to the single-step gait patterns (Fig. 6) for the 7-in and 4-in stairs, respectively. The 7-in stairs injected 253%, 195%, and 48% more energy compared to the 4-in stairs for the step-by-step, step-over-step, and two-steps gait patterns, respectively. Post-hoc analysis revealed that there were significant differences in the energy injection between the step-by-step and step-over-step gait patterns compared to the two-steps gait pattern ($p = 0$, $p = 0$, respectively), TABLE 1. However, the energy injected was not statistically significant between the step-by-step and step-over-step gait patterns ($p=0.909$), TABLE 1. The proposed controller enabled the prosthesis to inject a different amount of energy into the gait cycle depending on the gait pattern and stair height.

For the step-over-step gait pattern, a kinematic analysis was performed between the sound side and the prosthesis side for the thigh orientation, knee angle, and ankle angle, Fig. 8. Looking at the thigh angle and the knee angle, there was no

TABLE 2
STATISTICAL ANALYSIS FOR SOUND SIDE AND PROSTHESIS SIDE COMPARISON

Variable		Sound Side	Prosthesis Side	ANOVA	
				Side	Stair Height
Max Knee Flexion	4-in	$70.1 \pm 3.7^\circ$	$73.6 \pm 1.8^\circ$	$p > 0.3$	$F(1,13)=110.46$ $p=1.01 \times 10^{-7}$
	7-in	$88.5 \pm 2.9^\circ$	$88.3 \pm 2.6^\circ$		
Max Thigh Position	4-in	$-38.5 \pm 2.4^\circ$	$-41.0 \pm 3.3^\circ$	$p > 0.06$	$F(1,13)=167.80$ $p=8.31 \times 10^{-9}$
	7-in	$-53.9 \pm 1.6^\circ$	$-56.4 \pm 0.8^\circ$		
Max Ankle Plantarflexion	4-in	$-39.7 \pm 4.7^\circ$	$-25.2 \pm 0.07^\circ$	$F(1,13)=90.77$ $p=3.15 \times 10^{-7}$	$p > 0.01$
	7-in	$-51.4 \pm 2.3^\circ$	$-25.2 \pm 0.07^\circ$		
Max Ankle Dorsiflexion	4-in	$7.8 \pm 2.2^\circ$	$20.2 \pm 1.7^\circ$	$F(1,13)=43.61$ $p=1.71 \times 10^{-5}$	$p > 0.03$
	7-in	$14.8 \pm 1.5^\circ$	$20.0 \pm 1.0^\circ$		

statistical significance in the peak positions between the sound side and the powered prosthesis side ($p > 0.06$), TABLE 2. The peak knee angle of the knee on the sound side was $88.5 \pm 2.9^\circ$ and $70.1 \pm 3.7^\circ$ for the 7-in and 4-in stairs, respectively, TABLE 2. There was a statistically significant difference between stair heights for the sound side ($F(1,13)=167.80$ $p=8.31 \times 10^{-9}$ and $F(1,13)=110.46$, $p=1.01 \times 10^{-7}$), TABLE 2. At toe-off, or 0% stride, the sound side for the 4-in stairs experiences some initial flexion of $-8.0 \pm 2.2^\circ$ for the thigh and $17.1 \pm 2.5^\circ$ for the knee compared to the sound side for the 7-in stairs and both stair heights of the prosthesis side where the knee and thigh positions at toe-off are closer to a neutral position, Fig. 8. Additionally, the ankle on the sound side experiences $-32.1 \pm 10.8^\circ$ and $-27.6 \pm 11.0^\circ$ of plantarflexion at toe-off for the 7-in and 4-in stair height, respectively, where the prosthesis side starts with a more neutral ankle position for both the 7-in and 4-in stairs, respectively. The peak plantarflexion angle on the prosthesis side was $-25.2 \pm 0.07^\circ$ and $-25.2 \pm 0.07^\circ$ for the 7-in and 4-in stairs, respectively. The peak plantarflexion angle on the sound side was $-51.4 \pm 2.3^\circ$ and $-39.7 \pm 4.7^\circ$ for the 7-in and 4-in stairs, respectively. The sound side experiences 104% and 58% more plantarflexion compared to the prosthesis side for the 7-in and 4-in stairs, respectively, and is found statistically significant ($F(1,13)=90.77$, $p=3.15 \times 10^{-7}$), TABLE 2. The peak dorsiflexion angle on the prosthesis side was $20.0 \pm 1.0^\circ$ and $20.2 \pm 1.7^\circ$ for the 7-in and 4-in stairs, respectively. The peak dorsiflexion angle on the sound side was $14.8 \pm 1.5^\circ$ and $7.8 \pm 2.2^\circ$ for the 7-in and 4-in stairs, respectively. The peak dorsiflexion angle is also found statistically significant between the sound side and the powered prosthesis ($F(1,13)=43.61$, $p=1.71 \times 10^{-5}$), TABLE 2. Neither the peak plantarflexion nor dorsiflexion angle of the ankle had significant differences between stair heights ($p > 0.01$). The powered prosthesis was able to adapt to variations in stair height while following a similar movement to the sound side of the user.

IV. DISCUSSION

Ascending stairs in the real world with a powered prosthesis requires controllers that are robust to changes in stair geometry, gait pattern, and gait cadence. In swing, the powered prosthesis controller must synchronize the movements of the powered prosthetic joints with the movements of the user's residual limb to clear the step and to achieve proper foot placement for the next step to be taken. If the controller moves too fast or too slow

with respect to the user's residual limb, then the prosthesis will hit the stairs, causing the user to trip and fall. Because the step cadence changes depending on stair height and gait pattern, the prosthesis must continuously adapt the swing movement to the user's cadence. Moreover, the position of the prosthetic foot at the end of swing must match the stair height. If the prosthetic knee is too flexed, then the prosthetic foot is waiting above the step. If the prosthetic knee is not flexed enough, the prosthetic foot does not clear the last step. Prosthesis controllers using a fixed, preprogrammed swing trajectory [15]–[20] cannot synchronize with the user. Therefore, users must learn how to time their residual limb movements with the prosthesis to ensure that the step is cleared. Moreover, prosthesis controllers using a fixed, preprogrammed swing trajectory [15]–[20] cannot adapt to different stair heights and gait patterns. Thus, prosthesis controllers using a pre-programmed trajectory can only work for a specific cadence, stair height, and gait pattern. Outside of the specific tuning conditions, the powered prosthesis cannot provide proper toe clearance and foot placement. To address these limitations, we propose changing the prosthesis swing trajectory continuously, based on the movements of the user's residual limb.

The proposed *Lifting* controller continuously changes the knee and ankle angles as a function of the orientation of the user's residual limb. Besides clearing the step, the angle of the prosthetic joints at the end of swing is critical, as a poor prosthesis placement can result in the prosthesis pushing the user away from the stairs, which may cause the user to fall. Regardless of the stair height, right before the foot contacts the step to start stance phase, the knee joint must be flexed to an extent that ensures the prosthesis shank orientation is past the vertical line defined by gravity so that the user's center of mass is above the prosthesis [13]. We achieve this behavior by defining the knee flexion angle as a function of the user's residual limb orientation with respect to gravity (i.e. thigh angle). The closer the thigh angle is to a horizontal orientation, the more flexion occurring at the prosthetic knee joint (Fig. 5). Using this adaptive mechanism, a suitable prosthesis shank orientation is obtained for any stair height (Fig. 5). At the same time, the ankle must be dorsiflexed to ensure the prosthetic foot stays flat on the step. We achieve this behavior by controlling the ankle angle so that the prosthetic foot is perpendicular to gravity past a certain thigh orientation (e.g., 20°, (6)). Our results show that with the proposed controller the prosthetic foot is relatively flat when it contacts the step, regardless of the stair height or gait pattern (Fig. 5). Another advantage of continuously changing the prosthesis swing trajectory as a function of the user's residual limb orientation is that the prosthesis is intrinsically synchronized to the user's movements. If the user moves faster, the prosthesis moves faster (Fig. 5). If the user stops, the prosthesis stops. In contrast to controllers using pre-programmed trajectories [15]–[20], with this approach, the prosthesis does not start moving until the user starts moving. Thus, the proposed swing controller provides proper foot placement for different stair heights and gait patterns, while constantly synchronizing with the user.

The proposed linear relationship between the knee and ankle angles and the user's residual limb orientation is not sufficient to provide toe clearance during swing. To address this issue, the proposed swing controller partly defines the knee and ankle

angles based on the rotational velocity and vertical acceleration of the user's residual limb (2), (3), (5). Due to the velocity dependency, the higher the rotational velocity of the thigh orientation, the larger the knee flexion (Fig. 1). This velocity relationship is critical in ensuring toe clearance across the intermediate steps during swing for any stair height and gait pattern. However, pilot testing showed that rotational velocity dependency was not enough to clear the first step in all possible conditions. At the beginning of swing, the user's residual limb orientation and velocity are very small, especially for tall stairs or when two steps are climbed at the same time (Fig. 5). Thus, we use the vertical acceleration of the residual limb to partly define the desired prosthesis knee and ankle angles (3), (5). The acceleration dependency initiates the prosthesis knee flexion and ankle plantarflexion even if the residual limb is not rotating, such as at the beginning of swing (Fig. 5). By continuously updating the prosthesis joint angles as a function of the residual limb orientation, velocity, and acceleration, the proposed swing controller provides adequate toe clearance for different cadences (0.76 to 1.80 seconds), stair heights (4-7 inches), and gait patterns (step-over-step, step-by-step, two-steps).

Powered prostheses can assist individuals with an above-knee amputation during stair ascent by actively generating torque as necessary to lift the user up the step. Climbing stairs with variations in stair heights or gait patterns requires variations in levels of torque generation and mechanical energy injection [12]. The amount of energy injected by the prosthesis must increase proportionally to the stair height, such that more energy is provided for taller stairs compared to shorter stairs. Moreover, the location of the center of mass when the prosthesis generates torque is critical. During prosthetic torque generation, the location of the individual's center of mass must be above the powered prosthesis so that the prosthesis torque can push the body mass upward. If the user's center of mass is not above the prosthesis, then the torque generated by the prosthesis can push the individual backward, throwing them off balance or causing them to fall. Thus, controllers that use a fixed, pre-programmed stance torque profile or joint impedance cannot properly assist amputee users to ascend stairs in the real-world scenarios where every staircase is different. To address these limitations, we proposed adapting the prosthesis stance torque profile at each step.

The proposed stance controller changes the knee torque-angle relationship based on the prosthesis position. The adaptive knee torque-angle relationship (Fig. 2) enables the powered prosthesis to achieve multiple desired outcomes. The first outcome is the modulation of the net energy injected by the prosthesis as a function of the stair height. This outcome is achieved by increasing the maximum knee torque parameter in the torque-angle relationship proportionally to the knee flexion angle at the beginning of stance (Fig. 2). Because the knee flexion angle at the beginning of stance is proportional to the residual limb orientation and also to the stair height (Fig. 6), the taller the step, the greater the peak torque. As a result, the energy injected by the prosthesis is proportional to the stair height (Fig. 6). The second outcome of the proposed controller is that the torque at the very beginning of stance is always zero, regardless of the stair height (Fig. 6). This adaptation is important because if the torque at the beginning of stance, when the foot first contacts the step, is not zero, then the powered

prosthesis will prematurely apply torque and push the user backward, which may result in a fall and injury. The third outcome of the adaptive torque-angle relationship is the independence of energy injection from time. Inspired by biological leg behavior, the proposed controller sets the peak torque knee angle proportional to the knee range of motion to achieve full extension (Fig. 2). As a result, the torque-angle relationship is scaled linearly on the knee range of movement, and the knee angle at peak torque changes depending on the stair height and gait pattern. As the user moves through the extension of the prosthesis, the proportional knee torque is applied. This independence with respect to time allows the user to move at their desired cadence while still receiving the assistance needed. While the knee function is defined by the adaptive torque-angle relationship, the ankle movements are synchronized to the knee movements. Using a dedicated adaptive function (6), the ankle angle gradually returns to neutral as the knee extends during stance, regardless of the ankle angle at the beginning of stance (Fig. 7). By modulating the knee torque generated by the prosthesis at each step, the proposed stance controller enables the powered prosthesis to imitate the energy injected by the biological knee for different stair heights and different gait patterns.

The proposed controller allows for adaptability to cadence, gait pattern, and stair height with a few limitations. Non-amputee biomechanics show the ankle joint provides considerable energy during late stance. Our controller is not capable to imitate this important physiological function. Powered ankle push-off has been obtained in a previous study using a finite-state machine with a dedicated late-stance state signaled by an intermediate threshold on the ground reaction force [19], [22]. In our pilot studies, this late-stance state caused the ankle to push-off when the users shuffled instead of climbing stairs. This observation suggests that integrating ankle push-off in our stair ascent controller will require an understanding of the sound side position to ensure that the user wants to climb a step rather than shuffling. Future work should focus to address this limitation so that the proposed controller can more closely imitate the physiological ankle push-off. The analysis of the biological ankle biomechanics also suggests that at the transition from swing to stance the body weight is loaded on the middle to the front portion of the foot while the ankle joint is slightly plantarflexed [13]. In contrast, our controller aims for a flat foot placement at this transition to minimize the amount of joint movement occurring underneath the user. Further experiments are necessary to understand the impact of this discrepancy between the biological ankle behavior and that of our controller on the user.

Because the prosthesis movements are intrinsically synchronized to the user's residual limb movements, the proposed *Lifting* controller does not need to explicitly classify the intent of the user to ascend stairs. In other terms, the user can safely stand and shuffle around without triggering a predefined swing movement like in previous controllers. Removing the need for accurate user's intent classification may reduce the probability of the powered leg misbehaving while reducing the tuning and set-up time. Further work is necessary to quantify the impact of the proposed controller on the user's intent classification problem and, potentially, to integrate the proposed stair controller with walking controllers in a way that

does not require explicit classification of the user's intent to climb stairs or walk. This goal may be achieved in the future by using an adaptive minimum jerk controller as done previously in a walking/crossing obstacle controller [35].

V. CONCLUSION

The ability to enable climbing stairs step-over-step is one of the biggest advantages of powered prostheses over conventional passive devices. Safely climbing stairs in the real world requires the powered prosthesis controller to adapt to the variability in stair geometry, gait pattern, and gait cadence. The proposed *Lifting* controller enables a powered prosthesis to adapt the swing trajectory continuously, while coordinating with the movements of the user, without explicit classification of the environment. The proposed *Standing* controller enables a powered prosthesis to adapt the timing and amount of energy injected in stance based on the changing user's needs. This experiment with one individual with an above-knee amputation shows that the proposed stair controller supports climbing stairs with different heights at the preferred cadence and using different gait patterns. A controller with this capability may improve real-world ambulation and is not currently available to prosthesis users. Future work will be dedicated to evaluating the performance of the proposed controller in a larger clinical population and to assess functional mobility compared to passive prostheses and powered prostheses using other control approaches.

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REFERENCES

- [1] A. D. Segal *et al.*, "Kinematic and kinetic comparisons of transfemoral amputee gait using C-Leg and Mauch SNS prosthetic knees," *J. Rehabil. Res. Dev.*, vol. 43, no. 7, p. 857, 2006.
- [2] D. Winter, "Energy generation and absorption at the ankle and knee during fast, natural, and slow cadences," *Clin. Orthop. Relat. Res.*, vol. 175, pp. 147–154, 1983.
- [3] T. S. Bae, K. Choi, D. Hong, and M. Mun, "Dynamic analysis of above-knee amputee gait," *Clin. Biomech.*, vol. 22, no. 5, pp. 557–566, 2007.
- [4] D. J. Lura, M. W. Wernke, S. L. Carey, J. T. Kahle, R. M. Miro, and M. J. Highsmith, "Crossover study of amputee stair ascent and descent biomechanics using Genium and C-Leg prostheses with comparison to non-amputee control," *Gait Posture*, vol. 58, no. July, pp. 103–107, 2017.
- [5] H. Hobara *et al.*, "Lower extremity joint kinematics of stair ascent in transfemoral amputees," *Prosthet. Orthot. Int.*, vol. 35, no. 4, pp. 467–472, 2011.
- [6] B. J. Hafner, L. L. Willingham, N. C. Buell, K. J. Allyn, and D. G. Smith, "Evaluation of Function, Performance, and Preference as Transfemoral Amputees Transition From Mechanical to Microprocessor Control of the Prosthetic Knee," *Arch. Phys. Med. Rehabil.*, vol. 88, no. 2, pp. 207–217, 2007.

- [7] C. Gauthier-Gagnon, M.-C. Grise, and D. Potvin, "Enabling factors related to prosthetic use by people with transtibial and transfemoral amputation," *Arch. Phys. Med. Rehabil.*, vol. 80, no. 6, pp. 706–713, 1999.
- [8] H. Hobara, Y. Kobayashi, T. Nakamura, N. Yamasaki, and T. Ogata, "Foot clearance strategy for step-over-step Stair climbing in transfemoral amputees," *Prosthet. Orthot. Int.*, vol. 38, no. 4, pp. 332–335, 2014.
- [9] S. M. Jaegers, J. H. Arendzen, and H. J. de Jongh, "Changes in hip muscles after above-knee amputation," *Clin Orthop Relat Res*, no. 319, pp. 276–284, 1995.
- [10] Ö. Ülger, S. Topuz, K. Bayramlar, F. Erbahçeci, and G. ener, "Risk factors, frequency, and causes of falling in geriatric persons who has had a limb removed by amputation," *Top. Geriatr. Rehabil.*, vol. 26, no. 2, pp. 156–163, 2010.
- [11] M. Goldfarb, B. E. Lawson, and A. H. Shultz, "Realizing the promise of robotic leg prostheses," *Sci. Transl. Med.*, vol. 5, no. 210, pp. 1–5, 2013.
- [12] R. Riener, M. Rabuffetti, and C. Frigo, "Stair Ascent And Descent At Different Inclinations," *Gait Posture*, vol. 15, no. 1, pp. 32–44, Feb. 2002.
- [13] B. J. McFadyen and D. A. Winter, "An integrated biomechanical analysis of normal stair ascent and descent," *J. Biomech.*, vol. 21, no. 9, pp. 733–744, Jan. 1988.
- [14] C. D. Hoover, G. D. Fulk, and K. B. Fite, "Stair ascent with a powered transfemoral prosthesis under direct myoelectric control," *IEEE/ASME Trans. Mechatronics*, vol. 18, no. 3, pp. 1191–1200, 2013.
- [15] B. E. Lawson, H. A. Varol, A. Huff, E. Erdemir, and M. Goldfarb, "Control of stair ascent and descent with a powered transfemoral prosthesis," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 21, no. 3, pp. 466–473, Oct. 2013.
- [16] T. Lenzi, M. Cempini, L. Hargrove, and T. Kuiken, "Design, development, and testing of a lightweight hybrid robotic knee prosthesis," *Int. J. Rob. Res.*, vol. 37, no. 8, pp. 953–976, Jul. 2018.
- [17] M. Tran, L. Gabert, M. Cempini, and T. Lenzi, "A Lightweight, Efficient Fully Powered Knee Prosthesis with Actively Variable Transmission," *IEEE Robot. Autom. Lett.*, vol. 4, no. 2, pp. 1186–1193, 2019.
- [18] A. M. Simon *et al.*, "Configuring a powered knee and ankle prosthesis for transfemoral amputees within five specific ambulation modes," *PLoS One*, vol. 9, no. 6, 2014.
- [19] E. D. Ledoux and M. Goldfarb, "Control and Evaluation of a Powered Transfemoral Prosthesis for Stair Ascent," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 25, no. 7, pp. 917–924, 2017.
- [20] A. F. Azocar, L. M. Mooney, L. J. Hargrove, and E. J. Rouse, "Design and Characterization of an Open-Source Robotic Leg Prosthesis," *Proc. IEEE RAS EMBS Int. Conf. Biomed. Robot. Biomechatronics*, vol. 2018-August, pp. 111–118, 2018.
- [21] B. E. Lawson, J. Mitchell, D. Truex, A. Shultz, E. Ledoux, and M. Goldfarb, "A robotic leg prosthesis: Design, control, and implementation," *IEEE Robot. Autom. Mag.*, vol. 21, no. 4, pp. 70–81, 2014.
- [22] S. Culver, H. Bartlett, A. Shultz, and M. Goldfarb, "A stair ascent and descent controller for a powered ankle prosthesis," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 26, no. 5, pp. 993–1002, 2018.
- [23] A. M. Simon, N. P. Fey, S. B. Finucane, R. D. Lipschutz, and L. J. Hargrove, "Strategies to reduce the configuration time for a powered knee and ankle prosthesis across multiple ambulation modes," *IEEE Int. Conf. Rehabil. Robot.*, vol. 2013, pp. 1–6, Jun. 2013.
- [24] S. Rezazadeh, D. Quintero, N. Divekar, E. Reznick, L. Gray, and R. D. Gregg, "A Phase Variable Approach for Improved Rhythmic and Non-Rhythmic Control of a Powered Knee-Ankle Prosthesis," *IEEE Access*, vol. 7, pp. 109840–109855, 2019.
- [25] D. Quintero, D. J. Villarreal, D. J. Lambert, S. Kapp, and R. D. Gregg, "Continuous-Phase Control of a Powered Knee & Ankle Prosthesis: Amputee Experiments Across Speeds and Inclines," *IEEE Trans. Robot.*, pp. 1–16, 2018.
- [26] T. Lenzi, L. Hargrove, and J. W. Sensinger, "Preliminary evaluation of a new control approach to achieve speed adaptation in robotic transfemoral prostheses," in *2014 IEEE/RSJ International Conference on Intelligent Robots and Systems*, 2014, pp. 2049–2054.
- [27] T. Lenzi, L. Hargrove, and J. Sensinger, "Speed-Adaptation Mechanism: Robotic Prostheses Can Actively Regulate Joint Torque," *IEEE Robot. Autom. Mag.*, vol. 21, no. 4, pp. 94–107, Dec. 2014.
- [28] G. Aguirre-Ollinger, J. E. Colgate, M. A. Peshkin, and A. Goswami, "Design of an active one-degree-of-freedom lower-limb exoskeleton with inertia compensation," *Int. J. Rob. Res.*, vol. 30, no. 4, pp. 486–499, 2011.
- [29] L. Gabert, S. Hood, M. Tran, M. Cempini, and T. Lenzi, "A Compact, Lightweight Robotic Ankle-Foot Prosthesis: Featuring a Powered Polycentric Design," *IEEE Robot. Autom. Mag.*, vol. 27, no. 1, pp. 87–102, Mar. 2020.
- [30] J.-T. Zhang, A. C. Novak, B. Brouwer, and Q. Li, "Concurrent validation of Xsens MVN measurement of lower limb joint angular kinematics," *Physiol. Meas.*, vol. 34, no. 8, pp. N63–N69, Aug. 2013.
- [31] "504 Stairways." [Online]. Available: <https://www.ada-compliance.com/ada-compliance/504-stairways>.
- [32] M. Cempini, L. J. Hargrove, and T. Lenzi, "Design, development, and bench-top testing of a powered polycentric ankle prosthesis," *IEEE Int. Conf. Intell. Robot. Syst.*, vol. 2017-Sept, no. September, pp. 1064–1069, 2017.
- [33] T. Lenzi, M. Cempini, L. J. Hargrove, and T. A. Kuiken, "Actively variable transmission for robotic knee prostheses," *Proc. - IEEE Int. Conf. Robot. Autom.*, pp. 6665–6671, 2017.
- [34] L. Gabert and T. Lenzi, "Instrumented Pyramid Adapter for Amputee Gait Analysis and Powered Prosthesis Control," *IEEE Sens. J.*, vol. 19, no. 18, pp. 8272–8282, 2019.
- [35] J. Mendez, S. Hood, A. Gunnell, and T. Lenzi, "Powered knee and ankle prosthesis with indirect volitional swing control enables level-ground walking and crossing over obstacles," *Sci. Robot.*, vol. 5, no. 44, pp. 1–11, 2020.