A Phase Variable Approach for Improved Volitional and Rhythmic Control of a Powered Knee-Ankle Prosthesis

Siavash Rezazadeh1, David Quintero1,2, Nikhil Divekar1, Emma Reznick1, Leslie Gray3 and Robert D. Gregg1,2

Abstract—Although there has been recent progress in control of multi-joint prosthetic legs for periodic tasks such as walking, volitional control of these systems for non-periodic maneuvers is still an open problem. In this paper, we develop a new controller that is capable of both periodic walking and common volitional leg motions based on a piecewise holonomic phase variable through a finite state machine. The phase variable is constructed by measuring the thigh angle, and the transitions in the finite state machine are formulated through sensing foot contact along with attributes of a nominal reference gait trajectory. The controller was implemented on a powered knee-ankle prosthesis and tested with a transfemoral amputee subject, who successfully performed a wide range of periodic and non-periodic tasks, including low- and high-speed walking, quick start and stop, backward walking, walking over obstacles, and kicking a soccer ball. Use of the powered leg resulted in significant reductions in amputee compensations including vaulting and hip circumduction when compared to use of the take-home passive leg. The proposed approach is expected to provide better understanding of volitional motions and lead to more reliable control of multi-joint prostheses for a wider range of tasks.

Index Terms—powered prostheses, transfemoral amputees, rehabilitation robotics, volitional control.

I. INTRODUCTION

THE vast majority of lower-limb amputees use mechanically passive prosthetic legs, which can only dissipate energy during locomotion. This limits the ability of amputees to efficiently perform various ambulation modes, particularly walking at variable speeds or slopes. Furthermore, the biomechanical compensations required to walk with these passive devices generally cause joint discomfort and back pain during daily usage [1]. Powered prosthetic legs that provide actuation at the joints have the potential to improve amputee gait and eliminate these problems [2]–[6]. However, they require sophisticated control strategies, especially for multi-joint legs, in order to perform various activities in a natural and safe manner [7].

From a biomechanics perspective, the human gait cycle can be divided into different phases (namely, stance and swing phase) and sub-phases (for example, weight acceptance, push-off, early swing, etc.), each serving a specific purpose in locomotion [8]. This perspective was preserved in control design for powered lower-limb prostheses, which involves first detecting the correct sub-phase and then controlling that particular behavior of the prosthetic joints [5], [9]–[12]. The tuning has to be performed separately for each individual based on various physical parameters, for example, body mass, as well as functional parameters, for example, gait pattern. Due to the large number of parameters that need to be manually tuned, the process is typically arduous and difficult to automate, often taking multiple hours for each subject [10].

To address these issues, recent approaches have parameterized the gait cycle over a phase variable, i.e., a monotonic signal that represents the progression through the cycle. Aside from parameterizing the gait, ideally the phase variable is invariant across different subjects and does not depend on parameters such as the person’s mass or height [13]. In [14], the heel-to-toe movement of the Center of Pressure (CoP) served as the phase variable for determining progression through the stance phase, whereas the swing phase was controlled by two impedance-based states. In [15], the authors investigated additional phase variables for locomotion and found that the global thigh angle is a suitable piecewise monotonic signal that can be used to control the stance and swing phases separately. By also using the integral of the global thigh angle, the phase variable was made continuous across the gait cycle and was implemented in a powered knee-ankle prosthesis for use by amputee subjects [6].

Everyday tasks comprise both rhythmic activities, such as walking, as well as non-rhythmic activities, such as stepping over obstacles. A controller strictly based on behavior in a rhythmic task, such as the unified controller presented in [6], will encounter problems for non-rhythmic volitional motions. Previous studies such as [16]–[18] attempted to enhance volitional control using information contained in the bioelectric signals of the residual limb, such as those acquired from electromyography (EMG). However, EMG quality is highly dependent on physical factors such as electrode placement, movement artifact, and electromagnetic noise; physiological factors such as muscle and nerve fatigue; and anatomical factors such as volume conduction, which causes “mixing”.
of signals from different underlying muscles when using surface electrodes. In the case of transfemoral amputees, the muscles used for controlling the ankle joint do not exist. The information may still be recovered if a nerve re-innervation procedure was carried out during amputation, but sophisticated “unmixing” algorithms have to be used to decipher individual muscle activity from the EMG [19]. In light of this knowledge, we seek a more reliable solution using only mechanical measurements.

As a first attempt for such a control scheme, Villarreal et al. used the thigh angle and a stance/swing detection switch to implement a piecewise phase variable for volitional control [20]. However, the controller was problematic during transitions, as using solely the foot contact condition for transitioning between stance and swing phase variables would result in jumps and oscillations. To avoid such jumps, pushoff was eliminated, but it consequently made walking at greater speeds difficult and inefficient. Moreover, the undesired jumps would still occur when standing, as the subject shifted their weight to the sound leg. This motivated the design of a volitional controller based on a Finite State Machine (FSM) and preliminary experiments with an amputee subject demonstrated its functionality for different volitional and rhythmic tasks [21].

In this paper, we extend this investigation by comparing the performance of the powered leg using the proposed controller with that of a passive transfemoral prosthesis. Such comparison has been done in works such as [22] and [23] for powered ankle and powered knee-ankle prostheses, respectively. However, both these works rely on several sessions of training in order to obtain meaningful improvements with the powered prostheses. Specifically, in [22], it was shown that without training, the outcomes may only have slight improvements. In contrast, we show that using the presented controller, significant improvements can be observed immediately after a brief tuning/acclimation session (about 10 minutes) with the powered leg. In [23], the positive effects of a powered leg on back muscles and gait energy expenditure were investigated. Such symptoms often originate from compensations that amputees adopt to overcome the lack of power in passive prostheses [24]–[27]. In this work, we study the kinematic and kinetic attributes characterizing these compensations. The most common compensations associated with the use of passive prostheses are related to prosthetic foot clearance resulting from under-powered plantarflexion during pushoff and knee flexion during swing [28], [29]. Since individual amputees develop different compensatory mechanisms, we focused our analyses on traits that were the most apparent with the amputee participant, namely, vaulting and hip circumbduction. Vaulting emerges as excessive plantarflexion during midstance of the sound leg, whereas hip circumbduction is characterized by excessive hip abduction of the prosthetic side during swing, resulting in lateral deviation of the prosthetic foot. We hypothesized our powered leg would help mitigate these compensatory mechanisms by providing active plantarflexion and knee flexion to better aid the foot clearance.

A direct outcome of the phase-based controller is its ability to enable backward walking. For backward walking, two different gait patterns are recognizable. In a “step-to” pattern, the sound leg leads the motion while the prosthetic leg follows, seldom passing beyond the sound leg. In contrast, in a “step-through” pattern, both sound and prosthetic legs alternately lead the motion and the swing leg passes the stance leg. We analyze these gait patterns during backward walking trials to determine how the powered leg performs compared to the passive leg.

We also tested common volitional tasks that amputees may face in daily life. The first task we tested was stepping over an obstacle. We predicted the subject would be able to successfully maneuver his intact joints into a configuration that maintains the prosthetic knee and ankle in a sufficiently flexed position while he guiding his hip joint over the obstacle. This kind of maneuver is understandably difficult or even impossible with passive prostheses or the state-of-the-art powered prostheses designed for walking. Another volitional task that we tested was kicking a soccer ball. Kicking is a non-rhythmic movement that requires powerful but controlled extension of the knee. Functional abilities are compared between the powered and passive legs during these tasks.

The paper is organized as follows. In Section II, we describe the new phase variable, its states and corresponding transitions, and the implementation of the controller. Section III addresses the experimental setup, protocol, and methods. Next, in Section IV, the results of the experiments are presented. Based on that, in Section V, the obtained results are discussed and the performance of the controller for different tasks is analyzed. At the end, we present a series of conclusions together with suggestions for future works.

II. CONTROL DESIGN

This section presents the design of the proposed scheme for volitional and periodic control of a powered knee-ankle prosthesis. First, we explain the use of virtual constraints for formulating the desired knee and ankle joint trajectories. Next, we describe the design of our proposed phase variable for parameterizing the virtual constraints in different stages of the gait cycle. Finally, we discuss how the controller is implemented on a powered prosthesis.

A. Virtual Constraints

Virtual constraints, as introduced in [30], [31], are a useful tool to represent time-invariant trajectories, which can considerably simplify the process of controlling periodic orbits. Originally, virtual constraints were introduced as relationships among generalized positions (angles), which is analogous to a holonomic set of kinematic constraints. More recently, nonholonomic virtual constraints have also been used in legged robots applications [32], [33]. Generally, virtual constraints define the desired trajectories for the controlled degrees of freedom in the following form:

$$q^d_i(s) = h(s),$$

where $s$ is a monotonic function of positions (for holonomic virtual constraints), or positions and velocities (for nonholonomic virtual constraints), and is usually scaled between 0 and 1.
In legged robot applications, \( s \) is normally reset every step, and continuity is preserved by imposing equality conditions on \( h \) and \( \partial h/\partial q \) at \( s = 0 \) and \( s = 1 \). This is a convenient choice for legged robots, especially considering there are sensors on both legs for computing the phase variable. For a prosthetic leg application, in order to avoid attaching sensors on the sound leg of the subject, it is desirable to use only onboard sensors from the prosthesis. This is equivalent to resetting the phase variable at the end of each stride, rather than each step. In this case, \( \text{Fig. } 1 \) represents the desired, periodic trajectories for the entire stride \[34\].

Due to their dependence on velocities or integrals, nonholonomic virtual constraints are sensitive to changes in speed and are thus not suitable for a controller that is intended to work in a wide range of non-steady activities. A good example is the integral-based unified controller presented in \[6\], which worked well in normal-speed steady-state walking, but was unreliable for slow speeds and was unable to perform non-rhythmic motions. Therefore, we establish our volitional control scheme on a holonomic phase variable in order to make it speed-independent.

In \[13\], \[35\], Villarreal et al. used a perturbation experimental setup to examine and compare various combinations of thigh angle, its derivative, and its integral as invariant parameterizations of human walking gaits. Although the best phase variable found in this study was nonholonomic, a holonomic parameterization was a close second choice. Motivated by this result, and since the holonomicity of the thigh angle makes it an ideal selection for a volitional controller, we use this variable as the basis for our volitional controller. In what follows we show how this angle is used to construct our phase variable.

### B. Constructing the Volitional Phase Variable

As mentioned before, we aim to use thigh angle \( q_h \) (\text{Fig. } 1(b)), for defining our holonomic phase variable. In what follows, we will show how this variable can be used for this purpose and what other measurements are necessary.

\text{Fig. } 1(a) \text{ depicts the thigh, knee, and ankle angle trajectories during one stride of a normal able-bodied walking gait } \[8\]. \text{ Note that the thigh angle is not a monotonic signal throughout the stride. As a result, each value of } q_h \text{ corresponds to at least two points in the cycle (one in the descending part of } q_h \text{ and one in the ascending part), making the determination of a unique } s \text{ based solely on } q_h \text{ impossible. To avoid this problem, and also to keep the benefits of a holonomic system, we propose to use a set of piecwise holonomic virtual constraints. The idea is to divide the gait cycle into different sections, where each section corresponds to a monotonic (either ascending or descending) thigh angle trajectory.}

From \text{Fig. } 1(a), \text{ the thigh angle trajectory during a stride can roughly be divided into two monotonic sections (neglecting the small retraction section at the end); it is descending after heel strike (} t/T = 0 \text{) and through the stance phase until the trajectory reaches its minimum at } t/T = 0.53, \text{ and then becomes ascending. Note that the swing phase starts a little later, at } t/T = 0.63. \text{ An obvious way to transitioning between these two states is using the sign change of the thigh angle's rate, } \dot{q}_h. \text{ In practice this proves to be a very sensitive signal, because velocities can change rapidly, which results in large discontinuities in the virtual constraints and in undesirable transitions. For this reason and since these two monotonic sections approximately correspond to stance and swing phases, in } \[20\] \text{ a foot contact sensor was used for transitioning between these two states. The first problem with this approach is that the minimum thigh angle does not exactly correspond to the foot takeoff (} t/T = 0.53 \text{ versus } t/T = 0.63) \text{ and thus part of pushoff will be performed when the leg is already in swing. Moreover, this approach assumes that the thigh angle exactly follows the reference trajectory. If the minimum thigh angle is larger than the reference trajectory's minimum (shorter step), there will be a jump in the virtual constraints. Conversely, if the minimum thigh angle is less than the reference (longer step), the virtual constraint will saturate, which leaves pushoff half-completed. These undesirable features can be seen in the results of } \[20\].

\text{To resolve these problems, we propose to have two supplementary states (in addition to stance and swing) to represent pushoff. The result is depicted in } \text{Fig. } 2 \text{ in the form of an FSM with four states, where S1 and S2 pertain to the descending part of the thigh trajectory, and S3 and S4 correspond to the ascending part. Note that S1, S2, and S3 are all parts of the stance phase, and thus for all of these states } FC = 1 \text{ (} FC \text{ represents foot contact as a binary signal). For this reason, we use other variables to define these transition conditions. Namely, transitioning from S1 (stance) to S2 (pushoff onset) occurs at a specific thigh angle (} q_h = q_{po}, \text{ and transitioning from S2 to S3 (pre-swing) occurs when } q_h = 0. \text{ The tunable constant } q_{po} \text{ represents the thigh angle at the start of pushoff and its default value is obtained from the thigh angle at the maximum ankle angle in the reference trajectory (} q_{po} = -8.4^\circ). \text{ As previously mentioned, a transition based on velocity is accompanied with the risk of sensitivity and sudden jumps in virtual constraints. Although these jumps would be small due to the small range of thigh angles represented by S2 and S3, we propose a two-step approach to completely eliminate such discontinuities. In the first step, the transitions from S1 to S2 and from S2 to S3 are designed to be unidirectional, resulting in only one possible jump from S2 to S3. To eliminate this single jump, in the second step we reset the associated parameters based on the information from the sensors. This will be explained in the definition of } s \text{ in what follows.}

For S1 and S2, the phase variable can be computed from a shift and scale of the thigh angle:

\[
s = \frac{q_h^0 - q_h}{q_h^0 - q_h^{\text{min}}} \cdot c, \tag{2}
\]

where \( q_h^0 \) and \( q_h^{\text{min}} \) are constants whose default values are touchdown value and the minimum of the reference thigh angle trajectory, respectively. These two parameters can be tuned if the subject prefers a different step length. The constant \( c \) is also tunable and is related to the ratio of the stance phase to the whole cycle. The default value of \( c \) is the normalized time at which \( q_h \) reaches its minimum, which is 0.53 in \text{Fig. } 1(a).
Since the transitioning from S2 to S3 is based on the change of sign of $\dot{q}_h$, S3 pertains to the ascending part of the thigh angle. To form a continuous phase variable and to avoid jumps at each transitioning from S2 to S3, we record the values for $s$ and $q_h$ and name them $s_m$ and $q_{h,m}$, respectively. The phase variable in preswing (S3) and swing (S4) phases is then computed from

$$s = 1 + \frac{1 - s_m}{q_h^0 - q_{h,m}} \cdot (q_h - q_h^0). \quad (3)$$

Note that $s = s_m$ at $q_h = q_{h,m}$, and $s = 1$ at $q_h = q_h^0$. For both (2) and (3), the phase variable is saturated between 0 and 1.

An additional factor to consider for the preswing phase is the tendency of the leg to oscillate as the load is removed from it. This is eliminated by imposing a unidirectional filter on the phase variable in S3. That is, in the discrete time instance $k$:

$$s(k) \geq s(k-1), \quad \text{when } S = S3, \quad (4)$$

where $S$ is the current state. Note that this condition is not required for S2, as it transitions to S3 at the first instance when $\dot{q}_h$ (and hence $\dot{s}$) crosses zero.

The FSM of Fig. 2 together with the phase variable definition in (2) and (3) constitute a control paradigm based on a forward walking scheme. However, a volitional controller needs to also manage situations in which the motion is interrupted or even reversed. Due to the holonomic nature of the designed phase variable, it is invariant to the direction of motion. Therefore, the problem can only arise during transitions. The most critical situation happens when the leg is in swing and it touches the ground behind the body (backward walking). According to Fig. 2 the state transitions to S1 and then immediately to S2 (and perhaps S3), which leads to pushoff and does not allow the subject to put weight on the leg. In order to avoid this, we added another state, S5, to the FSM (Fig. 3). This new state keeps the leg in stance phase when walking backward, and it transitions to pushoff only if the subject resumes moving forward. With this new state, we define the transitions for our volitional controller as follows:

1) Transition from S1 or S5 to S4: The primary condition for transitioning between stance and swing is foot contact. However, in conditions such as standing still, if the leg is unloaded for a moment (i.e., shifting weight to the
sound leg), a transition to swing can result in a sudden and undesirable flexion of the knee. To avoid this, we require that the transition to swing happens either after pushoff (i.e., through S2 and S3), or directly from S1 or S5 to S4 at maximum thigh angle ($q_h = 0$). Obviously, for transitioning from S5 to pushoff, the state first needs to go to S1, as discussed next.

2) Transition from S5 to S1: This transition happens when the subject steps backward and then decides to move forward. The transition condition is given by $q_h < q_{h1}$, where $q_{h1}$ is a tunable constant. Note that $q_{h1} < q_{h2}$ in order to avoid direct transitioning to pushoff.

3) Transition from S4 to S1 or S5: Since stance is a more reliable state for the subjects (they can put their weight on the leg), the condition for transitioning from S4 to S1 or S5 is less strict compared to S1 and S5 to S4. When foot contact happens ($FC = 1$), the transition will be to S1 if $q_h \geq q_{h1}$, otherwise it will be to S5, where $q_{h1}$ is a tunable constant. Setting $q_{h1}$ to zero is equivalent to transition from S4 to S1 for a forward step or to S5 for a backward step.

Fig. 3 summarizes the states and the corresponding transitions.

C. Control Design Based on Virtual Constraints

Having computed the phase variable, the next step is obtaining the virtual constraints. In this work, we follow the approach of [6] in using Discrete Fourier Transform (DFT) in order to generate virtual constraints for knee and ankle, based on data from normal human walking provided in [8]. In this form the desired joint angles can be computed as

$$
q_i^d = h(s) = \frac{1}{2} \rho_0 + \frac{1}{2} \rho_{N/2} \cos(\pi N s)
+ \sum_{k=1}^{N/2-1} \left[ \rho_k \cos(\Omega_k s) - \psi_k \sin(\Omega_k s) \right],
$$

where $\rho_k$ and $\psi_k$ are the coefficients of real and imaginary parts of DFT, respectively.

In the next step, the desired knee and ankle angle obtained from (5), are imposed using a Proportional-Derivative (PD) position controller. Noting that the position error for joint $i$ is

$$
e_i = h(s) - q_i,
$$

the commanded motor torque is obtained from

$$
t_i = K_{p,i} e_i + K_{d,i} \dot{e}_i,
$$

where $K_{p,i} > 0$ and $K_{d,i} > 0$ are PD control gains for joint $i$.

Fig. 4(a) displays the block diagram of the proposed controller.

III. EXPERIMENTAL METHODS

A. Materials

The knee-ankle powered leg used for our experiments is shown in Fig. 4(b). Each joint is equipped with a Maxon EC-3pole 30, 200 Watt, three-phase brushless DC motor driving the joints through a timing belt and a Nook 2-mm lead ball screw. Due to greater torques in the ankle joint, the timing belt ratio for the ankle is twice that for the knee.

The joints and motors are equipped with optical encoders (Maxon 2RMHF for motors and US Digital EC35 for joints). An IMU sensor (LORD MicroStrain, 3DMGX4-25) is used to measure the global thigh angle, as shown in Fig. 4(b). Foot contact condition is determined using a force sensitive resistor sensor (FSR - FlexiForce A401, Tekscan Inc.) located inside the pyramid adapter of the prosthetic foot.

The computation and control is done offboard via a tethered connection to a dSPACE DS1007 system with Freescale OorIQ P5020, dual-core, 2 GHz PowerPC processor. The commanded torques from the computer are sent to an Elmo Gold Twitter R80/80 amplifier, which controls the motors. The powers for the motors and the FSR are provided through DC power supplies (Agilent Technologies 6673A for the motors and BK Precision 1761 for the FSR). See [6] for further details on the design of the prosthetic leg and the hardware specifications.

Motion capture data was collected using a 10-camera VICON system. A 16-marker lower body model was used (Plug-in Gait - VICON). Standard data processing procedures, including gap filling and filtering (fourth-order 6 Hz cut-off, 300 Hz cut-off Lowpass Butterworth for trajectory data, fourth-order 300 Hz cut-off Lowpass Butterworth for Force Plate data), were applied. A walking platform with a single 3-axial force plate (Kistler) positioned midway of the walkway length was constructed. Handrails were provided alongside the entire length of the platform. A non-instrumented treadmill with a safety harness was used for collection of continuous gait data.

B. Procedure

The controller was first tested and tuned through a set of walking experiments on a treadmill with an able-bodied subject using the powered leg with a bypass system [36]. Any tunable parameter discussed in Section III was appropriately changed until the subject was able to comfortably walk in different speeds. The joint trajectories for the trials with three different speeds (slow: 0.7 m/s; normal: 0.9 m/s; and fast: 1.1 m/s) were recorded for qualitative comparison with those of the amputee subject. After this set of tests, the main experiments were conducted with an amputee subject with the same set of parameters and with no change. The amputee participant was a 32-year old male with a height of 1.75 m, weight of 76 kg with the passive leg, and weight of 78.9 kg with the powered leg. His left leg was amputated 11 years before these experiments and he used an Ottobock 3R60 knee together with an Ability Rush ankle-foot prosthesis as his everyday passive leg. The participant had no prior experience with the powered prostheses device being tested, and he had no neuro-muscular disorder that is known to affect gait, balance, or muscle activity. The experimental protocol was reviewed and approved by the Institutional Review Board (IRB) at the University of Texas at Dallas.

Prior to the experiment with the amputee subject, standard procedures for fitting and tuning of the prosthetic leg were carried out by a licensed Prosthetist. For fitting the powered prostheses, the participant’s daily-use socket was used. The
finite state machine virtual constraints

\[ q_k, q_a \]

knee PD control
prosthetic leg

\[ q_k^d, q_a^d \]

\[ \varepsilon_k, \varepsilon_a \]

\[ \tau_k, \tau_a \]

(a)

Fig. 4. (a) Block diagram of the proposed volitional control for the knee-ankle prosthesis. \( q_k \) and \( q_a \) represent knee and ankle joint angles, respectively, and \( q_k^d \) and \( q_a^d \) are their desired values. (b) The powered knee-ankle prosthetic leg worn by the transfemoral amputee participant.

experiments were designed in two different sets, namely non-rhythmic and rhythmic. The non-rhythmic experiments were done on the walkway platform, while the rhythmic experiments were conducted on the treadmill. The subject first performed both sets of tasks with his passive prosthesis, and then he repeated them with the powered leg. As mentioned, for all these tasks, the control parameters introduced in Section II remained the same.

Walking forward between the handrails was repeated until we captured three clean foot strikes on the force plate for each leg (prosthetic, sound). Next, we examined the invariance of the controller to the direction of locomotion and the ability of reversing the direction of walking at the subject’s will. For this purpose, we asked the participant to walk backward on the walkway, as well as a combination of forward and backward transitions.

After that, we tested the ability of the subject to step over an obstacle, specifically an 85-mm high wooden block. The goal was to step over it first using the sound leg and then the prosthesis. No restrictions were placed on the number of practice trials. No additional guidance was provided regarding the ideal foot placement or maneuvering strategy for crossing the obstacle; the participant was encouraged to explore and practice his preferred strategy. After this experiment, the subject was asked to kick a soccer ball to demonstrate the fast extension of the powered knee following a quick forward motion of the hip in an activity other than walking. This test concluded the overground experiments.

Treadmill trials for rhythmic tasks were performed at three speeds: slow, normal, and fast. First, the participant was instructed to adjust the treadmill speed to his self selected speed (i.e., normal speed) with his passive prosthesis. The slow and fast speeds were then taken as the minimum and maximum speeds, respectively, at which the participant was able to maintain a stable and comfortable gait with his passive leg. Data were captured for 60, 60, and 45 seconds for slow, normal, and fast speeds, respectively, where recording was only started after the participant was able to produce a reasonably consistent gait cycle (as visually inspected). With the powered leg, the subject was able to reach higher speeds than the maximum speed with the passive leg. Thus, in addition to the previous speeds, we tested and recorded the maximum speed at which the subject was able to comfortably walk with the powered leg for the duration of 30 seconds.

C. Data Analysis

For comparing the joint angle trajectories of the prosthetic legs with those of the reference able-bodied subjects (adopted from [8]), we computed the Pearson correlation coefficient for each case, as in [23]. In the ideal case (trajectories identical to reference able-bodied trajectories), this coefficient would be equal to 1.

Vaulting is usually quantified by measuring peak ankle flexion power during single support [29]. However, in this paper, we utilize an alternative to kinetic quantification, using only kinematic parameters. This approach measures the peak sagittal-plane global foot angle (with respect to the ground) during single-support, which corresponds to the point of zero velocity, as shown in Fig. 5. This variable will be called vaulting angle for the purpose of statistical analyses.
Similar to [37], we quantified hip circumduction as the medio-lateral range of motion of the prosthetic ankle marker coordinates during a stride. This method corresponds better with what a clinician observes visually than other measures such as hip frontal-plane range of motion [38].

The symmetry index (SI) between the sound and prosthetic sides was quantified as

$$ SI = \frac{V_{\text{prosth}} - V_{\text{sound}}}{\frac{1}{2}(V_{\text{prosth}} + V_{\text{sound}})} $$

where $V_{\text{prosth}}$ and $V_{\text{sound}}$ are the gait variables corresponding to each side [39]. Based on this, $SI = 0$ represents perfect symmetry and increasing deviation from zero corresponds to increasing asymmetry.

For comparing differences in propulsion assistance between powered and passive legs, we analyzed propulsion and braking impulse [40], calculated by time integrals of the propulsive and braking portions of the force profile. We also calculated the symmetry index between the sound and prosthetic sides for each impulse, which were compared between the powered and passive use cases. Moreover, the symmetry of vertical force was computed with the ratio of the first force peak to the second force peak [27]. In the ideal case, this ratio would be equal to 1.

For backwards walking, we assessed the improvement in gait type from “step-to” gait (observed with the passive leg) to a “step-through” gait (observed with the powered leg) [41]. This asymmetry in backwards motion cannot be captured using standard step length symmetry assessments which are based on the distance covered by each foot with respect to itself. Instead, we assessed the distance along the direction of walking between the prosthetic versus sound ankle marker coordinates at their respective mid-stance positions. This allowed a relative (prosthetic versus sound) measure of foot positions.

The ability to cross over the obstacle was quantified by the height of the toe marker above the obstacle and also qualitatively assessed by plotting toe marker trajectories over the obstacle. Kicking the soccer ball was quantified by measuring ball velocity (estimated using coordinates of reflective markers affixed to the ball).

### D. Statistical analysis

Means and standard deviations were calculated for the biomechanical variables discussed in the previous section for each applicable condition resulting from a combination of 1) prosthetic device (powered, passive), 2) side (prosthetic, sound), and 3) speed (fast, normal, slow). Additionally, separate t-tests were performed to test the effect of prosthetic device (powered, passive) on dependent variables 1) vaulting angle and 2) hip circumduction for each speed (slow, normal, fast). This resulted in three t-tests performed for each dependent variable. To correct for multiple comparisons, a Holm-Bonferroni correction was applied to the significance level (alpha).

### IV. Results

A supplemental video of these experiments is available for download. The main goal of this paper was to demonstrate the ability of the controller to facilitate both rhythmic (walking) tasks as well as non-rhythmic (volitional) tasks and the improvements compared to the amputee subject’s passive leg. Our participant was able to successfully complete all (rhythmic and non-rhythmic) tasks that were part of the testing protocol as described in Section [III-B]. We present both qualitative as well as quantitative results in what follows.

#### A. Non-Rhythmic (Volitional) Tasks

The first non-rhythmic tasks were overground forward and backward walking between handrails, including start and stop. Fig. 6 displays the phase variable and joint angles through an overground forward walking trial. The subject started from rest (almost vertical leg), walked across the walkway, and stopped at the end. The change of the minimum ankle angle across strides is particularly interesting, as it represents the extent of pushoff. As the subject started from rest and increased his walking speed, the ankle plantarflexion also increased (i.e., larger pushoff) until the last stride where the subject decreased his speed and pushoff became smaller correspondingly.

The representative joint ankle powers for overground walking with powered and passive legs are displayed in Fig. [7] Although the passive ankle cannot produce power, its compliance allows for storage and release of energy through the stance phase. The comparison of the pushoff powers shows that the peak generated by the powered leg is more than three times larger compared to that of the passive leg. We will discuss the improvements in gait characteristics associated with this increase in the next subsection.

Fig. 8(a) shows mean antero-posterior force plots during forward walking trials with passive and powered prostheses. Passive legs intrinsically lack sufficient propulsive force, which results in compensatory reduction and asymmetry in braking forces on the subsequent stride. From the figure, it is clear that the powered leg provides greater propulsion and braking forces compared to the passive leg, which lead
to improved symmetry ratios. See Table I for impulse and symmetry index values for propulsive and braking forces.

Fig. 8(b) displays mean vertical ground reaction forces during overground walking. We calculated the ratio of the first force peak (corresponding to weight acceptance) to the second peak (corresponding to pushoff) as a measure of symmetry of walking [23]. This ratio was 1.17 with the passive leg and 1.13 with the powered leg. Again, the improved symmetry in the case of the powered leg can be attributed to the higher pushoff force applied by the powered ankle, which enhances the second force peak.

The results of a representative backward walking trial with the powered leg are depicted in Fig. 9. The holonomic nature of the controller enabled the subject to comfortably reverse his direction of motion and still maintain a smooth gait. Note that the phase variable has a reverse trajectory compared to Fig. 6. Because of the natural limitation of the passive prosthesis, the participant was forced to adopt a step-to pattern of walking, whereas he was able to walk using a normal step-through pattern using the powered leg. Our results show that use of the powered leg substantially improved the symmetry index for backwards walking: from 0.56 (passive leg) to 0.26 (powered leg). See Table II for details.

Crossing over an obstacle was another everyday volitional task that we tested and compared between the passive and powered legs. Fig. 10 shows the planar path of the toe marker task that we tested and compared between the passive and powered legs. Table II shows the planar path of the toe marker. It can be seen that the powered leg (blue solid curve) provided substantially greater toe clearance compared to the passive leg (dashed curve). In fact, in one of the trials, the participant toppled the obstacle with the passive leg (dotted curve). The maximum toe height reached with the passive leg was 0.106 m, whereas the powered leg reached 0.231 m.

As the last volitional experiment, we tested the performance of kicking a soccer ball with the passive and powered legs. We chose ball velocity as our performance metric. The trials with the passive leg resulted in velocities of 1.4 and 1.9 m/s (mean: 1.65 m/s), whereas the powered leg produced velocities of 5.0 and 5.5 m/s (mean: 5.25 m/s). The use of the powered prosthesis allows the participant to kick the soccer ball with a substantially higher velocity (218% increase in the mean ball velocity) when compared to using the passive prosthesis. Fig. 11 shows the thigh and knee angles as the subject kicks

---

**TABLE I**

Mean (SD) values of propulsion and braking impulses and their corresponding symmetry indexes: passive (P) versus powered (Pwr), and sound (S) versus prosthetic (P).

|                      | P  | S  | Pwr | Pwr |
|----------------------|----|----|-----|-----|
| Propulsion impulse (\%BW) | 0.65 | 1.99 | 1.25 | 1.98 |
| Propulsion impulse SI | 1.01 | 0.45 |     |     |
| Braking impulse (\%BW) | 0.87 | 1.68 | 1.46 | 1.65 |
| Braking impulse SI    | 0.64 | 0.13 |     |     |

**TABLE II**

Backward walking step symmetry: passive (P) versus powered (Pwr), and sound (S) versus prosthetic (P).

|                      | P  | S  | Pwr | Pwr |
|----------------------|----|----|-----|-----|
| Length (mm)          | 281 (32) | 498 (44) | 669 (44) | 513 (30) |
| SI                   | 0.56 | 0.26 |     |     |
the ball. Notice that the ball is kicked before maximum hip flexion (and hence maximum knee extension), but due to inertias, the leg continues moving forward. After reaching maximum flexion, the thigh retracts and the knee flexes for ground clearance. Finally, the thigh slightly extends forward, causing the knee to extend and the leg to rest on the ground. This shows the benefit of designing knee and ankle controllers based on following the motion of the thigh, which allows the subject to manage all of these maneuvers without difficulty.

B. Rhythmic Tasks

The knee and ankle angle trajectories for the trials with the passive leg on the treadmill are shown in Fig. [12] In particular, the absence of ankle plantarflexion (pushoff) in these trials is noticeable. In contrast, the ankle pushoff is quite conspicuous in the powered leg results of Figs. [13(a)] to [13(c)]. Note that since \( q_k^{\max} \) and \( q_k^{\min} \) were not changed in these trials, the minimum thigh angle is reached later than the reference trajectory during slow walking (Fig. [13(a)]). In other words, the ratio of stance to swing duration increases in order to provide extra time to achieve the minimum thigh angle while the foot is constrained to follow the treadmill speed. As the treadmill speed increases, the minimum thigh angle shifts to the left (Figs. [13(b)] and [13(c)] and the stance to swing duration ratio decreases. Also, the amplitude of the minimum thigh angle is consistently larger than the reference (about \(-17^\circ\) for all three speeds as opposed to \(-11^\circ\) for the reference trajectory). This means that the ankle pushoff was not fast enough to quickly reverse the direction of motion of the thigh and prepare it for the swing phase [42]. Similar trends are visible in the results of experiments with an able-bodied subject with the same control parameters (Figs. [14(a)] to [14(c)]. The Pearson correlation coefficients for the knee joint (Table III) confirm these observations. The effect of the overlong stance to swing duration ratio in slow speed with the powered leg resulted in a lower correlation than that of the passive leg (0.80 versus 0.93). However, with the increase of speed, the correlation of the powered leg improves, and in fast speed it significantly outperforms the passive leg (0.95 versus 0.75). These observations will be discussed in detail in the next section.

Next, we focus our analysis on common foot clearance compensation strategies employed by transfemoral amputees (i.e., vaulting and hip circumduction) for walking tasks. The goal is to determine whether our controller improves these compensations compared to the subject’s passive leg. Fig. [15]
Fig. 13. Mean ± std for phase variable, and commanded and measured joint angles as a function of normalized time for the treadmill tests with the amputee subject: (a) a 60-second trial with slow speed (0.7 m/s); (b) a 60-second trial with normal speed (1.0 m/s); and (c) a 30-second trial with maximum speed (1.6 m/s).

Fig. 14. Mean ± std for phase variable, and commanded and measured joint angles as a function of normalized time for treadmill test with the able-bodied subject; (a) a 30-second trial with slow speed (0.7 m/s); (b) a 30-second trial with normal speed (0.9 m/s); and (c) a 30-second trial with fast speed (1.1 m/s).
TABLE III
PEARSON CORRELATION COEFFICIENTS BETWEEN THE JOINT TRAJECTORIES OF EACH PROSTHETIC LEG (PASSIVE OR POWERED) AND THE REFERENCE ABLE-BODIED DATA [8] FOR DIFFERENT WALKING SPEEDS.

| Speed  | Pas knee | Pwr knee | Pas ankle | Pwr ankle |
|--------|----------|----------|-----------|-----------|
| Slow   | 0.93     | 0.80     | 0.15      | 0.71      |
| Normal | 0.96     | 0.96     | 0.16      | 0.94      |
| Fast   | 0.75     | 0.95     | -0.20     | 0.97      |

shows the mean global foot angles of the sound leg during the fast treadmill walking speed using passive and powered legs. Additionally, the global foot angle of the able-bodied reference data is shown for comparison [8]. It can be seen that the amputee subject employed a rising global foot angle during the single-support phase (for both passive and powered legs) when compared to the much more constant angle of able-bodied reference data during the same phase. Importantly, the vaulting angle (indicated by a red circle) was lower when using the powered leg when compared to using the passive leg. The statistical results are also in agreement with the graphical illustration. Use of the powered prosthesis facilitated a significant ($p < 0.001$, $p < 0.001$, $p < 0.001$) decrease in vaulting angle for fast, normal and slow speeds, respectively, when compared to using the passive leg. See Table IV for mean, SD, and symmetry index values.

V. DISCUSSION

As the supplemental video presents, the proposed controller enabled the amputee subject to accomplish a variety of volitional (walking forward and backward, instantaneous start and stops, walking over obstacles, shooting a soccer ball) and periodic (walking on a treadmill with different speeds) tasks. Although the designed virtual constraints were based on a periodic task, i.e., normal-speed walking kinematics, the holonomic nature of the controller helped the subject

cumduction symmetry ratios were improved with the powered leg for all speeds, showing that the amount of circumduction with the powered leg was closer to that of the sound leg (taken as a reference). See Table IV for mean, SD, and symmetry index values.

TABLE IV
ANKLE LATERAL DEVIATION (HIP CIRCUMDUCTION): POWERED VERSUS PASSIVE PROSTHESSES MEAN (SD) AT DIFFERENT SPEEDS. ASTERISKS (** AND *** ) INDICATE SIGNIFICANT DIFFERENCES ($p < 0.01$ AND $p < 0.001$, RESPECTIVELY) BETWEEN PASSIVE AND POWERED LEGS.

| Speed  | Pas P | Pwr P | SI Pas | SI Pwr |
|--------|-------|-------|--------|--------|
| Slow   | 46.4 (10.2) | 45.5 (5.7) | 0.2     | 0.1     |
| Normal | 71.8 (19.4) | 55.8 (13.5) | 0.3     | 0.1     |
| Fast   | 105.8 (9.5) | 70.3 (12.1) | 0.8     | 0.5     |
perform non-rhythmic tasks as well. Unlike previous controllers that used thigh angle to parameterize the gait [6], [20], the proposed controller is not limited to only one type of motion (rhythmic or non-rhythmic). In what follows, we discuss about the performance of the proposed controller and the observed improvements compared to that of the amputee subject’s passive leg. Since the goal of this study has been to investigate the acute biomechanical effects of the powered leg, we excluded the variables that are significantly dependent on adaptation time, such as step width [43].

A. Biomechanical Analysis of Rhythmic and Volitional Tasks

One of the main objectives of this paper is to demonstrate the ability of the powered prosthesis to facilitate both rhythmic and non-rhythmic volitional tasks. Compensatory mechanisms during walking (primarily for aiding foot clearance) are problematic in general because of higher energy expenditure [23], [44], [45] and deterioration of the intact joints [25], [26], [46]. We hypothesized that the assisted (powered) ankle plantarflexion during push-off and knee flexion during early swing would help reduce these compensations, namely, vaulting and hip circumduction.

Our amputee participant exhibited moderate to severe vaulting with his passive leg. Analysis with the vaulting angle metric shows that vaulting improved with the powered leg as compared to passive for all walking speeds tested. The improvement was consistent across walking trials as evidenced by the highly significant t-tests. Based on this result, we expect that the use of our powered device with the proposed controller can mitigate the deleterious effects of vaulting such as metatarsal pain and the sinus tarsi syndrome caused by overloading of the forefoot [29]. Moreover, we can expect a reduction in energy expenditure through reduced plantarflexion and vertical displacement of the whole body [47]. Likewise, the use of the powered leg led to a significant reduction in hip circumduction compared to the excessive level observed when using the passive leg. Similar benefits to those of vaulting reduction can be expected from the hip circumduction improvement as well.

We validated the assistance delivered through active push-off by analyzing force plate data. Our results confirm the observation in [49] that propulsion and braking forces are higher when using a powered prosthesis compared to a passive one. More importantly, we showed that the increases in braking and propulsive forces are associated with an improvement in the symmetry index. Similarly for the vertical force, the second force peak (associated with pushoff) was enhanced, resulting in a more symmetric walking force profile. In the case of passive prostheses, the lack of active pushoff and knee flexion during the stance phase results in reduced propulsion forces. To maintain a constant walking velocity, braking forces are usually reduced in the subsequent stance phase. Moreover, amputees generally tend to rely more on the sound leg for weight bearing and propulsion purposes, resulting in numerous clinical symptoms, for example, osteoarthritis and back pain [27]. A more balanced (symmetric) use of both legs is encouraged by clinicians to mitigate such symptoms. Thus the improvement in braking and propulsion force symmetry ratios is a significant highlight, considering its potential benefit in reducing common overuse symptoms.

The joint trajectories of the powered prosthesis had noticeably greater correlations with the reference able-bodied data compared to those of the passive leg. The passive leg only had a greater correlation at the knee for the slow speed, which was due to the powered leg’s use of a single parameterization for all speeds, as discussed in the previous section. As the speed increased, the greater propulsion force generated from the ankle pushoff of the powered leg resulted in more normative trajectories than the passive leg.

The joint ankle power of the powered leg (as depicted in Fig. 7) is in the same level as reported in [6] (about 100 W), and as expected, substantially increases pushoff power compared to the passive leg. The peak power is mainly limited by the motor’s speed and torque range as well as the high reflected inertia of the actuator system, which consumes a substantial part of the motor power for acceleration and deceleration. These factors also limit the knee flexion speed at the swing phase onset, which resulted in toe-stubbing at some strides. We expect that with the use of higher-power motors with lower reflected inertias (i.e., lower transmission ratios) such as [48], the controller will achieve pushoff powers comparable to that of able-bodied subjects. Despite these limitations, the comparison of the horizontal and vertical forces of the powered leg with those of the passive leg (Fig. 8) shows improved symmetry and significant increases in propulsive and braking impulses.

The speed-invariance of the controller and its improved pushoff management also allowed for greater walking speeds compared to the maximum speed achieved in [6], which was 1.2 m/s. With the present controller, the amputee subject was able to comfortably walk at 1.6 m/s, which can also be compared with the maximum walking speeds achieved in [9] (1.4 m/s) and [23] (1.6 m/s). It is worth noting that the subject who achieved 1.6 m/s in [23] was able to walk at the same speed using his passive leg, whereas our amputee subject was only able to reach a maximum speed of 1.3 m/s with his passive leg. In our preliminary experiments, the able-bodied subject wearing the prosthesis using a bypass was able to reach 1.8 m/s with the present controller. The limitation for high speeds primarily originates from the torque saturation of the motors and not from the controller. The inadequate ankle torque and power during stance and the limited speed of the knee actuator during swing require the users to compensate using torque from their hips, which quickly leads to fatigue at fast speeds. Overall, the results highlight the substantial improvement made in the controller performance as compared to [20] in which pushoff had to be eliminated to avoid abrupt jumps.

Backward walking is an interesting feature of our control approach that cannot be achieved with the traditional impedence-based control methods of powered knee-ankle prostheses, which are limited to working only in forward direction. As mentioned, for the backwards walking task, our participant utilized a step-to gait when using the passive prosthesis. The lead leg was the sound leg, which the prosthetic leg
followed. This choice of gait is not surprising as the passive knee is incapable of load bearing when behind the body. On the other hand, a more natural step-through gait was adopted with the powered prosthesis. Comparison of step symmetry for backwards walking shows that the powered leg enables more stable and more natural-looking maneuvers.

An analysis of the obstacle crossing task (Fig. 10) shows that our participant was easily able to clear the obstacle using his powered leg, with a clearance of 100 mm between the toe marker and top edge of the obstacle. While using the passive leg, this same task was extremely difficult. The obstacle was toppled over on one attempt and was barely cleared on the second try. The controller of the powered leg is designed to provide maximum knee flexion (clearance) at a hip angle of approximately 12° of flexion. Therefore, the optimal way to cross the obstacle for an amputee would be to maintain this hip angle and guide the hip joint (and thus the prosthesis) over the obstacle. Our amputee participant was quickly able to figure out this optimal strategy without any guidance from the experimenters. This is due to the holonomic mapping between the thigh angle and the foot position, which is easy for the subjects to learn. Once again, this emphasizes the intuitiveness of the controller for common volitional tasks.

Kicking the soccer ball requires a rapid yet controlled extension of the knee. Our participant was able to kick the soccer ball using the passive as well as the powered legs. However, the mean ball velocity was more than three times higher when using the powered leg. This result is impressive as it demonstrates the intuitive control of the powered knee and ankle at high speeds. Based on the control strategy, a fast extension of the knee can be achieved by a rapid flexion of the hip. Our amputee participant was able to gain an understanding of these mechanics and execute them after only a couple of practice trials. Again, no guidance or feedback was provided to the participant regarding the optimal mechanics.

B. Limitations

The use of a purely holonomic set of virtual constraints also has a limitation. There is a relatively flat section in the middle part of the phase variable plots (normalized time of about 0.5-0.6) in both Figs. 13(a) to 13(c) and Figs. 14(a) to 14(c), meaning that the rate of the phase variable is almost zero. To investigate the reason for this phenomenon, note that

\[ \dot{s} = \frac{ds}{dq_h} \dot{q}_h, \]  

which means for \( \dot{q}_h = 0 \), we will have \( \dot{s} = 0 \). This condition occurs during pushoff (transition from S2 to S3). As a result of \( \dot{s} = 0 \), the knee and ankle rates also tend to vanish, and pushoff becomes slower. This contributes to the thigh continuing backward, before the ankle plantarflexion increases enough to stop the backward motion and drives the thigh forward. This is intrinsic to the holonomic virtual constraints and can be regarded as a trade-off.

Another observation from the treadmill tests was the stance to swing duration ratio. Note that the leg’s joint kinematics and especially the maximum and minimum of thigh angle change as walking speed varies [49]. For the present study, we kept the kinematics (virtual constraints) unchanged, in order to demonstrate the ability of the controller to work in different situations with minimal tuning. As a result, for low speeds the stance to swing duration ratio was greater than expected. However, the decrease of stance to swing ratio and earlier pushoff in higher speeds is still in accordance with observations from able-bodied subjects [49]. Moreover, walking speed can be estimated using fairly straightforward methods (see [50], [51], for example), and the virtual constraints can be changed accordingly. A similar idea can incorporate the necessary kinematic differences between forward and backward trajectories. Note that adding these additional virtual constraints is merely a kinematic modification, whereas the dynamic joint attributes (i.e., joint impedances) remain unchanged. This provided invariance across subjects for the phase-based controller that was proposed and tested in our previous work [6]. The comparison between able-bodied and amputee users of the powered prosthesis (Figs. 13 and 14) offer some evidence that our updated phase variable retains this beneficial property. However, the purpose of this study was to demonstrate the potential for increased functional abilities rather than subject independence.

VI. Conclusion

A controller for volitional control of a range of periodic and non-periodic tasks was designed for powered knee-ankle prostheses and validated through experiments with an above-knee amputee subject. The controller uses a phase variable defined as a piecewise holonomic function of the thigh angle with transitions based on a finite state machine. In addition to volitional tasks, the controller enabled operation in an extensive range of walking speeds with statistically significant improvements in compensations associated with passive prostheses.

Although the controller facilitates a wider range of tasks than walking, it does not encapsulate tasks such as stair ascent and descent. However, the structure provided is flexible for incorporating new sets of kinematics in a task-recognition framework [51]–[53] or unifying framework [54].

An interesting extension of the controller would include a nonholonomic correction for the interruption in pushoff for a faster and smoother transition to the swing phase. Furthermore, we plan to test the controller on our newly designed leg [48], which provides greater torques as well as backdrivability. These investigations will provide a better understanding of the controller’s abilities, benefits, and potential improvements.

ACKNOWLEDGMENT

The authors would like to thank Christopher Nesler for his help in conducting the experiments.

REFERENCES

[1] M. Rabuffetti, M. Recalcati, and M. Ferrarin, “Trans-femoral amputee gait: Socket-pelvis constraints and compensation strategies,” Prosthetics and Orthotics International, vol. 29, no. 2, pp. 183–192, 2005.
[2] S. Au, M. Berniker, and H. Herr, “Powered ankle-foot prosthesis to assist level-ground and stair-descent gaits,” Neural Networks, vol. 21, no. 4, pp. 654–666, 2008.
of prosthetic alignment and different prosthetic components;” *Gait & posture*, vol. 16, no. 3, pp. 255–263, 2002.

[45] R. L. Waters and S. J. Mulroy, “Energy expenditure of walking in individuals with lower limb amputations,” in *Atlas of Limb Prosthetics: Surgical, Prosthetic, and Rehabilitation Principles*, D. Smith, J. Michael, and J. Bowker, Eds. American Academy of Orthopedic Surgeons, 2004, ch. 32, pp. 395–407.

[46] P. A. Struyf, C. M. van Heugten, M. W. Hitters, and R. J. Smeets, “The prevalence of osteoarthritis of the intact hip and knee among traumatic leg amputees,” *Archives of physical medicine and rehabilitation*, vol. 90, no. 3, pp. 440–446, 2009.

[47] R. L. Waters and S. Mulroy, “The energy expenditure of normal and pathologic gait,” *Gait & posture*, vol. 9, no. 3, pp. 207–231, 1999.

[48] T. Elery, S. Rezazadeh, C. Nesler, J. Doan, H. Zhu, and R. D. Gregg, “Design and Benchtop Validation of a Powered Knee-Ankle Prosthesis with High-Torque, Low-Impedance Actuators,” in *Proc. IEEE Int. Conf. Robot. Autom.*, Brisbane, Australia, 2018.

[49] C. Kirtley, M. W. Whittle, and R. Jefferson, “Influence of walking speed on gait parameters,” *Journal of Biomedical Engineering*, vol. 7, no. 4, pp. 282–288, 1985.

[50] D. Quintero, D. J. Villarreal, and R. D. Gregg, “Real-time continuous gait phase and speed estimation from a single sensor,” in *IEEE Conference on Control Technology and Applications (CICTA)*, 2017, pp. 847–852.

[51] H. Varol, F. Sup, and M. Goldfarb, “Multiclass Real-Time Intent Recognition of a Powered Lower Limb Prosthesis,” *IEEE Trans. Biomed. Eng.*, vol. 57, no. 3, pp. 542–551, Mar. 2010.

[52] A. J. Young and L. J. Hargrove, “A Classification Method for User-Independent Intent Recognition for Transfemoral Amputees Using Powered Lower Limb Prostheses,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 24, no. 2, pp. 217–225, Feb. 2016.

[53] H. L. Bartlett and M. Goldfarb, “A Phase Variable Approach for IMU-Based Locomotion Activity Recognition,” *IEEE Trans. Biomed. Eng.*, vol. 65, no. 6, pp. 1330–1338, 2018.

[54] K. R. Embry, D. J. Villarreal, and R. D. Gregg, “A unified parameterization of human gait across ambulation modes,” in *38th Annual Proc. IEEE Int. Conf. Eng. Med. Biol. Soc.*, Aug. 2016, pp. 2179–2183.

Siavash Rezazadeh (M’14) received his B.Sc. and M.Sc. from Sharif University of Technology, and his PhD from University of Alberta. Currently, he is a Research Scientist at the Locomotor Control Systems Laboratory, University of Texas at Dallas, Richardson, TX, working on design and control of prosthetic legs. Before joining Locolab, he worked at Dynamic Robotics Laboratory, Oregon State University on control of ATRIAS, a bipedal robot, for DARPA Robotics Challenge (DRC). His research interests include using fundamental concepts of mechanics for design and control of novel robots and mechanisms.

David Quintero (S’15) received the B.S. degree (2006) in mechanical engineering from Texas A&M University and the M.S. degree (2008) in mechanical engineering from Stanford University. He received the Ph.D. degree (2018) in mechanical engineering from the University of Texas at Dallas. He is now an Assistant Professor at San Francisco State University. His research is in robotics, controls, system identification, and wearable sensing for applications in rehabilitation.

Emma Reznick received the B.A. degree (2016) in ACS Biochemistry with an emphasis in Electrical Engineering from Colorado College. After working as a Post-Baccalaureate Researcher at the University of Texas at Dallas, she is now pursuing a Ph.D. in bioengineering at the same institution. Her research interests relate to biomechanics and accessible tuning of powered prostheses.

Leslie Gray received her B.S. degree (2002) in Prosthetics-Orthotics from UT Southwestern Medical Center, the M.Ed. degree (2007) in Instructional Technology from the University of Texas at Brownsville, and is board certified in orthotics (2003) and prosthetics (2004). She is an Assistant Professor and serves as Director of the Prosthetics-Orthotics Program at the University of Texas Southwestern Medical Center in Dallas, TX.

Nikhil Divekar (S’18) received the B.S. and M.S. degrees in mechatronics engineering and biomedical engineering from University of Cape Town (UCT) in 2008 and 2013, respectively. He is currently working toward a Ph.D. degree in biomedical engineering at University of Texas at Dallas (UTD). Prior to joining UTD, he was a research officer in biomechanics at the Department of Exercise Science and Sports Medicine of UCT. His research interests include clinical biomechanics, neurophysiological basis of rehabilitation and wearable robotics for applications in rehabilitation.

Robert D. Gregg (S’08-M’10-SM’16) received the B.S. degree (2006) in electrical engineering and computer sciences from the University of California, Berkeley and the M.S. (2007) and Ph.D. (2010) degrees in electrical and computer engineering from the University of Illinois at Urbana-Champaign.

He joined the Departments of Bioengineering and Mechanical Engineering at the University of Texas at Dallas (UTD) as an Assistant Professor in 2013. Prior to joining UTD, he was a Research Scientist at the Rehabilitation Institute of Chicago and a Postdoctoral Fellow at Northwestern University. His research is in the control of bipedal locomotion with applications to autonomous and wearable robots.