Controlling Powered Prosthesis Kinematics over Continuous Transitions Between Walk and Stair Ascent

Shihao Cheng, Curt A. Laubscher, and Robert D. Gregg

Abstract— **One of the primary benefits of emerging powered prosthetic legs is their ability to facilitate step-over-step stair ascent by providing positive mechanical work. Existing control methods typically have distinct steady-state activity modes for walking and stair ascent, where activity transitions involve discretely switching between controllers and often must be initiated with a particular leg. However, these discrete transitions do not necessarily replicate able-bodied joint biomechanics, which have been shown to continuously adjust over a transition stride. This paper presents a phase-based kinematic controller for a powered knee-ankle prosthesis that enables continuous, biomimetic transitions between walking and stair ascent. The controller tracks joint angles from a data-driven kinematic model that continuously interpolates between the steady-state kinematic models, and it allows both the prosthetic and intact leg to lead the transitions. Results from experiments with two transfemoral amputee participants indicate that knee and ankle kinematics smoothly transition between walking and stair ascent, with comparable or lower root mean square errors compared to variations from able-bodied data.**

I. Introduction

Stairs are a challenging activity of daily life and are often responsible for fall-related accidents [1], [2], particularly during the transition strides before or after ascending stairs [3]. These transitions can be especially challenging for individuals with transfemoral amputation as they have lost direct control over their lower limb [4]. Although passive prostheses enable daily activities, the lack of positive work typically limits users to ascending stairs step-to-step (instead of normative step-over-step) and requires compensation using their residual and intact limbs (e.g., hip-hiking, vaulting) [5]. These compensations increase the risk of secondary injuries such as back pain and knee osteoarthritis from the intact limb [6], [7]. In addition, users are instructed to lead the transitions to stairs with a particular leg [8]. For example, the walk to stair ascent transition requires the intact leg to step on the stairs first. This can slow down the locomotion when approaching the transitions and sometimes requires a complete stop to adjust the foot placement or step length [9].

Powered prosthetic legs can reduce the risk of secondary injury by providing net-positive work at the knee and ankle joints [10], [11]. They enable step-over-step stair ascent, which more closely mimics able-bodied biomechanics and is a desirable characteristic for prostheses [12]. This allows for the possibility of leading transitions with either the prosthetic or intact leg, making locomotion more flexible for the user. Although powered devices such as the Össur PowerKnee have been commercialized to facilitate step-over-step stair ascent, the walk to stair transition requires the user to follow a specific procedure including a complete stop, which is undesirable in repeated daily use [13]. Therefore, an ideal powered prosthetic leg controller should allow for prompt and uninterrupted transitions, led by either the prosthetic or intact leg, while emulating the anatomical function of the biological leg to minimize the strain on intact limbs.

Recent studies focus on developing controllers that can transition from/to stairs without a pause from the user. A common method to accomplish transitions is to switch between steady-state controllers during the transition strides [14]–[16]. While steady-state controllers often aim to improve symmetry and biomimicry by tracking reference kinematics/kinetics from an able-bodied dataset [17]–[19], discrete transitions do not necessarily resemble normative behavior. Research on lower-limb joint biomechanics has found significant differences in stair transitions compared to steady-state activities [20], suggesting explicit consideration of transition biomechanics is necessary. In addition, our previous work [21] observed and modeled a continuous kinematic transition that lasts over the entire transition stride for the leg leading the transition. Therefore, discrete switching between steady-state controllers is insufficient to produce biomimetic transitions led by the prosthetic leg. Kim et al. [22] proposed a deep neural network that employs latent and time sequence features to generate desired impedance parameters for various steady-state activities. While they did not consider biomimetic transitions data in training their neural network, their proposed method could potentially be extended to do so. However, such neural network-based methods are prone to bias from data imbalance, where transition activities in particular may be lacking in quantity.

To solve these challenges, this paper presents a phase-based kinematic controller for a powered knee-ankle prosthesis that enables both prosthetic and intact limb leading transitions to/from stair ascent. Continuous and seamless transitions can be achieved by establishing relationships between transitional and steady-state kinematics. When the prosthetic leg leads the transition, we utilize our recent kinematic model [21] which continuously interpolates between preceding and subsequent steady-state kinematics over the transition stride. When the intact leg leads the transition, that leg handles the continuous kinematic adjustments, and the prosthetic leg controller can employ a simple switching-based strategy.

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S. Cheng, C. Laubscher, and R. Gregg are with the Department of Robotics, University of Michigan, Ann Arbor, MI, 48109 USA. Contact: {chengsh, claub, rdgregg}@umich.edu

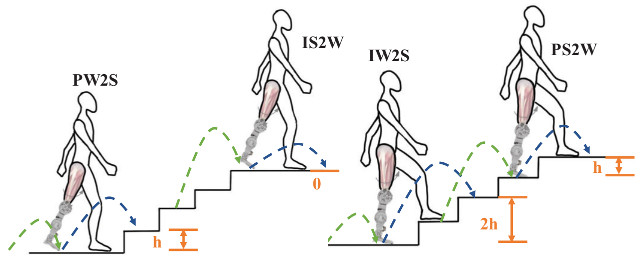

Fig. 1. Transitions between walking and stair ascent for prosthetic and intact leading conditions. IW2S and IS2W respectively denote walk-to-stair and stair-to-walk transitions led by the intact leg, whereas PW2S and PS2W refer to these transitions being led by the prosthetic leg. The green dashed curves denote preceding swing trajectories, and the blue dashed curves denote the subsequent transition swing trajectories. Elevation changes are indicated in units of stair height (h).

This transition control method is experimentally validated by investigating the resulting kinematics during all four possible transitions (prosthetic vs. intact leading for walk to stair ascent and vice versa) with two transfemoral amputee participants. The main contribution of the paper can be summarized as follows:

- Define and analyze the four transitions between walk and stair ascent led with either the intact or prosthetic leg by considering the elevation change within those strides (Fig. 1).
- Compare the joint kinematics between different transitions and expand the transition model [21] from prosthetic leading to intact leading by switching the steady-state kinematics models at a predefined transition point.
- Implement and validate a novel kinematics-based controller that enables biomimetic transitions between walk and stair ascent, led by either the intact or prosthetic leg, with two transfemoral amputee participants.

II. Control Method

This section defines the different transition cases between walk and stair ascent, reviews the transition kinematics models proposed in our previous work [21], and expands these models to the case of the intact leg leading the transition. We then introduce the continuous transition controller built upon these kinematics models.

A. Transition Kinematics Model

A transition stride between stairs and walking is a stride with one heel strike on the staircase and either the preceding or subsequent heel strike off the staircase, as shown in Fig. 1. The top step on a staircase is considered a stair step. Depending on which leg contacts the new terrain first, the transition to/from stairs can be led by either the intact or prosthetic leg. Therefore, we define two walk to stair ascent (W2S) transitions, one for the intact leg leading (IW2S) and one for the prosthetic leg leading (PW2S). Similarly, we define two stair ascent to walk (S2W) transitions, one for intact leading (IS2W) and another for prosthetic leading (PS2W). Unlike steady-state stair ascent, the two prosthetic-leading transitions (PW2S and PS2W) involve the limb rising a single step rather than two. Therefore, these transitions need to be modeled separately from the steady-state activities.

Next, we give a concise review of the transition kinematics models from our previous work [21], which were based on the steady-state modeling framework of Embry et al. [23]. In the latter, steady-state walking (SSW) kinematics $\Theta^w(\phi,\chi^w)$ were modeled as a continuous function of gait phase (ϕ), representing stride progression, and task (χ), comprising forward speed (v) and ground inclination (ι). This function was calculated as a linear combination of task-dependent Bernstein polynomials with phase-dependent coefficients taken as finite Fourier series. A convex optimization problem, detailed in Section II-E in [23], was solved for the best fit of the kinematics model to mean able-bodied (AB) values for all recorded combinations of walking speeds (0.8, 1.0, 1.2 $\mathrm{m\,s^{-1}}$) and inclines (-10 to 10°).

Cheng et al. [21] expanded this kinematics model to steady-state stair climbing (SSS) over staircase inclines between -35 and 35 degrees ($\Theta^s(\phi,\chi^s)$) based on the same algebraic structure as SSW. Using a modified task function space, the kinematics model was also expanded to transitions between walking and stair ascent from the perspective of the leading leg. The leading leg's joint kinematics tended to gradually change from the preceding steady-state activity to the subsequent steady-state activity over the transition stride. Thus, the transition kinematics were defined to lie within the convex hull of the two steady-state activity kinematics, with an offset to account for brief deviations from this assumption:

$$\begin{aligned}\Theta^{ts}(\phi,\chi) =& \alpha(\phi,\chi)\Theta^w(\phi,\chi^w) \\ &+(1-\alpha(\phi,\chi))\Theta^s(\phi,\chi^s)+O(\phi,\chi)\gamma(\phi,\chi),\end{aligned} \tag{1}$$

where Θ^{ts} was a vector containing the knee θ^k and ankle θ^a joint angles of the transition models. At each point in time, the indicator function $O(\phi,\chi)$ ascertained whether the transition joint angle falls outside the convex hull formed by the steady-state walk and stair models so that the linear offset angle $\gamma(\phi,\chi)$ can be applied to account for the offset if true. The convex coefficient $\alpha(\phi,\chi)\in[0,1]$ described how transition kinematics change (in a monotonic but nonlinear manner) between the two steady-states during the transition. Both $\alpha(\phi,\chi)$ and $\gamma(\phi,\chi)$ were defined as a weighted sum of a set of functions of gait phase and task conditions, similar to $\Theta^w(\phi,\chi^w)$. Then, a convex optimization solved for the unknown coefficients in $\alpha(\phi,\chi)$ and $\gamma(\phi,\chi)$ to fit the transition model to the mean AB transition kinematics, as detailed in [21, Section II-C].

In this paper, we control prosthetic leading transitions using this continuous model. However, prior work did not consider the perspective of the trailing leg, which is the prosthesis during an intact-led transition. To determine how to control the prosthesis in this case, we analyze the trailing leg's joint kinematics in the AB dataset [24], see Fig. 2. For W2S, the trailing leg's kinematics resemble steady-state walking over the entirety of stance for the knee and about 70% of stance for the ankle. After toe off, the leg rises two steps in a similar manner to the swing phase of steady-

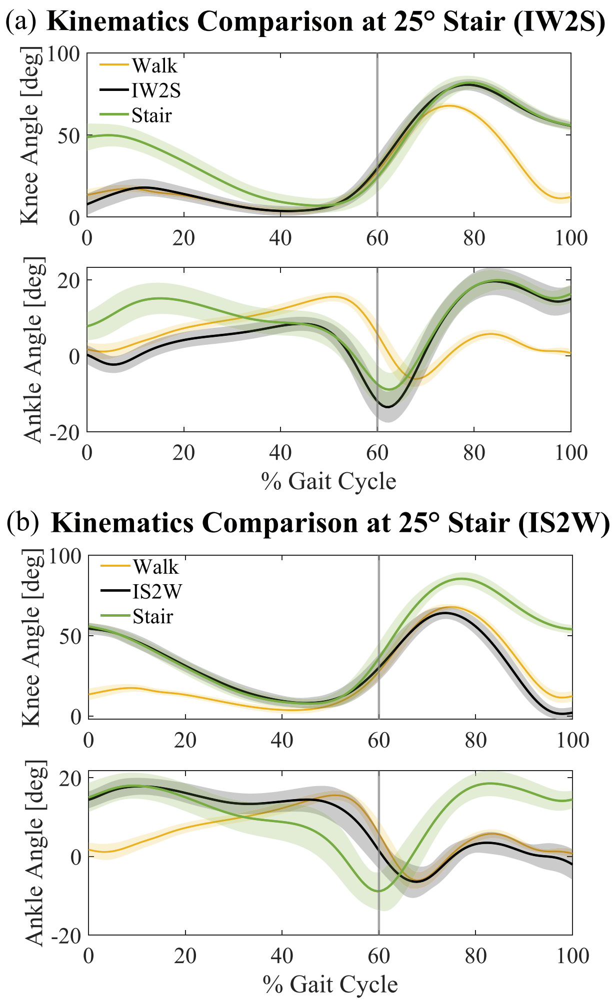

Fig. 2. Knee and ankle kinematics comparison between trailing leg during walk-stair transitions (i.e., prosthetic leg during intact-led transitions), steady-state walking stride, and steady-state stair ascent stride. (a) The transition from walk to stair ascent at 25° inclination. (b) The transition from stair ascent to walk at 25° inclination. The grey vertical line indicates 60% of the gait cycle as an approximate average toe off phase [24], [25].

state stair ascent (for both the knee and ankle). Because toe off can be robustly detected with onboard sensors of the powered prosthetic leg (e.g., load cell), we determine that the prosthetic leg during IW2S can be controlled by discretely switching between steady-state activities at toe off. In other words, the intact leg handles the continuous transition adjustments described previously. Similarly, in IS2W, the prosthetic leg can use steady-state stair ascent kinematics before toe off, after which the elevation will not change, similar to the swing phase of steady-state walking. Hence, we choose the same transition point at toe off for a discrete switch between steady-state activity controllers during IS2W.

B. Continuous-Varying Position Controller

The kinematics models described in Section II-A are used to control the powered prosthetic leg over the various steady-state and transition activities, as shown in Fig. 3. The thigh angle in the sagittal plane is used to estimate the gait

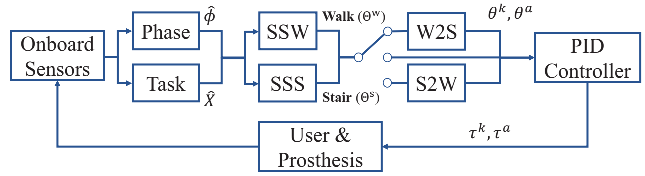

Fig. 3. Block diagram of the position-based transition controller based on the steady-state and transition kinematics models [21]. The gait phase ($\hat{\phi}$) is estimated in real-time with fixed task variables ($\hat{\chi}$) for the experiment. SSW, SSS, W2S, and S2W represent steady-state walking, steady-state stair, walk-to-stair, and stair-to-walk models, respectively. Depending on the current activity (walk or stair), different kinematics models will be used to generate virtual constraints for knee and ankle joints. A proportional–integral–derivative (PID) controller calculates the required motor torques (τ^k and τ^a) to track the desired joint angles.

phase ($\hat{\phi}$) in real-time using a finite-state-machine to switch between different monotonic regions as detailed in our prior work [26], [27]. The task conditions $\hat{\chi} = (v, \iota)$ are fixed to be $\hat{\chi}^w = (1\,\mathrm{m\,s^{-1}}, 0°)$ and $\hat{\chi}^s = (1\,\mathrm{m\,s^{-1}}, 26.5°)$ for walking and stair ascending, respectively, in this study. During SSW or SSS, the corresponding steady-state model is used to generate desired joint angles to be tracked as virtual constraints [17]. Switching between steady-state models occurs during intact-led transitions as described previously. W2S or S2W models are used for prosthetic-led transitions. Since this paper focuses on the mid-level kinematic controller, transitions are triggered manually with a specified leading leg during the experiment (i.e., we assume ideal high-level classification). Ultimately, a proportional–integral–derivative (PID) controller is employed to follow the reference joint angles of both the knee and ankle. Discontinuities in the reference joint angles, which could arise due to errors in the kinematic models, were mitigated via a rate limiter.

III. Experimental Methods

This section describes the powered prosthetic leg hardware, experimental methods, and data analysis.

A. Experiment Setup

The kinematic controller was implemented on the powered knee-ankle prosthesis from [28], which is equipped with low-impedance actuators and G-SOLO Twitter R80A/80VDC drivers (Elmo Motion Control, Petah Tikva, Israel). The motors provide enough torque through 22:1 single-stage stepped-planet compound planetary gear transmissions to perform demanding activities such as stair ascent [11]. An inertial measurement unit 3DM-CX5-25 (LORD Microstrain, Williston, VT) measures the residual thigh's global orientation, and motor positions are tracked with E5 optical quadrature encoders (US Digital, Vancouver, WA). Additionally, a 6-axis load cell (M3564F, Sunrise Instruments, Nanning, China) is positioned at the distal part of the shank, directly above the prosthetic foot. The controller and all the sensors sampled at a frequency of 500 Hz.

To test the control framework, we carried out an experiment involving two transfemoral (TF) amputee participants. Table I summarizes participant anthropometrics and other

TABLE I
PARTICIPANT INFORMATION

ID	Sex	Age (yrs)	Height (cm)	Mass (kg)	K-Level*
TF01	M	18	183	68	4
TF02	M	26	192	116	4

* The etiology of amputation was congenital for both participants.

Fig. 4. Experiment setup for the controller evaluation between walking, transitions, and stair ascending. A parallel bar and a five-step staircase were placed in sequence. As a precautionary measure, the participant wore a safety harness. A licensed prosthetist fitted the powered prosthetic leg [28] to TF participants for proper alignment.

general information. Both participants had more than 5 hours of experience using the powered prosthesis prior to the experimental study conducted in this paper. The University of Michigan Institutional Review Board granted approval for the research protocol on August 28, 2020, with protocol number HUM00166976. Each subject provided written consent before the experiment.

Parallel bars were placed leading into a five-step staircase (step height: 15.2 cm / 6 inch, incline: 26.4°) with handrails as shown in Fig. 4. An overhead safety harness was fastened to the participant during the experiment as a precautionary measure. A licensed prosthetist fitted the powered prosthetic leg to the TF participants with proper alignment to ensure symmetry with the intact leg. We provided at least two hours of acclimation time for each participant to adapt to the powered prosthetic leg and reduce the participants' compensation motions before the actual data collection.

For the main experiment, we performed 48 total trials divided into 6 segments, alternating which leg is leading between segments. At the start of each segment, we instructed the participant to lead the transition from walking to stair ascent with different legs (i.e., either prosthetic or intact leg). For each trial, the participant started with a pre-transition level walking stride, followed by the transition stride to stair ascent, one or two stair ascending strides, a transition stride back to level walking, and one or two post-transition level walking strides on the stair platform. Due to the experimental setup consisting of an odd number of step on the staircase, a sequence of stair strides beginning with IW2S will end with PS2W, and a sequence of stair strides beginning with PW2S will end with IS2W (see Fig. 2). Thus, the two full sequences with all activities were Walk $\rightarrow$ PW2S $\rightarrow$ Stair $\rightarrow$ IS2W $\rightarrow$ Walk and Walk $\rightarrow$ IW2S $\rightarrow$ Stair $\rightarrow$ PS2W $\rightarrow$ Walk. We assigned a 2-5 minute break between each segment to prevent muscle fatigue. Biomechanical data of the powered prosthetic leg were recorded with a sampling frequency of 250 Hz, and the entire experiment was video-recorded.

B. Data Analysis

To show the knee and ankle joint transition kinematics continuously vary between the adjacent steady-state activities, we plotted the averaged kinematics for each activity within the two full sequences. Then, we compared the kinematics and phase from the recorded data of the prosthetic leg with the AB data for each of the four transitions we considered for this study. Thigh, knee, and ankle kinematics were compared to the mean AB kinematics with the closet inclination (25°) available in the dataset [24] to assess biomimicry. The calculated phase variable was compared against the ideal case (straight line from 0 to 1 over the gait cycle).

To quantify the kinematic errors of the prosthetic knee and ankle joints, we calculated the root mean square error (RMSE) with respect to the mean nominal kinematics. Because the phase-based controller allows the participants to progress at different speeds or even pause during each stride, phase shifts can occur throughout locomotion. Therefore, we calculated the RMSE over phase instead of normalized time. However, since the recorded phase variable did not necessarily sweep the full range from 0 to 1, the RMSE over phase only accounts for the errors between actual heel strike events. As a baseline for comparison, we also calculated RMSEs for individual AB trials and average nominal AB knee and ankle kinematics, which assumes the phase progresses ideally.

IV. RESULTS AND DISCUSSION

This section evaluates and discusses the biomimicry of the prosthetic kinematics observed during the different types of transitions. Limitations and future works are also discussed.

A. Continuous Transitions

The intra-subject mean kinematics for each activity in the two transition sequences are shown in Fig. 5. Overall, kinematics resemble AB data from [24], with a summary of RMSE means and standard deviations reported in Fig. 6. This is taken with respect to phase to mitigate errors arising purely from phase offsets which can occur due to possible variations in the rate of gait progression, which is made possible through the phase-based controller. For transitions led by the prosthetic leg, the mean kinematics trajectory gradually varies between walking and stair ascending, while the transitions led by the intact leg consist of the same kinematics as the two adjacent steady-state activities. Small discontinuities between activities in Fig. 5 arise from averaging strides for each activity type rather than plotting a continuous sequence of strides.

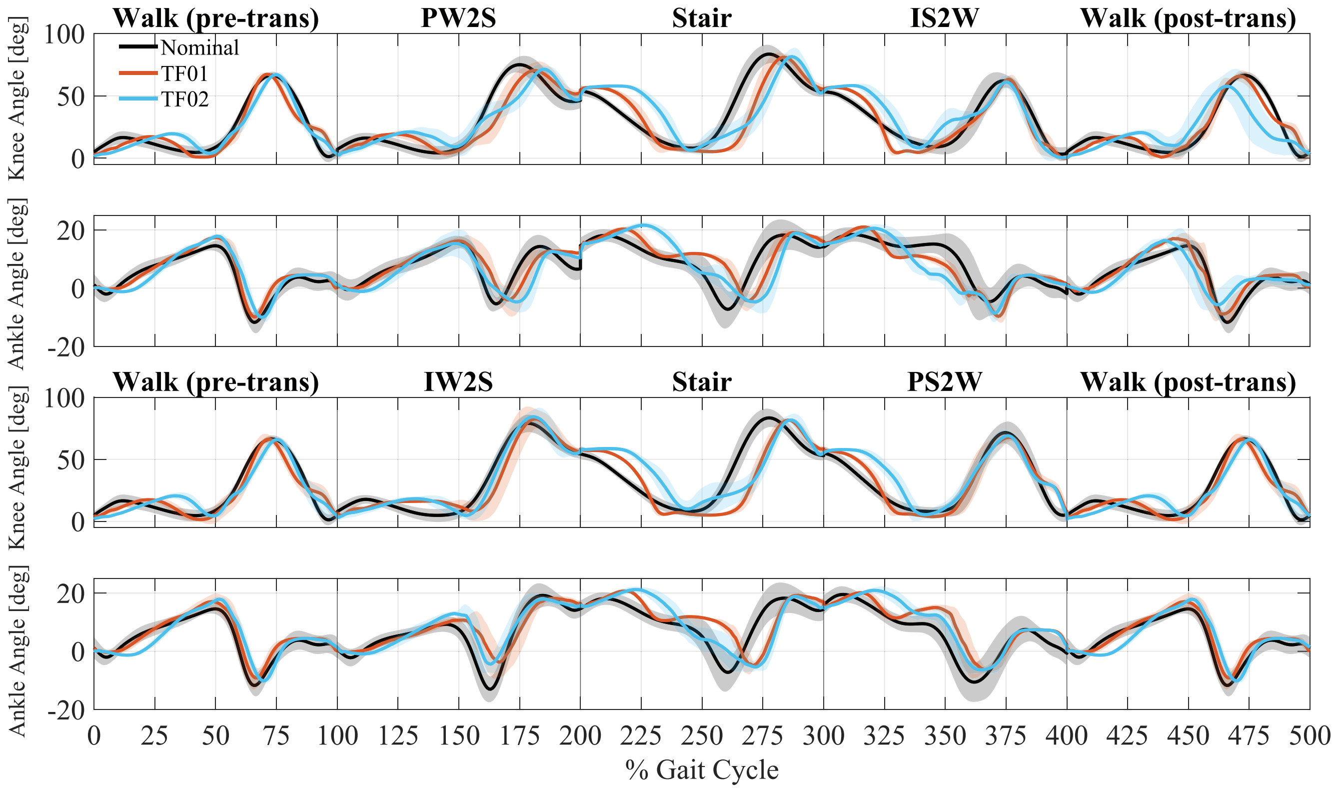

Fig. 5. Intra-subject mean kinematics for each activity in the two continuous transition sequences, including W2S and S2W transitions led by the prosthetic leg and the intact legs, compared to able-bodied data (nominal). Solid curves show the mean and shaded regions represent ± 1 standard deviation. Strides of the same activities were pooled before averaging for each transition sequence.

B. Walk to Stair Ascent Transitions

For the W2S transitions in Fig. 7, kinematics roughly resemble nominal AB values, with the change from walk to stair ascent stride being gradual in the PW2S transition and discrete in the IW2S transition. Deviations from AB data arise mainly from a phase lag in stance for PW2S and IW2S and in swing for just PW2S, shown in Fig. 7. In stance, the phase lag for both TF01 and TF02 results from the thigh angle progressing more nonlinearly than AB data. This is common with the chosen control strategy and was observed in our prior work for steady-state walking and stair ascent [27]. For both PW2S and IW2S, the lower values in the thigh angle at the beginning of the gait cycle are due to how TF participants tend to take a smaller step to prepare for the transition. Therefore, the thigh trajectories have a shallower slope than nominal, which contributes to stance phase lag in both kinds of transitions. Interestingly, the stance lag is more pronounced for IW2S than PW2S. This may be because the IW2S transition does not involve a gradual change between activities, and the desired thigh kinematics for phase calculation during stance comes from walking.

During swing in PW2S, the phase lag mostly occurs in the early-to-mid swing phase, possibly because neither participant uses their passive prosthesis to lead the transition to stair ascent. As a result of this inexperience, the thigh does not flex as much as nominal, causing both participants to experience a heel strike shortly after reaching a lower-than-expected maximum thigh angle. This can lead to the phase variable not reaching 1, particularly for TF01. Consequently, this causes a mismatch between knee and ankle angles compared to AB data by approximately 5° for PW2S, though it falls within one standard deviation of the nominal data.

Additional deviation in the kinematics of both PW2S and IW2S, not accounted for by non-ideal phase progression, is partially due to the kinematic model imperfectly matching the nominal transition kinematics. This can account for RMSEs up to 3.05° for the knee and 1.72° for the ankle in PW2S [21], and 2.02° for the knee and 5.67° for the ankle in IW2S. The remaining errors come from control tracking performance, which has a maximum RMSE of 5.26° for the knee and 2.60° for the ankle. The reported values in Fig. 6 are comparable to those seen in AB data with values being within one standard deviation. The only exceptions are the knee and ankle values from TF01 being lower than nominal values from AB data in PW2S. This suggests the controller for W2S transitions is capable of producing kinematics with errors that are at least comparable to AB data for the participants in this study. The likely reason for TF02 having higher RMSEs than TF01 in PW2S is that TF02 consistently checked for foot clearance before the prosthetic leg transitioned to the new terrain after toe off and tended to swing the thigh fast once the clearance was confirmed. This is reflected by TF02's thigh angle trajectory around toe off in Fig. 7 (top).

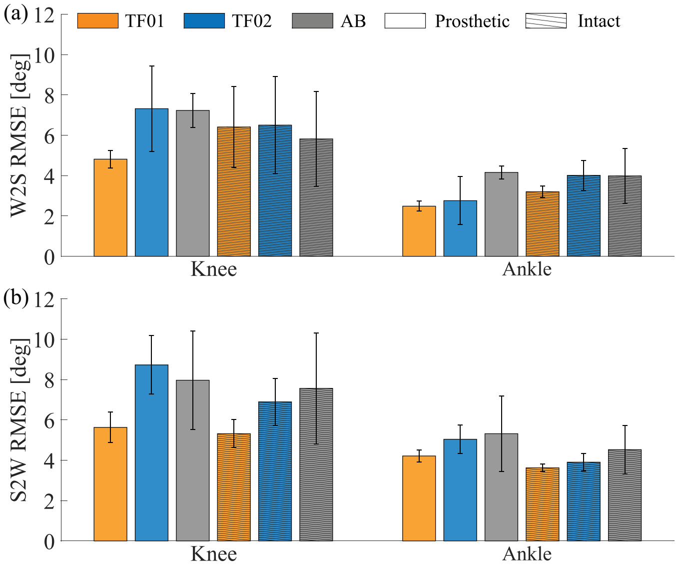

Fig. 6. Root mean square errors (RMSEs) of the transition kinematics (parameterized by gait phase) for knee and ankle compared to the mean able-bodied data for both TF01 and TF02 participants. RMSEs of AB data [24] are also calculated as a baseline to compare the performance of the transition controllers. Bars with and without shadows represent transitions led by the prosthetic and intact legs, respectively. The error bars show ± 1 standard deviation from the mean.

C. Stair Ascent to Walk Transitions

For the S2W transitions in Fig. 8, kinematics also roughly resemble nominal AB values. Deviations from the AB data primarily arise from a combination of phase lead and lag at various parts of the gait cycle, shown in Fig. 8. In both PS2W and IS2W, early stance exhibits a phase lag, and mid-to-late stance exhibits a phase lead. The early-stance phase lag and late-stance phase lead for both TF01 and TF02 results from the thigh angle at heel strike being greater than nominal, in combination with the thigh angle progressing quicker in most of stance. The higher thigh angle at heel strike may be attributed to a combination of the experimental stair incline ($26.5°$) being greater than the nominal dataset ($25°$) and more freedom in foot placement before IS2W. The quicker progression of the thigh angle may be explained by the shorter swing time of the sound leg observed in the experiment due to the limited training time with the powered prosthesis. In the second half of stance for IS2W, the two participants exhibit different behavior. A single minimum peak of the thigh angle is observed for TF01 but two peaks are observed for TF02 around 40 to 60% of the gait cycle. The double peaks can be explained by the phase lead during 30 to 60% of the gait cycle. This period of gait corresponds to stance when the controller uses the SSS model, which causes knee flexion around 60% of the gait cycle resulting in thigh flexion due to the kinematically closed chain [29]. These effects in combination result in phase matching approximately the nominal phase at toe off.

In swing for PS2W, kinematics and phase closely match AB data until late swing where there is a phase lag. This arises because the thigh continues to flex further than expected. In swing for IS2W, phase behaves similarly to normative values. Unlike in the W2S transitions, heel strike occurs when phase reached close to 1. The well-matched phase in PS2W in late stance and most of swing results in perfect alignment of both joints with the nominal kinematics for near 75% of the gait cycle in both participants.

Similar to the W2S transitions, deviations of PS2W and IS2W kinematics from nominal can be partly explained by kinematic modeling errors. For PS2W, the modeling RMSEs are up to $2.78°$ for the knee and $1.59°$ for the ankle [21]. For IS2W, the RMSEs are up to $4.90°$ for the knee and $4.92°$ for the ankle. Other errors come from control tracking performance, which has a maximum RMSE of $5.69°$ for the knee and $3.84°$ for the ankle. The values of RMSEs over phase in Fig. 6 for the S2W transitions are comparable to those seen in AB data, all within one standard deviation of the reference AB data. Again, this suggests the controller for S2W transitions is capable of producing kinematics with errors that are at least comparable to AB data.

D. Limitations and Future Work

As demonstrated in Fig. 6, TF02 exhibits greater RMSEs over phase than TF01 for both joints during W2S and S2W transitions. One possible explanation is that TF02 was unable to perform step-over-step stair ascent with his passive leg, which leads to significant compensations during stair ascent and transitions, such as vaulting from their intact leg and hip-hiking. Those compensatory behaviors were challenging to correct within the allotted acclimation time. In contrast, TF01 can ascend stairs in a step-over-step manner resulting in thigh motion more closely resembling that of able-bodied individuals, implying that longer training time can improve experimental outcomes by mitigating compensations.

Another notable observation is that with the exception of the knee joint of TF02, transitions led by the prosthetic leg generally exhibit lower RMSEs over phase than those led by the intact leg, thanks to the continuous transition models [21]. For intact leg leading transitions, the transition point at toe off may not be entirely accurate according to Fig. 2; it describes the transition in knee kinematics well but lags behind for the ankle. Choosing an earlier transition point between 40 to 50% of the gait cycle could enhance transition controller performance, but requires an accurate estimation of phase during stance. Furthermore, ankle kinematics during intact-led transitions differ significantly from both level walk and stair ascent across the entire gait cycle [20]. Hence, although both participants executed intact-led transitions comfortably and without balance loss, constructing a similar continuous kinematics model for intact-led transitions may improve controller performance and achieve better biomimicry.

Additionally, this experiment involved manual activity classification to validate the proposed mid-level controller, but clinical use necessitates a real-time classifier. It is also more difficult to detect a transition at the start of a transition stride than toward the end, as is usually done with switching-based activity transitions [14], [15]. Early transitions have previously been detected by incorporating environmental features collected by a laser scanning system [30] or a

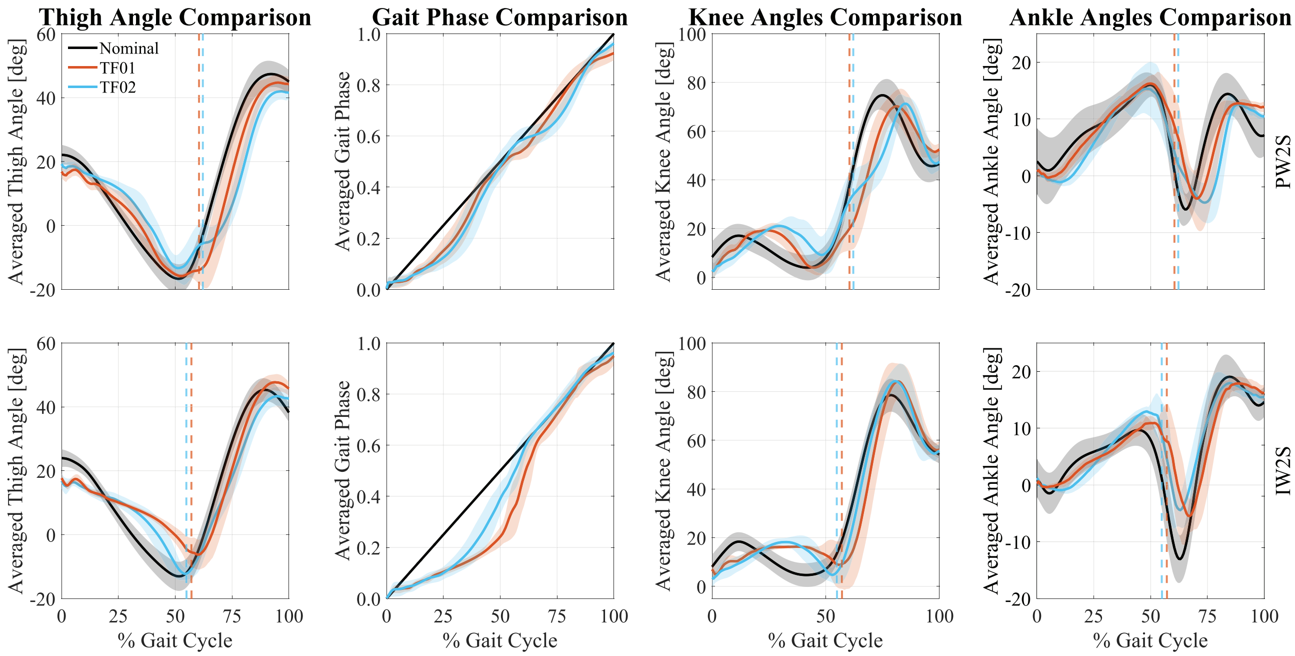

Fig. 7. Comparison between the recorded kinematics of the transition controller during walk to stair ascent transitions with the corresponding nominal kinematics from an able-bodied dataset [24]. Phase variables from the transition controller are compared against the ideal phase. PW2S (top) and IW2S (bottom) represent transitions led by the prosthetic leg and intact leg, respectively. Shaded regions indicate ± 1 standard deviation. The vertical lines demonstrate the average toe off phases for the corresponding data. Nominal toe off information is not reported in [24] for transitions.

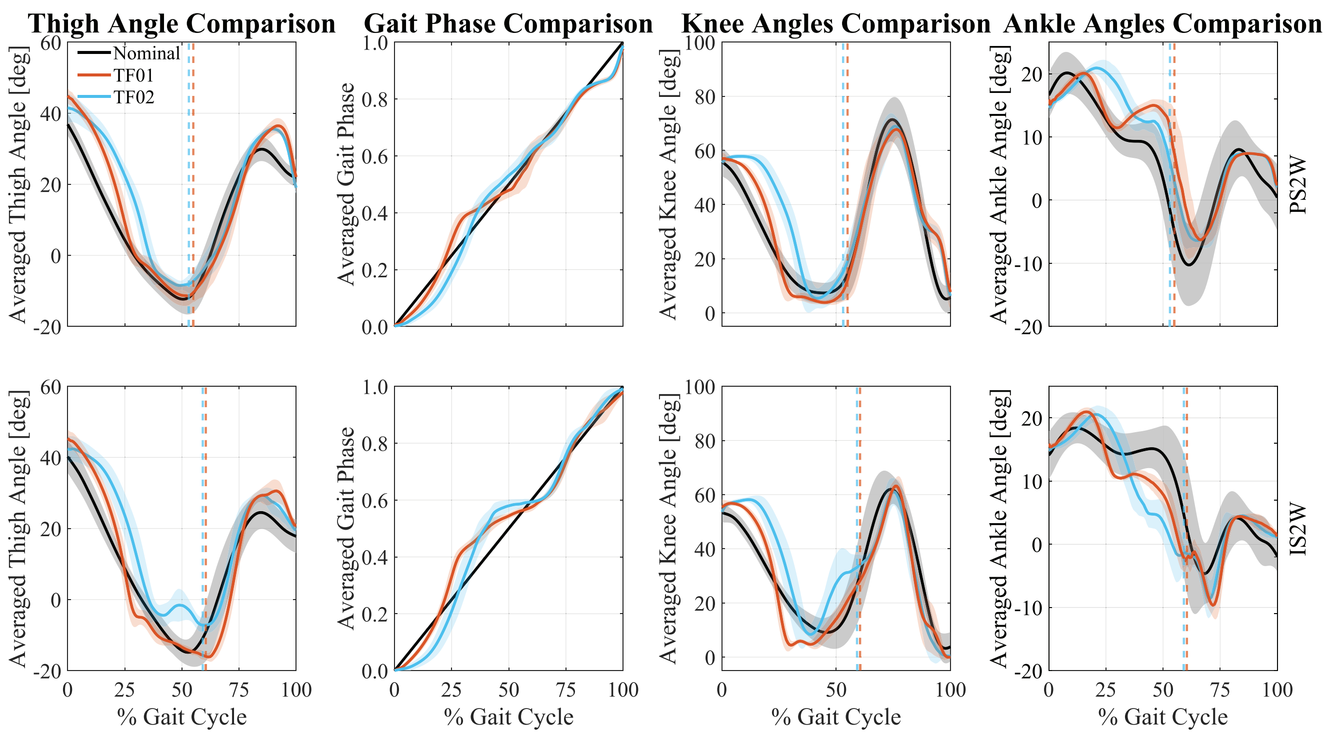

Fig. 8. Comparison between the recorded kinematics of the transition controller during stair ascent to walk transitions with the corresponding nominal kinematics from an able-bodied dataset [24]. Phase variables from the transition controller are compared against the ideal phase. PS2W (top) and IS2W (bottom) represent transitions led by the prosthetic leg and intact leg, respectively. Shaded regions indicate ± 1 standard deviation. The vertical lines demonstrate the average toe off phases for the corresponding data. Nominal toe off information is not reported in [24] for transitions.

depth camera [31], [32], which could trigger the proposed controller to achieve a smooth and seamless transition stride. Future work will implement a pre-transition classification method, perform experiments with more amputee participants on different stair inclines, and expand the proposed control framework to other activities such as stair descent.

V. CONCLUSIONS

This paper presented a continuously-varying, phase-based kinematic controller for a powered prosthetic leg that smoothly and seamlessly transitions between walking and stair ascending led by either the prosthetic or intact leg. We defined transitions led by either leg by analyzing the foot's elevation change between each case. A transition controller was created that accounts for the four different transition types based on models of joint kinematics. The transition controller was validated with two transfemoral amputee participants, demonstrating that the proposed controller can accomplish these transitions continuously and seamlessly with joint kinematics resembling nominal able-bodied data. We believe this work will eventually enable powered prostheses used by individuals with lower-limb loss to more naturally ambulate in the real world.

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