Minimum Jerk Swing Control allows Variable Cadence in Powered Transfemoral Prostheses

T. Lenzi, *IEEE Member*, L. J. Hargrove, *IEEE Member*, and J. W. Sensinger, *IEEE Member*

Abstract—**We present a novel swing phase controller for powered transfemoral prostheses based on minimum jerk theory. The proposed controller allows physiologically appropriate swing movement at any walking speed, regardless of the stance controller action. Preliminary validation in a transfemoral amputee subject demonstrates that the proposed controller provides physiological swing timing, without speed- or patient-specific tuning.**

I. Introduction

Powered prostheses have the potential to improve the walking ability of individuals with transfemoral amputations. Thanks to the use of battery-operated servomotors, powered prostheses allow positive net-energy tasks, such as step-over-step stair ambulation [1] and sit-to-stand transitions [2], while restoring more natural walking kinetics and kinematics compared to passive prostheses [3]. In stance phase, prosthesis torque can be regulated to obtain physiological body support and propulsion, [4] possibly reducing the metabolic cost of walking [5]. In swing phase, a biologically accurate movement can be generated to allow the timely placement of the foot in preparation for subsequent heel strike without requiring any additional effort from the user [6]. However, biologically appropriate control is necessary to exploit the full potential of motorized prostheses [7].

Passive transfemoral prostheses rely on passive dynamics to generate the swing movement [8], which can be performed at a constant or variable pace. However, propulsion of the swing movement is generated entirely by the user, who must pull the thigh forward at the end of stance (i.e., exaggerating the hip flexion torque) to initiate the swing movement. This unnatural action produces an asymmetric gait pattern [9].

Powered prostheses can overcome this limitation by mimicking the action of biological muscles to actively propel and control swing movement. However, attaining biologically accurate swing requires continuous adaptation of swing movement duration with walking speed and cadence [10], a function that no powered transfemoral prosthesis has yet provided.

Powered transfemoral prostheses largely rely on impedance-inspired control [3], an approach that does not allow direct regulation of swing duration. Impedance-inspired control defines joint torque as a parametric function of angle and velocity, with different stiffness, damping, and equilibrium values for each discrete phase of the gait cycle [11]. Swing duration therefore depends on the dynamic interaction of the prosthetic leg with the user and the environment during the swing phase, as well as on leg angle and velocity at the transition between stance and swing phase. Swing trajectory can be modified by regulating the impedance parameters of the prosthesis [12], though swing duration cannot be defined a priori (i.e., it is not a controlled parameter). Because impedance-inspired control needs user- and speed-specific tuning to obtain desired swing duration, variable cadence is hardly achievable.

To overcome this limitation, we propose a new control approach for swing phase that relies on a minimum jerk trajectory. Using this approach, we can obtain a biologically accurate swing movement with direct control of swing duration that is independent of joint angle and velocity at the stance-to-swing phase transition. Direct control of swing movement duration facilitates natural gait symmetry for any walking speed and cadence. Swing phase duration can be set to be proportional to stance phase duration at each step in order to restore the physiological relationship between the two phases of the gait cycle. Minimum jerk control can attain biologically appropriate swing movement without subject- or speed-specific tuning. Notably, we can enforce a desired maximum knee flexion in swing phase independent of walking speed and cadence, thus ensuring proper foot clearance in all conditions. Moreover, we can regulate the desired swing terminal angle as needed for walking up or down a ramp [13], again independently of swing movement duration, without any need for tuning.

In this paper, we present the design and implementation of the proposed swing controller on the Vanderbilt transfemoral prosthesis [12] (see Fig. 1), and preliminary evaluation of this controller in a transfemoral amputee subject walking on a treadmill at three different speeds. The experimental protocol comprised walking with the motorized prosthesis or a prescribed passive prosthesis. Experimental results showed that minimum jerk control allowed biologically appropriate swing movement by automatically adapting swing duration with walking speed and cadence. The subject improved swing timing when using the powered prosthesis compared to using his passive prosthesis. The

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Fig. 1. The Vanderbilt prosthesis.

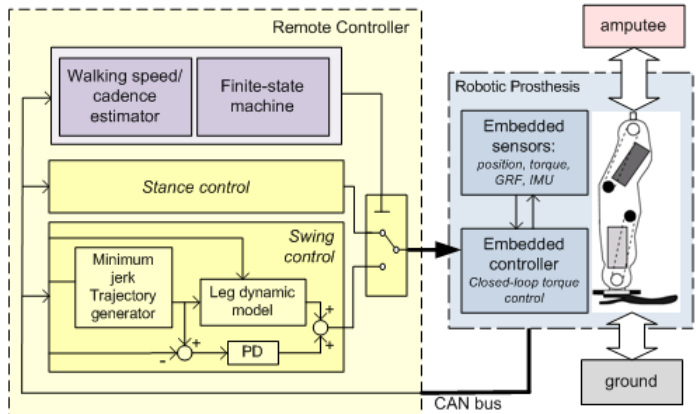

Fig. 2. Block diagram of the prosthesis controller.

swing trajectory was always smoother when using the powered device.

II. Methods

A. Controller design and implementation

The block diagram of the prosthesis controller is shown in Fig. 2. The overall control architecture comprises three stages: (1) identification of user and prosthesis status (i.e., walking speed/cadence estimator, finite-state machine); (2) planning of prosthesis joint torque (i.e., stance and swing phase controllers); and (3) attainment of desired torque in prosthesis joints (i.e., embedded closed-loop control). Hereafter, we provide a detailed description of the swing controller only, which is the focus of this paper. Additional details on the overall controller can be found in [14].

The swing phase controller enforces a minimum jerk position trajectory that approximates the behavior of an intact leg at different walking speeds. As the finite-state machine enters into swing mode (i.e., prosthetic foot off the ground), the swing phase controller computes the coefficients of a 5th order polynomial function that generates the desired minimum jerk position trajectory. The duration of the ankle and knee movement during swing phase is set to be 0.30 and 0.45 times the previous stance duration respectively, based on able-bodied biomechanics [10]. The initial joint angle and velocity of the trajectory are equal to their respective measured values at the end of stance phase. The final joint angle, velocity, and acceleration are set to zero for both the ankle and knee joints. Whereas a unique minimum jerk trajectory is used for the ankle joint, the knee trajectory comprises two parts. The first starts with the knee angle and velocity measured at the end of stance phase and ends at the point of maximum knee flexion with zero velocity. The second part starts from the maximum knee flexion and ends with the knee fully extended and zero velocity and acceleration. The acceleration at maximum knee flexion as well as the starting acceleration of the ankle joint was optimized based on able-bodied data. The maximum knee flexion angle is regulated based on user anthropometry to ensure an appropriate foot clearance despite the fixed shank length of the prosthesis. The desired angular trajectory is enforced by relying on a strong feed-forward torque command and a weaker feedback position control with proportional and derivative terms. Whereas the feed-forward command accounts for the inertial, gravitational, and frictional components, the feedback loop allows accommodation to the contingent disturbances that occur during swing phase movement, and, in addition, compensates for possible inaccuracies of the prosthesis dynamic model.

The proposed control framework has been implemented on a self-contained ankle and knee prosthesis previously developed at Vanderbilt University [12]. This prosthesis is battery operated and uses brushless DC motors to deliver biomechanically appropriate torque and power at the knee and ankle joints. An embedded control system runs the closed-loop torque controllers for the ankle and knee joints. A remote computer using a hard real-time operative system (xPC target, Mathworks, USA) runs the algorithms for the estimate of gait phase and walking speed, as well as the stance phase and swing phase controllers. Communication between the embedded and remote systems is handled by a high-speed CAN bus (CAN-AC2-PCI, Softing, USA). Communication, processing, and data recording run on the remote control system at the fixed sampling rate of 1 KHz.

B. The experimental procedure and data analysis

As a preliminary evaluation, we tested the proposed controllers on a transfemoral amputee patient (30 years old, 1.86m, 86.2 Kg). The experimental protocol was approved by the Northwestern University Institutional Review Board, and the participant provided informed consent before the experiment took place. A certified prosthetist fit the subject with the Vanderbilt prosthesis. The subject then practiced walking with the prosthesis on a treadmill for about 30 minutes at different speeds. After this familiarization phase, we assessed the self-selected speed, which was 0.85 m/s and defined the low and high speed for the main experiment as 0.70 and 1.0 m/s respectively. The patient performed three two-minute sessions at each previously selected walking speed, with at least two minutes of rest between each session. The subject then repeated the test using his prescribed prosthesis (an Elite blade foot and a KX06 knee, Endolite, Miamisburg, OH), to which we added sensors—electro-

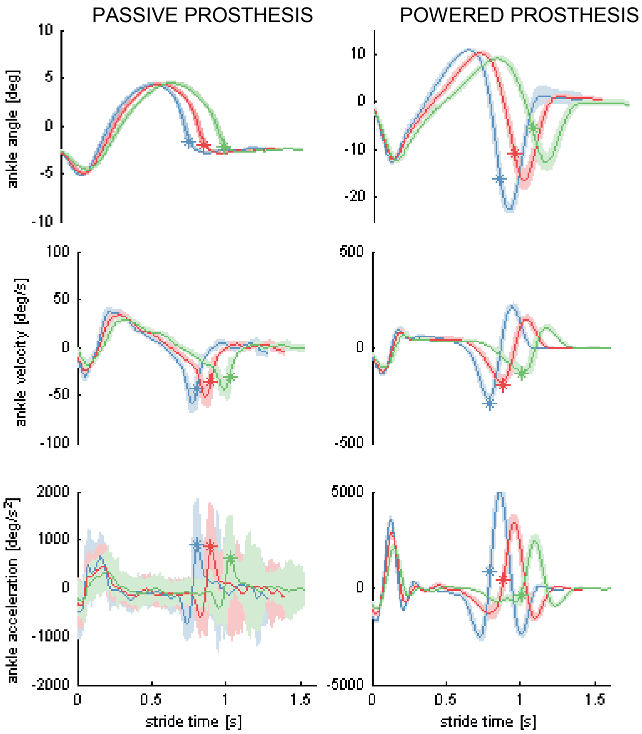

Fig. 3 Ankle kinematics for the three different walking speeds

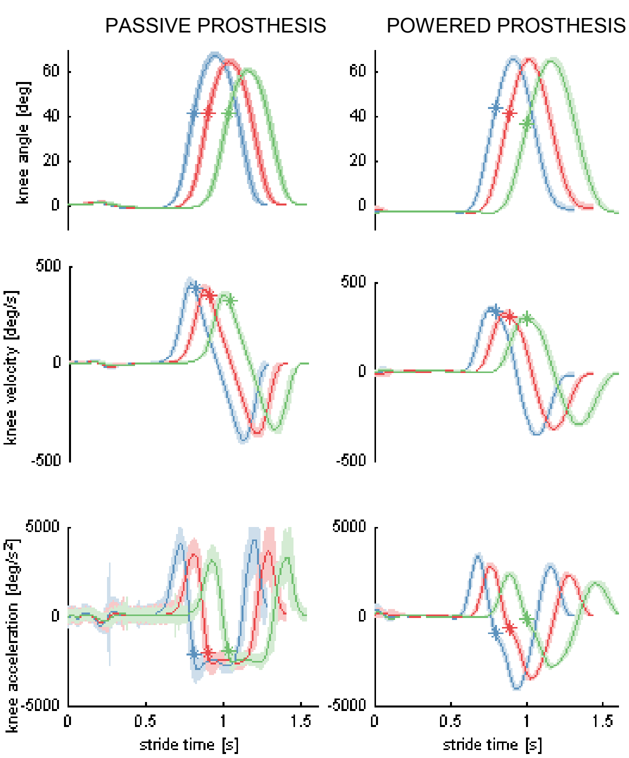

Fig. 4. Knee kinematics for the three different walking speeds.

mechanical goniometers and a foot-switch sensor—to record ankle and knee joint angle, as well as heel and toe contact with the ground.

Prosthesis angle profiles were recorded using the sensors located on the prosthesis Joint velocity and acceleration were obtained in post-processing. To attenuate the sensor noise for proper data analysis, we low-pass filtered all data using a back and forth low-pass first-order Butterworth filter with cutoff frequency of 10 Hz. For each walking speed, we separated raw data into strides (i.e., the time interval between two consecutive heel-strike events on the prosthesis side) using the output of the local ground reaction force sensor for the robotic prosthesis and the foot-switch sensors for the passive prosthesis. Within each stride, we computed the duration of stance-phase, swing-phase, and stride. The first and final three strides for each walking session were omitted from the analysis to avoid including non-steady state walking. Finally, we computed the angle, velocity, and acceleration profiles for the ankle and knee joint averaged over all the steady-state strides recorded at each constant walking speed. Only the third repetition for each walking speed was considered in the analysis, to avoid adaptation effects. All data processing was performed using Matlab (The MathWorks, Natick, MA, USA).

III. Results and Discussion

Fig. 3 and Fig. 4 show the angle, velocity, and acceleration profiles averaged over all the strides recorded at the same walking speed for the ankle and knee, respectively. Solid color lines indicate the averaged profiles; shaded areas represent +/- one standard deviation. Different colors indicate different walking speeds. Markers show the average transition times from stance to swing phase.

The powered ankle kinematics largely differed from those of the passive ankle (Fig.3). During stance phase, the passive ankle was generally stiffer than the powered ankle and did not provide plantarflexion movement in late stance. This difference is due to the stance phase controller, and thus is not further discussed in this paper. Importantly, the powered ankle movement in swing phase was automatically adapted to walking speed in order to complete the dorsiflexion movement in a physiologically appropriate time. The dorsiflexion movement was completed in a shorter time at higher walking speeds, despite the increased plantarflexion angle and velocity at the transition between stance and swing phase, which was due to speed-dependent action of the stance controller.

Knee kinematics (Fig. 4) were also significantly different between the passive and powered prostheses. The maximum knee flexion angle was independent of walking speed for the powered prosthesis, whereas it increased with walking speed for the passive prosthesis. This indicates that the passive prosthesis failed to fully compensate for the increased momentum of the prosthetic leg, caused by a higher knee flexion speed at the start of swing phase.

Focusing on knee extension, we noted that the powered prosthesis completed the swing movement (i.e., knee velocity reached zero) equally in advance of the end of swing phase for all walking speeds. On the other hand, with the passive prosthesis, the time lapse between the end of knee extension and the end of swing phase varied with walking speed: Whereas at slow speed (green line), the knee extension movement was completed well in advance of the

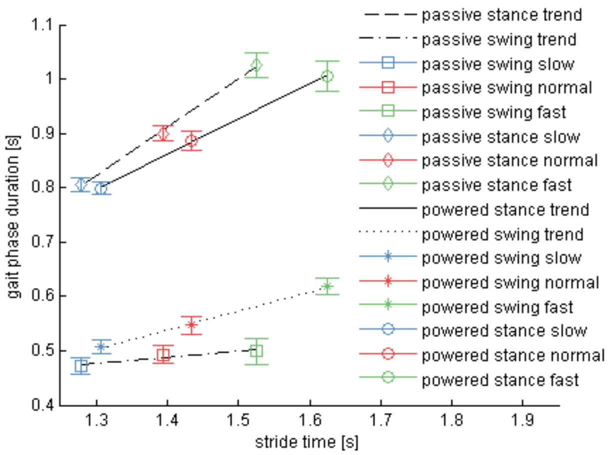

Fig. 5. Stance and swing phase duration as a function of stride duration.

end of swing phase (i.e., the subject waited with the prosthetic knee fully extended before contacting the ground), at the highest speed (blue line), the knee extension coincided with the end of swing phase (i.e., knee velocity just reached zero when the foot contacted the ground). This analysis indicates that using the passive prosthesis provided a much more limited control of swing movement; the fastest possible swing movement was reached at the highest speed of the test (i.e., 1.0 m/s).

The averaged knee velocity peaks were slightly smaller for the motorized prosthesis. In the first part of swing phase, the passive prosthesis had a constant negative acceleration that decelerated the initial knee flexion movement and accelerated the subsequent knee extension movement. This negative acceleration was equal for all walking speeds. Toward the end of swing phase, the acceleration became positive, showing a bell-shaped profile with a peak proportional to walking speed. The powered prosthesis showed instead a smoother acceleration trajectory, with a bell-shape profile during both the negative and positive acceleration phases and peaks proportional to walking speed. A smoother swing was obtained for all walking speeds using the powered prosthesis.

Fig. 5 shows the stance and swing phase duration for both passive and powered prosthesis for all walking speeds. When walking with the powered prosthesis, the stride duration was longer, though the difference decreased with walking speed (0.10, 0.04, and 0.02 s for 0.7, 0.85 and 1.0 m/s, respectively). This indicates that the subject took longer steps with the powered prosthesis, possibly better approximating able-bodied behavior [10]. With the powered prosthesis, swing duration was 38.3%, 38.2%, and 38.4% of stride duration for high, normal, and low walking speed, respectively. In contrast, swing duration with the passive prosthesis equaled 36.7%, 35.2%, and 32.8% of stride duration for the same three walking speeds, respectively. The proposed controller achieved physiological swing duration regardless of the walking speed, outperforming the passive device. In contrast to impedance-inspired control, the proposed controller achieved biologically accurate stance and swing timings at any speed without the need for tuning.

IV. Conclusion

In this paper, we present and validate a novel control algorithm for the swing phase of a motorized prosthesis. Using a simple principle of minimum jerk, it was possible to provide direct control of swing movement duration. This allowed us to set a simple rule to normalize the stance-swing proportion inside each gait cycle, regardless of the walking speed. Experimental results showed that this simple control improved swing timing in a transfemoral amputee using the powered prosthesis when compared to using a passive prosthesis.

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