Proportional EMG Control of Ankle Plantar Flexion in a Powered Transtibial Prosthesis

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***Abstract*—The human calf muscle generates 80% of the mechanical work to walk throughout stance-phase, powered plantar flexion. Powered plantar flexion is not only important for walking energetics, but also to minimize the impact on the leading leg at heel-strike. For unilateral transtibial amputees, it has recently been shown that knee load on the leading, intact limb decreases as powered plantar flexion in the trailing prosthetic ankle increases. Not surprisingly, excessive loads on the leading, intact knee are believed to be causative of knee osteoarthritis, a leading secondary impairment in lower-extremity amputees. In this study, we hypothesize that a transtibial amputee can learn how to control a powered ankle-foot prosthesis using a volitional electromyographic (EMG) control to directly modulate ankle powered plantar flexion. We here present preliminary data, and find that an amputee participant is able to modulate toe-off angle, net ankle work and peak power across a broad range of walking speeds by volitionally modulating calf EMG activity. The modulation of these key gait parameters is shown to be comparable to the dynamical response of the same powered prosthesis controlled intrinsically (No EMG), suggesting that transtibial amputees can achieve an adequate level of powered plantar flexion controllability using direct volitional EMG control.**

Keywords—ankle-foot prosthesis; powered prosthesis; EMG; volitional control; proportional control

I. INTRODUCTION

Dynamic walking models have illustrated the importance of calf-muscle activation in the trailing leg during human walking. Pushing off just before toe-off, or powered plantar flexion, decreases the collision cost at heel-strike in the leading leg, and keeps the metabolic cost at a minimum [2]. Powered plantar flexion is hence not only important to conserve energy but also to minimize mechanical loading borne by the leading leg. Leg amputees using ankle-foot prostheses that are incapable of normative levels of powered plantar flexion are at a disadvantage, as the biological calf-muscles generate nearly 80% of the mechanical work required to complete each gait cycle [3, 4]. Further, the loading of the leading leg at heel-strike determines the peak knee external adduction moment (EAM). In the general population, excessive levels of EAM at heel-strike has been linked to the development of knee osteoarthritis (OA) of the leading limb, a common secondary impairment in lower-extremity amputees [5]. It has recently been shown that powered plantar flexion minimizes knee EAM. In this study, the larger the prosthetic ankle-foot push-off work was, the lower the 1st peak of the knee EAM became [6].

Furthermore, research on ankle-foot orthoses has illustrated the importance of intrinsic versus extrinsic control method. Participants using proportional electromyographic (EMG) control to activate artificial pneumatic muscles performed closer to their natural powered plantar flexion movements than when using an intrinsic foot-switch control [7], and were able to reduce their net metabolic power using these exoskeletons in a laboratory environment [8]. Current ankle-foot prostheses are capable of biomimetic behavior using intrinsic controllers during locomotion at steady speeds [9]. Adding a volitional EMG controller may yield important advantages such as generalizing biomimetic behaviour of the prosthesis to transitional walking speeds and incorporating sensorimotor feedback from the residual limb. Furthermore, continued use and activation of the residual limb muscles may prevent muscle atrophy and a decrease in limb volume, which directly affects the prosthetic socket's goodness of fit.

In this study, we hypothesize that a transtibial amputee can learn how to control a powered ankle-foot prosthesis using a volitional EMG control to directly modulate ankle powered plantar flexion throughout the mid to late stance period. We anticipate that a transtibial amputee can control the push-off response of the powered prosthesis volitionally, using a hybrid controller that combines proportional EMG control with its intrinsic controller, and achieve the same level of controllability as the purely intrinsic controller in level ground walking. As a preliminary evaluation of this hypothesis, we monitor prosthetic powered plantar flexion on a transtibial amputee at three walking speeds (1, 1.25, 1.5m/sec) using both the hybrid and intrisic control paradigms.

II. STATE MACHINE AND CONTROL SCHEME

A. Intrinsic Controller

A finite state machine synchronized to the gait cycle of the participant implements top-level control of the prosthesis. A typical gait cycle for level ground walking is defined as starting with the heel strike of one foot and ending with the next heel strike of the same foot. A complete gait cycle consists of two phases: stance and swing. The stance phase

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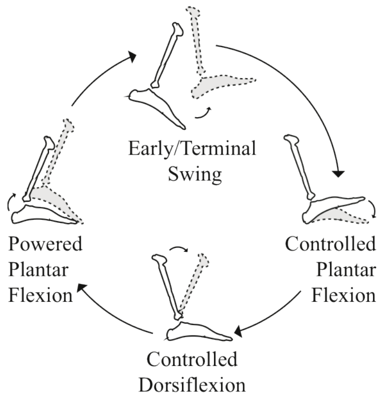

Fig. 1. States of the intrinsic controller that are monitored by the EMG state machine.

begins at heel strike and terminates upon toe off; the swing phase takes up the remainder of the cycle. Each phase can be further divided for better characterization. This paper adapts the same convention used by Au et al. [10]. As illustrated in Fig. 1, the stance phase consists of three sub-phases: Controlled Plantar Flexion (CP), Controlled Dorsiflexion (CD), and Powered Plantar Flexion (PP). The swing phase is divided into early swing state (ESW) and terminal swing state (TSW). For a formal definition and detailed analysis of each stage of the gait cycle please refer to Au et al. [10]. What is particularly pertinent to this paper is that during PP, the ankle performs net positive work that is equal to or greater than the work absorbed during CP and CD.

The intrinsic controller used in the current paradigm has previously been used for the BiOM bionic ankle-foot prosthesis. While amputees wearing commercially available passive-elastic prostheses require 10-30% more metabolic energy to walk at the same velocities as non-amputess it has been shown that the BiOM is capable of normalizing walking gait and metabolic cost using this intrinsic controller [9].

B. Powered Plantar Flexion

The PP state is important as it emulates the biological calf-muscles, which generate nearly 80% of the mechanical work required to complete each gait cycle. The push-off provided before toe-off adapts to the participants walking characteristics. An increase in the sensed prosthetic ankle joint torque triggers an increase in the torque generated by the actuator during mid- to late stance phase, resulting in an increase in net positive ankle work production. When the intrinsic controller is used, the gain term (P_{FF}) in equation 1 depends on the walking velocity. When proportional myoelectric control is used the gain componenet is linearly dependent on the EMG signal. During late stance PP, when the ankle begins to plantar flex a spring function is applied, having stiffness K and eqilibrium θ_0 to ensure a biomimetic plantar-flexion toe-off angle.

$$\tau_{Pff} = P_{FF} * \tau^3_{measured} + K_0(\theta - \theta_0) \qquad (1)$$

C. Proportional Myoelectric Controller

Since the intrinsic controller is able to emulate biological ankle behavior it continuously controls the ankle unless interrupted by the extrinsic EMG controller. The purpose for the EMG controller is to modulate the gain parameter of torque commanded on the powered ankle during level ground walking. The EMG controller is used to interrupt the intrinsic controller and to take over control of the ankle as described below. Thus the state-machine is built to monitor and shadow four essential states of the intrinsic controller. Fig. 1 shows the state-machine used on the controller.

The diagram should be read starting from controlled plantar flexion (CP), which is defined as ranging from heel strike until foot flat, cf. Fig. 4. The EMG controller is idle during CP, upon arriving at CD the EMG controller starts to record EMG from the residual limb muscle (lateral Gastrocnemius) of a transtibial amputee. The processed EMG amplitude is then linearly mapped to the gain parameter overriding the value set by the intrinsic controller, P_{FF}, in equation 1. This gain parameter determines the amount of torque to command from the powered ankle motor using a positive torque feedback control paradigm. Thus, the EMG signal is used to modulate the gain, P_{FF}, or sensitivity, of the positive torque feedback control (equation 1) which changes the amount of torque exerted by the ankle during PP. After PP, the ankle enters the swing state, which includes both ESW and TSW. The cycle repeats as the ankle enters CP. For more information on the intrinsic control, see [9].

III. Hardware

A. BiOM Ankle-Foot Prosthesis

The ankle–foot prosthesis used for this study is developed by iWalk, LLC. This prosthesis is a successor to the series of prototypes developed in the Biomechatronics Group of the MIT Media Laboratory. It is a completely self contained device having the mass (1.8 kg) and size of the intact biological ankle–foot complex. The basic architecture of the electromechanical design is depicted in Fig. 2. It consists of a unidirectional spring in parallel to an actuator with a series spring [1, 11]. The prosthesis is capable of varying impedance during stance, providing power during powered plantar flexion and performing PD control during swing.

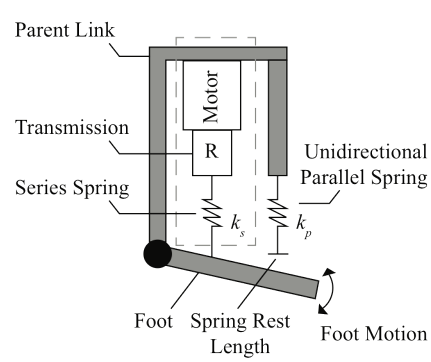

Fig. 2. Schematic mechanical design of the powered ankle-foot prosthesis (adapted from Au et al. [1]).

B. EMG Measurement Unit

The EMG measurement module was designed and implemented in the Biomechatronics group. At the input stage the module uses a commercially-available pre-amplifier designed by Motion Lab Systems with a pick up gain of 20. Due to the lack of physical space between the participant's residual limb and the socket, the pre-amplifier was not directly connected to the gastrocnemius muscle. Instead a customized liner (Alps, St Peterstburg, Florida, USA) with integrated, conductive fabric electrodes was used. This technique was developed at Northwestern University. The pre-amplifier is attached to the fabric leads above the socket. A 32bit analogue-digital converter (ADC) and a microcontroller (Pic32MX575F512H) were used to process the signal. An onboard IEEE 802.11g wireless radio streamed the data over a local wifi network.

IV. EMG Signal Processing

The EMG signal is sampled at 1.5kHz and later down-sampled to 500Hz on the microcontroller. A standard preprocessing method is used to obtain the EMG profile [12]. The raw signal obtained from the pre-amplifier is low-pass filtered with f_{cutoff}=15Hz, rectified by taking the absolute value, and then normalized by the maximum muscle contraction (MVC) measured before experimentation. A moving window (200ms) is used to calculate the moving average of the preprocessed EMG signal. The MVC was measuered in a standing position as this signal was stronger in the residual muscle than in a sitting position. This is most likely due to an improved contact between the electrode and the weightbearing residual limb.

A. Threshold Detection

For proportional torque control, a threshold value is needed to distinguish between the EMG signal measured due to voluntary muscle contraction and the baseline noise due to motion artefacts. The participant is therefore asked to walk 10 steps each with and without flexing the gastrocnemius during controlled dorsiflexion. In these trials the ankle-foot prosthesis is controlled by the intrinsic controller. The threshold represents the lowest EMG amplitude to best separate the two conditions.

V. Experiments

A preliminary clinical study was conducted to evaluate the performance of the proposed EMG-driven, finite state controller. One participant, a bilateral transtibial amputee tested the device (male, weight: 81.9kg, height: 1.89m). The participant wore the powered prosthesis on his right leg and a conventional passive prosthesis on his left leg. Initial walking experiments were conducted in the Biomechatronics Group within the MIT Media Lab. MITs Committee on the Use of Humans as Experimental Subjects (COUHES) approved the study. The participant volunteered and was permitted to withdraw at any time and for any reason. Before partaking in the study, the participant read and signed a statement acknowledging his informed consent.

The participant performed two experimental blocks of level-ground walking. Each block was recorded on a separate day and consisted of 7 walking trials of 10 steps each (35 gait cycles) at three different speeds: slow (1m/s), medium (1.25m/s) and fast (1.5m/s). For the first block, the participant was instructed to walk without flexing his residual limb muscles. In this 'idle' condition the unintentional EMG activity did not interfere with the EMG controller, leaving control of the prosthetic ankle to the intrinsic controller. In the second block, the participant was asked to consciously flex his residual limb muscles during controlled dorsiflexion to modulate the gain of the powered plantar flexion torque.

Ankle angle, torque and state machine transitions were recorded on the powered prosthesis and sampled at 500Hz. All data were subsequently parsed into gait-cycles starting at heel-strike, interpolated and downsampled to 1000 data points per gait cycle. Only trials within 5% of error (walking speed) were accepted. Ankle net work was calculated by integrating ankle torque (in Nm) with respect to ankle angle (in radians) per each parsed gait cycle. Ankle power was calculated by taking the time derivative of the calculated ankle work. Further, toe-off angle, net ankle work, and peak ankle power were recorded for each gait cycle. The ensemble average and standard deviation were then calculated for all gait cycles and walking speeds.

VI. Results and Discussion

The intrinsic controller on the powered ankle-foot prosthesis had previously been shown to display biomimetic behaviors during steady level-ground walking [9]. Here we report preliminary data illustrating that the participant was able to achieve comparable behavior using the EMG controller.

A. Comparison to Intrinsic Controller

1) Ankle Angle at Toe-Off

Fig. 3a shows the prosthetic ankle angle at toe-off over the different walking speeds. The ankle angle from the hybrid control experiments closely resembles that of the intrinsic controller across the tested speeds. Comparable to non-amputee biomechanics, the prosthetic ankle angle at toe-off is shown to increase with increasing walking speed [13].

2) Net Work

The average net work per gait cycle and speed is shown in Fig. 3b. As for biological limbs, the two data sets show that ankle work increases with increasing walking velocity. Both the intrinsic controller and the hybrid EMG controller exert a similar amount of work for the three speeds tested. The net work of the hybrid controller is within one standard deviation of the intrinsic controller.

3) Peak Power

Fig. 3c shows the average peak power for the prosthesis across the three walking speeds. The profile of the hybrid controller closely resembles that of the intrinsic controller, the former again lying within one standard deviation of the latter.

4) Ankle Joint Torque vs Ankle Angle

Ankle torques versus ankle angles, as measured from the prosthetic ankle, using the hybrid controller and the intrinsic

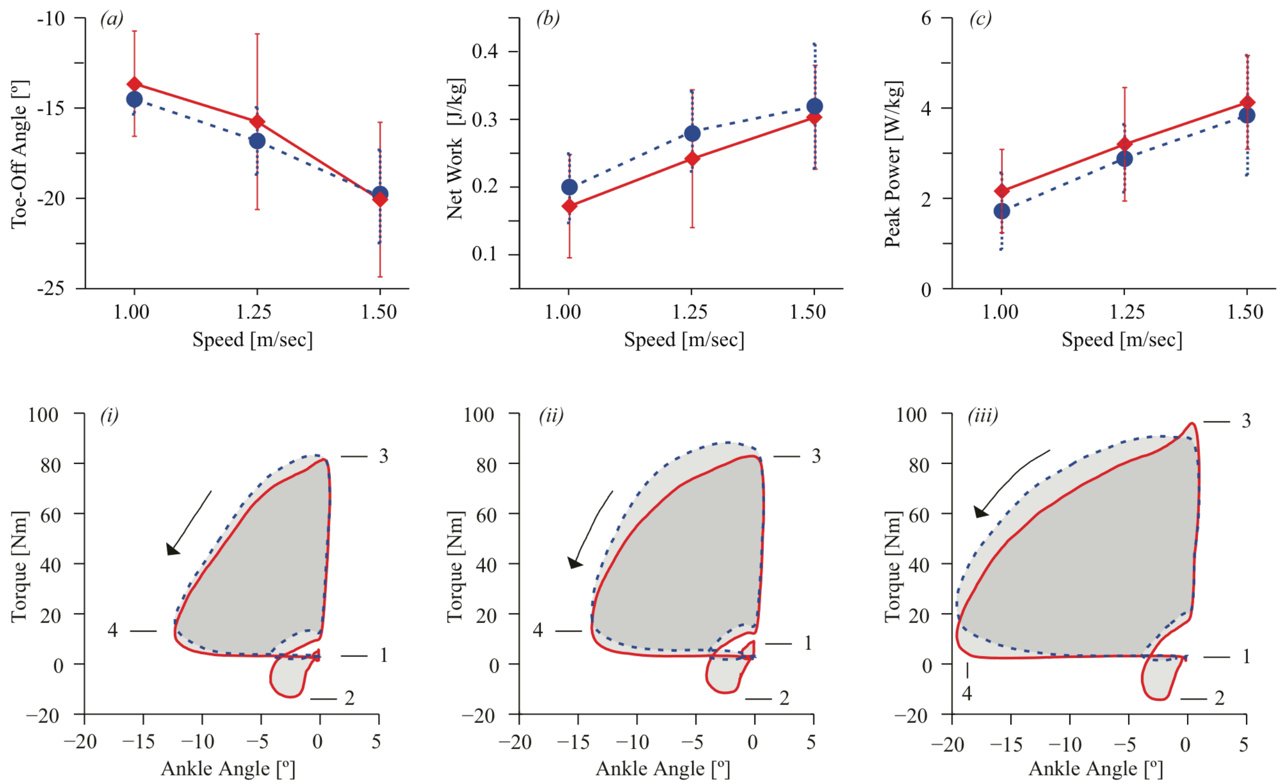

Fig. 3. Ankle angle, work, peak power and torque versus angle across walking speeds. (a) The average ankle angle at toe-off and (b) the ankle work per step during the stance phase of walking using the hybrid controller (solid, red line) are within 1 standard deviation (sd) of the mean data for the intrinsic controller (dotted, blue line, error bars indicate one sd). (c) The peak powers are also within one sd across all tested walking speeds. Below - Preliminary data illustrating the prosthetic ankle joint torque versus ankle angle. *(i)* 1.00m/s *(ii)* 1.25m/s *(iii)* 1.50m/s. The grey shaded area represents the positive work provided by the prosthesis. 1 – heel strike, 2 – foot flat, 3 – maximum dorsiflexion, 4 – toe-off.

controller are shown in Fig. 3, panels i-iii, for the three walking speeds, respectively. As before, there is a quantitative resemblance between all measurements, showing toe-off angle and net work trending with walking speed in a comparable manner between the controllers.

B. Comparison to Biological Limb

Table 1 provides a direct comparison between the dynamic behaviour of the prosthesis using the two controllers under investigation and the biological ankle-foot complex at 1.25m/s. Biological norm for each quantity are provided for direct comparison. Toe-off angle and net work were taken from Herr and Grabowski [9] and peak power from Winter [13]. Toe-off angle and peak power are within one standard deviation of the biological limb data. This indicates that biomimetic behaviour may be achieved using the hybrid EMG controller. In this preliminary data set the net work performed by the participant was higher than that of the biological norms. This is most likely due to an over-tuning of the gain parameter (P_{FF}) in (1), which was further modulated by the EMG signal. In a future investigation, as a resolution to this difficulty, the gain for the positive torque feedback will be carefully tuned until there is quantitative agreement with normative data.

TABLE I. DYNAMIC BEHAVIOR OF THE POWERED PROSTHESIS.

Controller	Toe-off Angle [°]	Ankle Net Work [J/kg]	Peak Power [W/kg]
Intrinsic (sd)	17 (2)	0.28 (0.06)	3 (1)
Hybrid	16 (5)	0.24 (0.10)	3 (1)
Biol. Norm	18 (3)	0.10 (0.05)	3 (1)

In summary, the preliminary data presented here illustrate that the participant was able to reliably use the hybrid EMG controller in concert with the existing intrinsic controller during level ground walking. The net work exerted and the toe-off angle, ankle torque, and power profiles closely resemble those of the intrinsic controller, which has previously been shown to produce biomimetic behavior during level-ground walking [9]. Further studies are needed to confirm the current findings and perform a full evaluation of the hybrid controller.

VII. CONCLUSION

Proportional EMG control of powered plantar flexion provides users the benefit of volitonal control over their prosthesis during mid to late stance. This includes walking at a steady speed, as indicated by the preliminary data, but readily extends to a range of walking transitions. The hybrid controller has the potential of restoring walking velocity-related feedback by allowing the user to tune the torque gain through flexing the corresponding residual limb muscle. This approach would perhaps allow control of the prosthesis to be at least partially integrated with the biological sensorimotor control loop enabling signals from neuromuscular reflexes as well as supra-spinal locomotor regions to influence prosthesis control. Future studies are required to determine if the hybrid EMG controller can improve the participant's metabolic cost of walking and minimize the impact at heel-strike on the leading, biological limb and thus prevent the accelerated development of knee osteoarthritis in lower-extremity amputees.

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