

PERFORMANCE OF A POWERED ANKLE EXOSKELETON USING NEUROMUSCULAR MODEL-BASED CONTROL OVER A RANGE OF WALKING SPEEDS

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INTRODUCTION

Powered exoskeletons can improve locomotion performance by providing assistive torques to offset biological moments generated by lower-limb joints. So far, the majority of exoskeleton controllers rely on *a priori* tuning of parameters based on a reduced set of locomotor tasks (*e.g.* level walking at preferred speed). This limits their effectiveness to a small subset of functional conditions. To be useful in the real-world, exoskeleton controllers need to be able to automatically adjust to the user and the environment.

The purpose of this study was to explore whether ankle exoskeleton control based on a neuromuscular model could adapt to the increasing mechanical demands associated with increased walking speed (*i.e.*, increased ankle power).

A neuromuscular model (NMM) based controller simulating a Hill-type muscle-tendon with positive force feedback has previously been implemented in an ankle-foot prosthesis [1]. By emulating physiological neuromuscular mechanics, the prosthesis adapted to variable gait speed by way of modulating net work output at each speed. To date, the performance of NMM based controllers has yet to be explored in lower-limb exoskeletons.

We hypothesize that a powered ankle exoskeleton employing a NMM-based controller would automatically adapt to walking speed by 1) increasing net work at increased speeds and 2) maintain reductions in metabolic cost across speeds.

METHODS

Exoskeletons: We provided bilateral plantarflexion assistance via a tethered ankle exoskeleton device. Off-board motors and Bowden cable transmission delivered torque to the custom carbon fiber exoskeletons.

Controller: We designed the NMM based controller in Simulink (MathWorks, USA) and implemented it through a dedicated real-time control system which

also handled signal IO (500 Hz, dSPACE, Germany). We calculated applied exoskeleton torque from a load cell (500 Hz, LCM Systems Ltd, UK) placed in series with the applied assistance, and a goniometer (500 Hz, Biometrics, UK) attached to the exoskeleton joint provided real-time ankle angle.

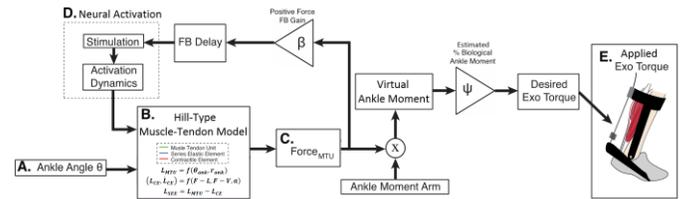


Figure 1: Neuromuscular model controller block diagram. Biological force, as computed by the internal muscle-tendon state, is scaled and applied across a moment arm to calculate desired exoskeleton torque.

The NMM based controller implemented in our system was similar to a reflex-based force feedback controller previously demonstrated [1]. A Hill-type muscle-tendon model was the basis for the emulated muscle-tendon unit (MTU) consisting of a contractile element (CE) with both active and passive properties and a series elastic element (SEE) (Fig.1B). We calculated MTU length from ankle angle and musculoskeletal geometry (1.1) and CE dynamics was subject to force-length, force-velocity and activation (1.2). We calculated SEE length as the difference in length of the CE from the MTU (1.3).

$$L_{MTU} = f(\theta_{ank}, r_{ank}) \quad (1.1)$$

$$(L_{CE}, \dot{L}_{CE}) = f(F-L, F-V, a) \quad (1.2)$$

$$L_{SEE} = L_{MTU} - L_{CE} \quad (1.3)$$

MTU force (F_{MTU}) was a function of the model's SEE stiffness and strain. In the reflex pathway, we normalized MTU force to F_{max} and applied a gain (1.2); a 10ms delay was added to emulate a positive force feedback neural reflex signal (Stim) and CE activation dynamics (ACT) were modeled to close the feedback loop (Fig.1D).

We calculated desired exoskeleton torque as a portion of the estimated biological moment.

$$\tau_{exo} = F_{MTU} \times r_{ankle} \times \psi \quad (1.4)$$

where ψ represented a percentage of the estimated biological torque. We calculated and recorded internal model states (F_{MTU} , L_{MTU} , L_{CE} , $STIM$, ACT) in dSpace for 5 second periods for offline analysis.

Testing Protocol: Two healthy young adults (M/77.4 kg, and F/75.2 kg) completed the IRB approved protocol. Each subject walked in two conditions (with exoskeleton-No Power and with exoskeleton-Power) and at three speeds for each condition (1.25, 1.50, and 1.75 m/s) for a total of six trials presented in random order. Metabolic power was estimated from indirect calorimetry data collected in minute 5 and 6 of each trial (OxyCon Mobile, Carefusion, USA).

RESULTS AND DISCUSSION

Mechanical performance of the neuromuscular model (NMM) based controller diminished as walking speed increased. Average exoskeleton torque decreased as speed increased from 1.25 to 1.50 to 1.75 m/s (15.0, 11.6, 8.9 Nm) and peak torque followed a similar trend (Fig. 2a). Net work performed per stride remained nearly constant across speed as the exoskeleton dissipated small amounts of energy in each case.

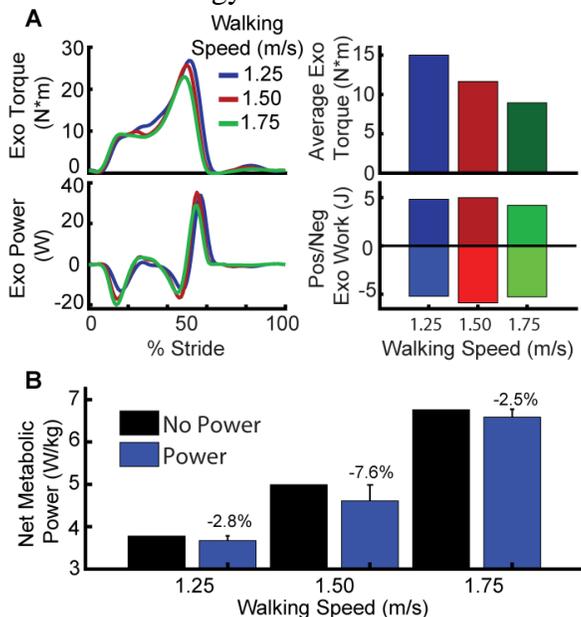


Figure 2: Trial results of two subjects walking at different speeds and conditions. **A.** Exoskeleton Torque and Mechanical Power over the stride along with Avg. Torque and Work for each speed. **B.** Net metabolic rate with % reduction displayed for PWR trial compared to NPWR trial at each speed.

Exoskeleton mechanical assistance resulted in reduced metabolic cost across all walking speeds. Compared to the exoskeleton-No Power condition, the reductions in net metabolic rate due to exoskeleton assistance were 2.8%, 7.6%, and 2.5% at 1.25, 1.50 and 1.75 m/s respectively (Fig. 2b).

CONCLUSIONS

Our initial results indicate that the adaptive properties of NMM-based control seen in prostheses may not generalize to lower-limb exoskeletons. This finding may not be that surprising given that this biologically-inspired control framework is designed to mimic the mechanical performance of biological plantarflexors. In fact, gastrocnemius and soleus muscles exhibit reduced force production at higher walking speeds [2] due to unfavorable force-velocity effects [2,3]. Previous studies on lower-limb prostheses using this control strategy may have avoided this constraint by operating at relatively low walking speeds (≤ 1.4 m/s) [4].

Perhaps an improved class of NMM-based controllers, incorporating a feedforward contribution to neural drive, would lead to improved adaptive performance. For example, adding the user's EMG or some other phase-based signal generator that responds positively to locomotion task demand, on top of reflexive positive force feedback could help counteract the aforementioned performance deficits due to force-velocity effects in the CE component of the NMM.

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