SPEED-DEPENDENT, PROPORTIONAL MYOELECTRIC EXOSKELETON CONTROLLER WITH ADAPTIVE GAINS

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INTRODUCTION

Chronic stroke patients typically have limited forward propulsion during walking due to weak and uncoordinated ankle function in their paretic limb[1]. A powered exoskeleton capable of applying mechanical assistance during the propulsive phase of gait is a promising approach to enhance mobility and quality of life post-stroke.

Previous work by Takahashi et al. shows encouraging, but limited improvements to steady state walking in stroke survivors by applying a powered ankle exoskeleton[2]. The powered ankle exoskeleton controller used by Takahashi et al, applied assistance proportional to the paretic soleus muscle activation (proportional myoelectic control) during the propulsive phase of gait (via anterior-posterior propulsive force gating). Although the exoskeleton increased the paretic limb plantar flexion moment, there was no reduction in metabolic cost. The researchers identified two major areas for improvement in the controller for future investigations. 1.) The propulsive assistance of the exoskeleton caused reduced paretic soleus activation, limiting the potential for the exoskeleton's assistance during push off 2.) The treadmill was constrained to a velocity at which subjects were already comfortable walking, and thus could have limited the effects of the exoskeleton.

Our goal was to create an ankle exoskeleton controller that could account for these deficits by 1.) ensuring the control signal is saturated despite a reduction in soleus activation and 2.) automatically modulating the magnitude of propulsive assistance (exoskeleton torque) based on walking speed.

METHODS

Controller: Our controller was based on that of Takahashi et al., however, we applied two additional gains (Fig. 1, grey boxes) before computing the final assistive plantar flexion torque applied by our powered ankle exoskeleton (exoskeleton torque). To compute these gains and the resulting control signal, we acquired soleus electromyographs, anterior-posterior ground reaction forces, and treadmill belt speed in real-time and processed them via a control board (dSPACE, Germany) to the controller on a desktop computer. The appropriate force, for a desired exoskeleton torque, was then applied to the exoskeleton from a benchtop motor (Baldor Electric Co) through a Bowden-cable transmission system. We fabricated custom carbon fiber ankle exoskeletons and measured applied ankle exoskeleton torque with a load cell (500 Hz, LCM Systems Ltd, UK) placed in series with the applied assistance.

The control diagram (Fig. 1) shows that similar to the Takahashi approach, the myoelectric component of the controller was based on raw soleus EMG (960 Hz, Biometrics, UK) processed to obtain an EMG envelope. The EMG envelope was gated based on the occurrence of the propulsive anterior-posterior GRF to apply assistance only during the propulsive phase of gait. An adaptive EMG gain (similar to Koller et al.) was calculated from a moving average of EMG peaks over five steps (Fig. 1, green ticks)[3]. The product of the

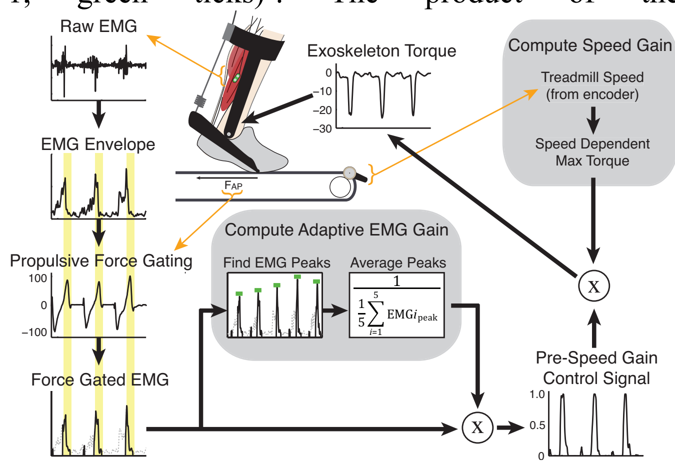

Figure 1: Diagram of exoskeleton controller. Grey boxes designate the gains applied in the controller. Yellow highlighting: propulsive phase of the stride cycle. Green ticks: peak EMG used to compute the adaptive EMG gain.

adaptive EMG gain and the force-gated EMG signal resulted in the pre-speed gain control signal. An encoder (1024cpr Encoder Products Company, USA) installed on a treadmill belt roller (Bertec, USA) supplied instantaneous treadmill velocity input to the controller. From the velocity input, we calculated a velocity dependent gain representing 25% of the max biological planar flexion moment at the given speed. This velocity dependent gain was then applied to the normalized pre-speed gain to calculate the exoskeleton torque profile.

Testing Protocol: Three able-bodied subjects (male = 2) completed the IRB approved protocol. Subjects walked during two randomized conditions: without the exoskeleton (CON) and with the powered exoskeleton on the right leg (EXO). Presentation of the conditions was randomized. During both conditions, subjects completed a continuous walking protocol. The treadmill velocity began at 1.0m/s and increased every minute by 0.1m/s until a walking speed of 1.9 m/s was achieved. Soleus muscle activation, ground reaction forces, pre-speed gain control signal, and exoskeleton load cell data were simultaneously collected for 20 seconds at each speed in VICON (960Hz, UK). Post processing was completed in Visual3D (CMotion, USA) and MATLAB (Mathworks, USA).

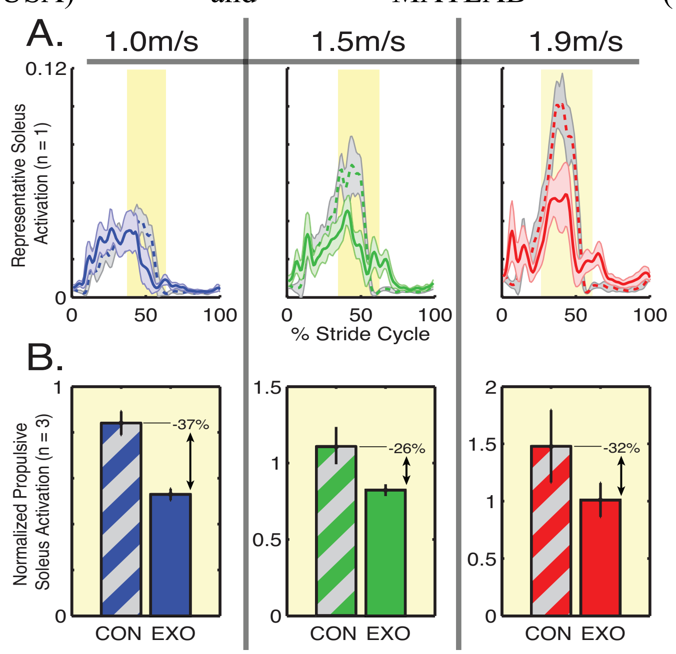

Figure 2: Soleus EMG (mean±std) **A.** Representative data comparing CON and EXO. Yellow highlight indicates propulsion. **B.** CON vs. EXO iEMG during propulsion.

RESULTS AND DISCUSSION

Consistent with previous work, right soleus EMG activity decreased when powered exoskeleton assistance was applied (Fig. 2A). On average there was a 37, 26, and 32% reduction in integrated soleus EMG (iEMG) during the propulsive phase of gait when compared to CON during walking at 1.0, 1.5, and 1.9m/s, respectively (Fig. 2B).

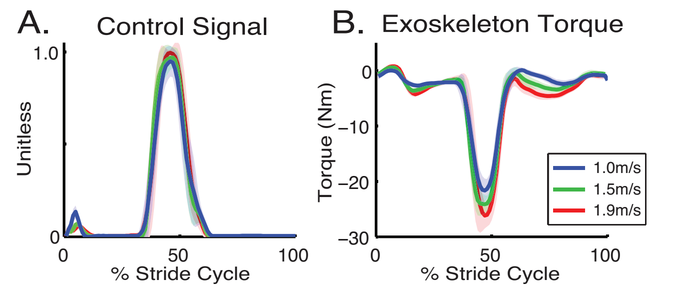

Figure 3: Representative subject data (mean±std) **A.** Pre-speed gain control signal and **B.** Exoskeleton torque.

Despite the reductions in soleus EMG, the pre-speed gain control signal maintained saturation at 1. (Fig. 3A). Thus, verifying the effectiveness of the adaptive EMG gain and preventing an undesired reduction in the exoskeleton's propulsive assistance. Furthermore, the automatic modulation of propulsive assistance was verified by the increase in exoskeleton torque corresponding to the increase in treadmill velocity (Fig. 3B). Thus, verifying the effectiveness of the speed gain.

CONCLUSION

Our novel exoskeleton control scheme can account for changes in both the user's EMG activity and walking speed, and mitigates the deficits previously identified by Takahashi et al. Future work is aimed at using this control strategy to improve walking mechanics and energetics in stroke survivors.

REFERENCES

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[2] Takahashi, K.Z et al., J NeuroEng Rehab. 2015;12:23.
[3] Koller, JR et al., J NeuroEng Rehab. 2015; 12:97

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