Physiological Effects of a Simple Adaptive Ankle Exoskeleton

Jeffrey R. Koller, Daniel A. Jacobs, Daniel P. Ferris, C. David Remy
The University of Michigan, Ann Arbor, MI, USA
jrkoller@umich.edu, jacobsda@umich.edu, ferrisdp@umich.edu, cdremy@umich.edu

1 Introduction

A major problem in exoskeleton design is the unnatural movement that comes from a poor human machine interface. Innovative controller designs are a primary focus in making the interface between man and machine more fluid and intuitive. One controls approach is to detect the user's intent to move and then command the exoskeleton's actuators to act appropriately. These types of controllers rely on intrinsic mechanical measurements such as angles, forces/torques, or accelerations to sense the user's intent. Once the controller has detected intent, it then commands actuators to move in parallel with the user to assist them through predefined motions [1]. These control schemes commonly lag behind the user because the intrinsic mechanical measurements are only sensed once the user has already begun movement [2]. Additionally, many of these controllers are constraining the user to move in a particular predefined manner, one that may not be to the user's preference.

An alternative controls approach is to use the neural signals from the wearer directly as control inputs to the exoskeleton. We can record the neural signals sent to the wearer's muscles using electrodes placed on the skin's surface. Our research group has used electromyography (EMG) from the wearer to directly control actuation timing and amplitude of multiple walking exoskeletons. These controllers create a control signal by multiplying the EMG linear envelope by a *constant* mapping gain. This proportional myoelectric control scheme has resulted in significant reductions in the wearer's metabolic cost when implemented on a simple ankle exoskeleton using the wearer's soleus muscle as the control input [3]. However, the constant mapping gain of this control scheme could be a limiting factor of the controller.

With past ankle exoskeletons our group has hand tuned a calibration mapping gain so that the control signals barely saturated during unpowered walking. This calibration mapping gain was then doubled for powered walking to encourage a reduction in the wearer's own muscle recruitment [4]. A couple assumptions are taking place when this constant mapping gain is imposed. First is that the researcher is assuming that all subjects can reduce their peak soleus electromyography by half. By doubling the calibration mapping gain, the wearer must halve their peak muscle recruitment to achieve the full control range of the exoskeleton. The second is that 50% peak muscle recruitment may not be the subjects preferred way to walk with the exoskeleton. The doubled calibration mapping gain could be considered a constraint on the user. This constraint conforms subjects to walk in a manner that has been predefined by the researcher as the "best" way to walk with the device.

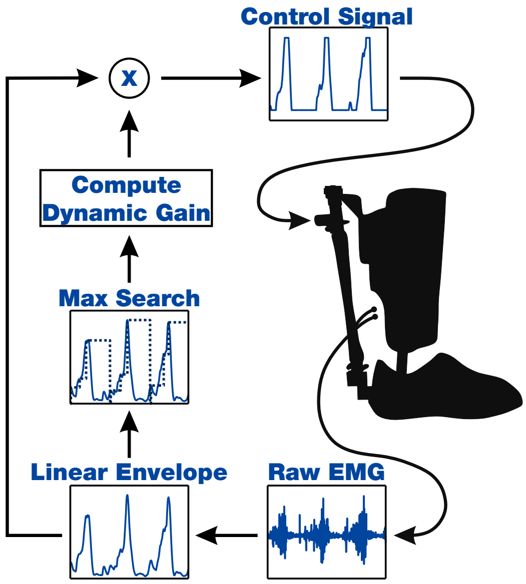

Figure 1: Our proportional myoelectric control algorithm dynamically adjusts the mapping gain as the user adapts to the device. The controller performs a max search on each stride which is added to a moving average of the previous strides. The dynamic gain is calculated using this average and a user defined maximum mapping voltage. The calculated dynamic gain is then multiplied by the linear envelope prior to being sent to the actuators.

In this study we have created an adaptive proportional myoelectric controller in order to give the wearer full control range regardless of their muscle recruitment. We hypothesized that the adaptable controller will reduce the metabolic cost of walking.

2 Approach

We designed and built a simple one degree of freedom ankle exoskeleton (2.09 kg) for this study. This exoskeleton assisted with plantar flexion and was powered by artificial pneumatic muscles. The exoskeleton was controlled by an off board real-time processor.

2.1 Controller Design

We used the processed soleus EMG to proportionally command control inputs to the exoskeleton actuators. Our controller first processed the raw EMG to get the signal's lin-

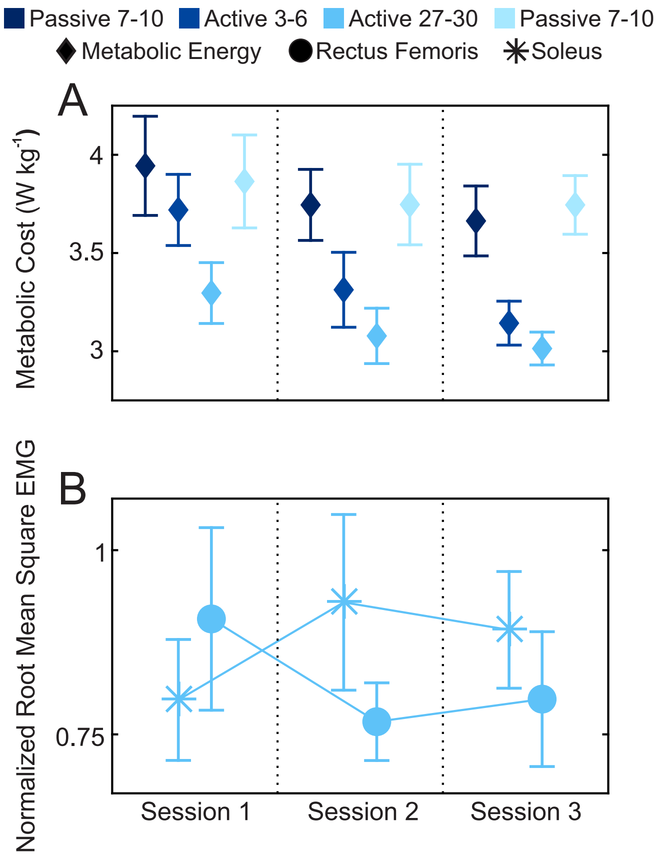

Figure 2: (A) Subjects walked in the exoskeleton continuously for a total of 50 minutes; 10 minutes unpowered (passive), 30 minutes powered (active), and then 10 minutes unpowered. Each measurement was an average of metabolics over a three minute time span. A reduction occurs across training days when comparing unpowered to powered conditions. The final session resulted in an average metabolic cost of 3.01 $\pm$ 0.08 W kg^{-1} (s.e.m.) during the last three minutes of the active walking. (B) When considering the EMG from the end of the active condition across training days we see an increase in soleus activity and a decrease in rectus femoris activity.

ear envelope. Then the controller conducted a real-time max search of the linear envelope on a stride by stride basis. Once the maximum of a stride was found it was added to a moving average of the previous fifty strides. Our controller then calculated the mapping gain necessary for this moving average maximum solues activity to map to a user defined desired maximum actuator voltage. This desired maximum actuator voltage remained constant across all subjects and testing days.

2.2 Experimental Design

We tested eight subjects (male, 21 $\pm$ 1 years, 74.0 $\pm$ 2.7 kg, 180.0 $\pm$ 2.8 cm; means $\pm$ s.e.m.) during treadmill walking at 1.2 m s^{-1}. All subjects walked in the device continuously for 50 minutes on three separate training days. Each training day was identical and consisted of 10 minutes of unpowered (passive) walking, 30 minutes of powered (active) walking, followed by 10 minutes of unpowered walking. We collected electromyography, metabolic, kinematic, and kinetic data across all sessions.

3 Results

We found that by the end of the third session, our adaptive controller was choosing gains that were a 1.50 $\pm$ 0.14 (mean $\pm$ s.e.m.) scaling of the would be calibration mapping gain. Previous studies used a constant scaling factor of 2.0.

We found a significant reduction in metabolic cost on all training days ($P<0.05$, repeated ANOVA analysis) when comparing passive to active walking with the exoskeleton. There was an average metabolic reduction of 0.65 $\pm$ 0.13 W kg^{-1} or 16.9 $\pm$ 2.8% (mean $\pm$ s.e.m.) by the third training day. Our metabolic results are similar to those found in previous studies where a constant mapping gain was used [4].

Interestingly, the root mean square (r.m.s.) of the soleus EMG increased across the end of each training session. This contradicts the findings of Sawicki et al. where a constant mapping gain was used [4]. We believe this discrepancy comes from the fact that subjects are not constrained to walk in a specific way when using our adaptive controller.

This increase in the r.m.s. of soleus EMG was coupled with a decrease in the r.m.s. of rectus femoris EMG. This suggests subjects found it energetically economical to increase effort at the ankle in exchange for decreased effort at the hip. Findings from Lewis et al. show that there is a trade off between ankle and hip dynamics during walking [5]. Our results suggest this trade off may come with energetic benefits.

Beyond demonstrating the benefits of an adaptive EMG controlled exoskeleton, our findings open an interesting discussion as to what is the best way of providing assistance to the human gait. Our data suggests that subjects voluntarily increased effort at the ankle in order to reduce effort at the hip. Similar strategies (as opposed to replicating ‘normal joint action) might prove beneficial for intrinsically controlled exoskeletons or active prostheses.

References

[1] K. Anam and A. A. Al-Jumaily, “Active exoskeleton control systems: State of the art,” *Procedia Engineering*, vol. 41, pp. 988–994, 2012.

[2] D. P. Ferris, G. S. Sawicki, and M. A. Daley, “A physiologist’s perspective on robotic exoskeletons for human locomotion,” *International Journal of Humanoid Robotics*, vol. 4, no. 03, pp. 507–528, 2007.

[3] D. P. Ferris, K. E. Gordon, G. S. Sawicki, and A. Peethambaran, “An improved powered ankle–foot orthosis using proportional myoelectric control,” *Gait & posture*, vol. 23, no. 4, pp. 425–428, 2006.

[4] G. S. Sawicki and D. P. Ferris, “Mechanics and energetics of level walking with powered ankle exoskeletons,” *Journal of Experimental Biology*, vol. 211, no. 9, pp. 1402–1413, 2008.

[5] C. L. Lewis and D. P. Ferris, “Walking with increased ankle pushoff decreases hip muscle moments,” *Journal of biomechanics*, vol. 41, no. 10, pp. 2082–2089, 2008.