# Heuristic-based online adaptation of ankle exoskeleton assistance using plantarflexor electromyography

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## INTRODUCTION

People often change their coordination strategies as they learn to walk with ankle exoskeletons [1], yet most current exoskeleton control approaches do not appropriately account for these changes. Time-based assistance techniques, in which the exoskeleton is actuated at a specific point in the gait cycle [2, 3], keep device behavior static regardless of human adaptation. Proportional myoelectric control, in which torque provided from the exoskeleton is directly proportional to the user's muscle activity [4], requires a certain level of muscle activity be maintained for the exoskeleton to supply torque, preventing muscle activity from being fully supplanted. Control techniques that adjust exoskeleton behavior online in response to measured changes in human coordination patterns would allow the human and device to co-adapt and potentially result in improved human-robot interaction.

The goal of this project is to use soleus muscle activity, measured online, to update the desired ankle exoskeleton torque trajectory every step. Ideally, the exoskeleton will learn the changing pattern of soleus muscle activity over time and supplant the role the soleus muscle plays in plantarflexion. We expect to be able to tie our immediate goal, to drive soleus muscle activity down towards zero, to a higher level outcome, such as reduced whole-body metabolic rate.

## METHODS

We developed an electromyography-based adaptive controller that optimizes the desired exoskeleton torque pattern in real time. The foundation of the control scheme is given by the following equation:

$$\tau_{\mathrm{des}}(i, n+1) = k_1 \cdot EMG_{\mathrm{SOL}}(i+d, n) \qquad (1)$$
$$- k_2 \cdot EMG_{\mathrm{TA}}(i+d, n) + \tau_{\mathrm{des}}(i, n)$$
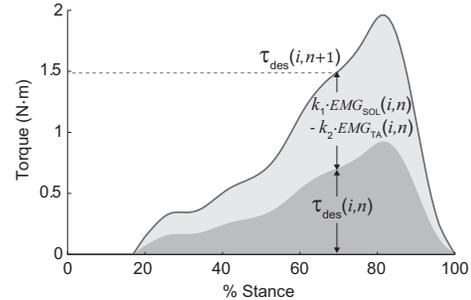


Fig. 1: Schematic depicting the desired exoskeleton torque as defined by the proposed control scheme. Desired exoskeleton torque on the next step is the sum of the desired torque, soleus muscle activty, and tibialis anterior muscle activity on the current step.

where $\tau_{\mathrm{des}}$ is the desired exoskeleton plantarflexion torque, $i$ is the time index of the current step, $n$ is the step number, $d$ is a delay term, $k_1$ is a positive gain, $EMG_{\mathrm{SOL}}$ is soleus (plantarflexor) muscle activity, $k_2$ is a positive gain, and $EMG_{\mathrm{TA}}$ is tibialis anterior (dorsiflexor) muscle activity. The sampling rate of our system is 500 Hz, thus the time index $i$ is updated every 2 ms and reset every heel strike.

The desired torque curve on the next walking step is a function of soleus muscle activity, tibialis anterior muscle activity, and desired torque on the current walking step (Fig. 1). Soleus muscle activity ($EMG_{\mathrm{SOL}}$) acts to increase desired torque. As torque increases and soleus muscle activity decreases, the soleus muscle's contribution to desired torque will decrease, thereby slowing the growth of the desired torque curve over time. Tibialis anterior muscle activity ($EMG_{\mathrm{TA}}$) is meant to help stabilize the controller if soleus muscle activity does not decrease as expected. In theory, tibialis anterior muscle activity would increase if the exoskeleton applies an undesirable amount of plantarflexion torque in an attempt

to resist excessive plantarflexion or to ensure certain ankle kinematics. Finally, the contribution of the desired torque on the current walking step, $\tau_{\text{des}}(i,n)$, to the desired torque on the subsequent walking step, $\tau_{\text{des}}(i,n+1)$, introduces history dependence. This is desirable because a reduction in soleus muscle activity does not necessarily result in a reduction in exoskeleton torque, thereby enabling the exoskeleton to fully supplant the soleus muscle's contribution to plantarflexion. The full formulation of the control scheme includes several additional terms necessary to stabilize the controller and prevent torque from growing unbounded.

## RESULTS

We measured soleus muscle activity and metabolic rate in N = 10 able-bodied participants as they walked for 30 minutes with the heuristic-based adaptive controller applied to bilateral ankle exoskeletons. The control approach significantly reduced soleus muscle activity below that observed during walking with the exoskeletons while they applied zero torque and during walking in normal shoes. Furthermore, metabolic rate during walking with the adaptive controller was 22% below that observed during walking with the exoskeletons while they applied zero torque (p = $1 \cdot 10^{-4}$).
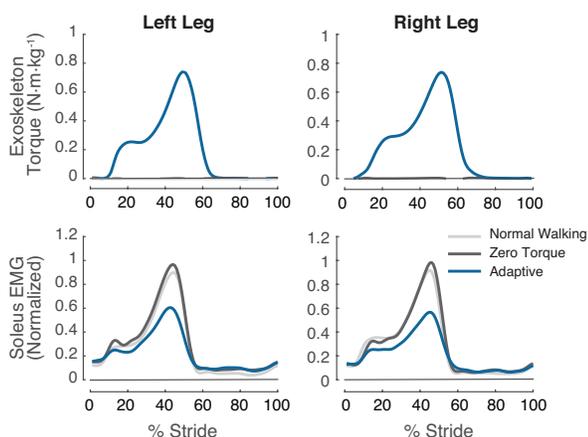


Fig. 2: Results for N = 10 participants show that our heuristic-based adaptive control scheme can generate exoskeleton torque profiles (*top row*) that significantly decrease soleus muscle activity (*bottom row*) below that observed during walking with zero torque from the exoskeletons and during walking with normal shoes.

## DISCUSSION

This control approach proved effective at driving soleus muscle activity, and consequently metabolic rate, down in participants. Due to the fact that exoskeleton torque was allowed to evolve independently for each participant and for each leg, there was large variation in the evolved exoskeleton torque profiles across individuals and across the left and right legs. Desired exoskeleton torque stabilized relatively quickly for most participants while metabolic rate took longer to converge. For some subjects, metabolic rate had not reached steady-state after 30 minutes of walking, suggesting that more exposure might be necessary to obtain the greatest benefit.

## CONCLUSIONS

Using this approach, we hope to learn more about human motor adaptation and how to best interact with the human musculoskeletal system using exoskeletons. Through the analysis of optimized exoskeleton torque trajectories from multiple different subjects, we may be able to extract common characteristics from which we can derive general principles about human locomotion and effective human-robot coordination strategies. We expect this control strategy to be naturally transferable to multiple joints and allow for full lower-limb assistance during human walking.

## FUNDING

## REFERENCES

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