Hip Exoskeleton Emulator to Explore Spring-Like Assistance Strategies During Walking

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Summary

Knowledge of effective control strategies for reducing the metabolic cost of walking with active exoskeletons is limited. Here, we used a hip exoskeleton emulator to assess how stiffness and reference angle of a simplified impedance controller influence metabolic cost. In general, impedance parameter settings biased toward hip exoskeleton extension torque were more effective at reducing metabolic cost during walking.

Introduction

Unpowered elastic exoskeletons have reduced the metabolic cost of walking when applied at the ankle [2] and running, when applied at the hip [3]. To date, no study has examined whether spring-like assistance at the hip can reduce the metabolic cost of walking [1]. This is surprising given that the hip joint is a major source of torque during walking and exhibits spring-like behavior during stance phase. Simplified impedance controllers can also exhibit spring-like behavior whereby, assistance torque is a function of joint angle (Eq.1). Here, we used a tethered hip exoskeleton with a simplified impedance controller (Eq.1) to examine the influence of the stiffness and reference angle of a virtual spring on a user's metabolic cost during walking with mechanical assistance. We hypothesized that spring-like control parameters that generate torque assistance profiles with magnitude and timing similar to physiological hip torques would result in the lowest metabolic cost.

Methods

We recruited one male volunteer who gave informed and written consent to participate, per the approved IRB protocol. We used a cable-driven bilateral hip exoskeleton emulator (HuMoTech, LLC.) to apply torques in real-time according to,

$$\tau_{exo} = k(\theta_{hip} - \theta_0) \qquad \text{Eq. 1}$$

where k is a stiffness and θ_0 is the reference angle of a virtual spring acting at the hip. We measured θ_{hip} using encoders mounted on the exoskeleton frame (+ is extension). We set exoskeleton torque to zero during late swing. The volunteer performed one 6-min trial for each combination of k and θ_0 (3x3) at 1.0 m/s. We chose θ_0's with flexion bias, extension bias, and neutral (*i.e.,* no bias – the hip joint angle where physiological torques shifted from extension to flexion during stance). We chose k's to generate torques of up to ~25% of the physiological peak during walking (k = 0.2, 0.4, & 0.6 Nm/deg). We used indirect calorimetry and a standard physiological equation to measure net metabolic power (W/kg).

Results and Discussion

Exoskeleton torques responded differently to changes in k vs θ_0. For a θ_0 = -12º, increasing k from 0.2 to 0.6 Nm/deg increased the range of applied torque from 0.07 to 0.29 Nm/kg. For a fixed k (0.6 Nm/deg), θ_0 changed both timing and magnitude of extension and flexion torques. Timing of the transition of extension to flexion torque shifted from 35% to 50% of the gait cycle in flexion vs extension bias conditions. This resulted in decreased peak flexion torques (-0.18 to -0.07 Nm/kg) and increased peak extension torques (0.02 to 0.21 Nm/kg).

Parameter combinations that provided increased extension assistance (both timing and magnitude) corresponded with lower metabolic cost. The highest metabolic cost (3.80 W/kg) was measured with the flexion biased reference angle and lowest stiffness, while the lowest metabolic cost (3.29 W/kg) was measured with the extension biased reference angle and highest stiffness.

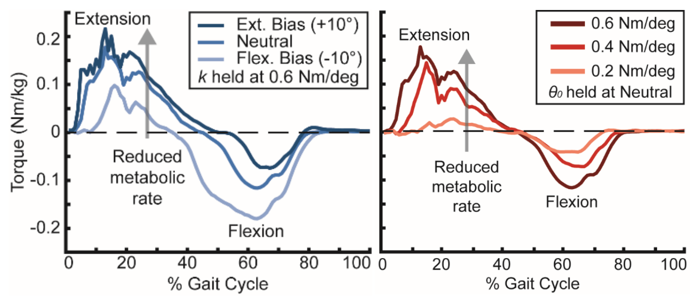

Figure 1: Hip exoskeleton torques across θ_0's (Left) and k's (Right)

In agreement with our hypothesis and other studies using powered hip exoskeletons [3], our initial result that exoskeleton controllers which tend toward physiological torque profiles during early stance hip extension are most effective at reducing metabolic cost.

Conclusions

Our preliminary results suggest spring-like properties of hip exoskeletons can be tuned to reduce the metabolic cost during walking. It remains unclear whether continuing to amplify hip exoskeleton extension assistance beyond levels studied here will lead to further decreases in metabolic cost (*i.e.,* is more better?). We will explore a broader parameter space and employ human-in-the-loop optimization procedures to find the optimal control parameters. Long-term, we hope to translate optimal control concepts from emulator studies to state-of-the-art pseudo-passive hip exoskeleton technology. Ideally, these devices will adapt to both user and environment under real-world locomotion scenarios by modifying impedance parameters in real-time.

References

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