EFFECTS OF A PASSIVE ELASTIC EXOSKELETON DURING WALKING

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INTRODUCTION

Locomotor function is often limited following a neurological injury, although the underlying mechanisms are not always clear. While walking appears to be a complex task from a neural control perspective, many aspects of gait may be predicted simply from the passive mechanics of the body, the focus of a field commonly termed dynamic walking.

Following a stroke or spinal cord injury, gait speed is primarily limited by maximal step frequency [1, 2]. Recent dynamic walking model simulations predicted that step frequency is primarily controlled by actuation at the hip and knee joints [3]. This actuation does not require complex active control, but may be provided by passive elastic elements. Such an actuation method can produce human-like gait speeds and leg swing kinematic patterns without the need for energetic input at the hip or knee. These simulation results suggest that neurological patients may have limited maximal step frequencies due to an inability to appropriately activate their proximal leg musculature.

Based on our previous model simulations, we developed a passive exoskeleton which is able to store and return mechanical energy using springs which parallel the users' legs. The springs are situated such that a single spring produces torque at the hip and knee, a configuration our simulations suggest is necessary to produce typical leg swing kinematics. These two-joint springs are similar in concept to bi-articular muscles of the upper leg, specifically the rectus femoris and biceps femoris.

We hypothesized that storage and return of mechanical energy by the exoskeleton's elastic elements would reduce the need for subjects to actively produce joint torques. We expected that the mechanical assistance provided by the springs would allow subjects to reduce the swing phase activity of muscles acting across the hip and knee.

METHODS

The exoskeleton was lightweight (3.5 kg), consisting of a carbon fiber waist belt attached to adjustable length aluminum segments lateral to the thigh and shank (Fig. 1). Thrust bearing hinges connected the rigid segments, and were aligned with the subjects' hip and knee joints. Padded plastic cuffs around the shanks and thighs mechanically linked the user to the device. Extension springs (stiffness = 2000 N/m) were placed in series with wires running along grooved aluminum sheaves at the hip (moment arm = 7.6 cm) and knee (moment arm = 2.5 cm). The ratio between the hip and knee moment arms was based on earlier simulations [3].

Four young (27±3 yrs), healthy subjects walked on a treadmill at 1.25 m/s under three conditions: *No Exoskeleton*; Exoskeleton with *No Springs*; Exoskeleton With *Springs*. Trial order was randomized. For each trial, subjects were given 3 minutes to reach a steady state, and data was recorded for the next 30 seconds.

We collected bilateral EMG data from the rectus femoris (RF) and biceps femoris (BF). Electrodes were placed over the muscle bellies, avoiding contact with the exoskeleton. Pressure footswitches were used to quantify stride timing. For analysis, processed EMG data for each muscle were divided

Figure 1. We built a passive exoskeleton with two-joint springs crossing the hip and knee. The design of this device was based on previous model simulations, with the goal of providing swing phase gait assistance.

into strides, averaged, and normalized by the peak value across all trials. For one subject, we quantified spring joint torques using load cells placed in series with the springs.

RESULTS AND DISCUSSION

The net joint torques produced by the springs varied as expected. At the beginning of the swing phase, the springs produced hip flexion and knee extension torques. Later in swing the effects of the springs reversed, producing hip extension and knee flexion torques. This swing phase torque profile is similar to the pattern of active torque production typically seen in human walking [4].

As expected, the hip and knee joint torques produced by the springs late in the swing phase allowed subjects to reduce BF muscle activity (Fig. 2). Averaged across subjects, this effect was significant (p=0.010). Donning the exoskeleton slightly increased muscle activity, likely due to the mass of the device. But adding springs decreased BF activity below the level seen during normal walking. This implies that subjects were able to adapt their muscle activation pattern to take advantage of the exoskeleton assistance.

The effects of the exoskeleton on RF activity were less clear (Fig. 3). RF activity was quite low during the Pre-Swing phase, when we would expect the exoskeleton assistance to be most beneficial. While the springs appeared to slightly decrease RF activity, this change was not significant (p=0.52). Unexpectedly, the addition of springs significantly (p=0.049) decreased RF activity during early stance, when RF activity was maximal. It is possible that the springs acted to stabilize the knee at heel strike, allowing subjects to decrease co-contraction.

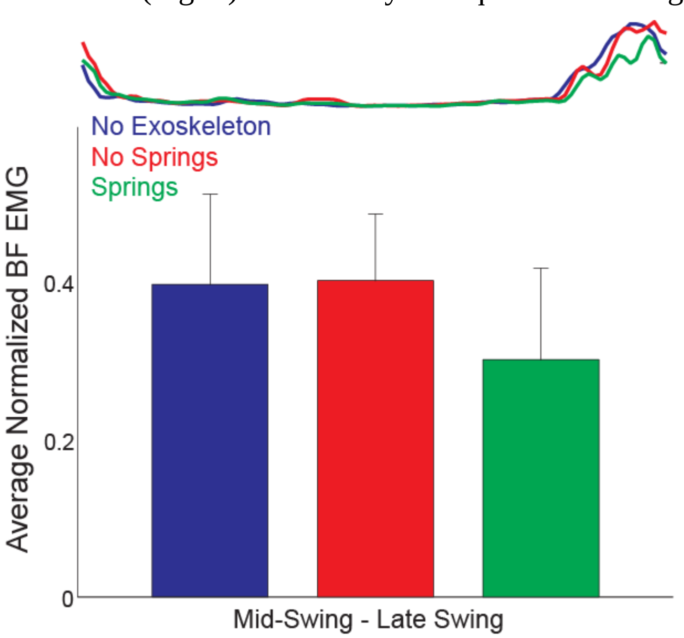

Figure 2. Changes in BF activity are illustrated for a single subject (top) and averaged group data (bottom). Single subject data is plotted from heelstrike to heelstrike, while group data is averaged over Mid and Late Swing (73-100% gait cycle).

CONCLUSIONS

A simple passive exoskeleton worn while walking allows users to decrease muscle activity during specific gait phases. Such a device has the potential to provide useful gait assistance during rehabilitation, but future work must investigate effects outside of the sagittal plane.

REFERENCES

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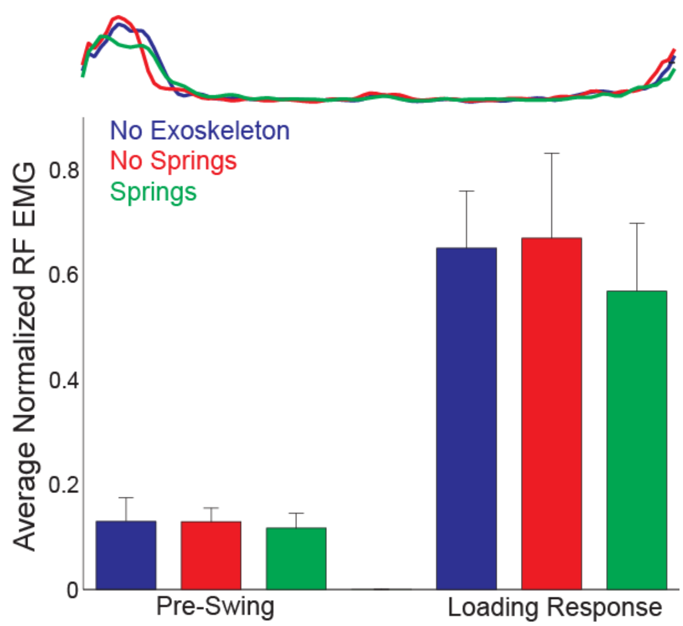

Figure 3. Changes in RF activity for a single subject (top) and averaged group data (bottom). Group data is averaged separately over Pre-Swing (50-58% gait cycle) and Loading Response (1-10% gait cycle).