AN ANKLE-FOOT ORTHOSIS POWERED BY ARTIFICIAL MUSCLES

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INTRODUCTION

Increasing evidence suggests that locomotor training can greatly improve human walking ability after neurological injury (Barbeau et al. 1998). After stroke or spinal cord injury, treadmill stepping with manual assistance and partial body weight support facilitates re-learning how to walk. However, manual assistance is labor intensive and highly variable from therapist to therapist. A lightweight powered lower limb orthosis that can assist locomotor training would decrease labor requirements and provide more consistent therapy. In addition, a powered orthosis would also provide the means to design controlled motor learning studies on human locomotion. Thus, we designed a pneumatically powered, myoelectrically controlled ankle-foot orthosis as a tool for rehabilitation and studying human locomotor adaptation.

METHODS

Figure 1: The orthosis has a carbon-fiber shell and a hinge joint at the ankle. Two identical artificial pneumatic muscles (McKibben muscles) are attached. The dorsiflexor is contracted and the plantarflexor is relaxed in the figure at right.

The artificial pneumatic muscles are lightweight and can produce high power outputs. They are made from latex tubing surrounded by a braided polyester shell. Inflating the tubing causes the shell to expand radially and shorten axially. A desktop PC used programs written in SIMULINK and implemented in dSPACE to regulate the pressure in the artificial pneumatic muscles proportional to the amplitude of low-pass filtered EMG. Force transducers mounted in series measured tension of the artificial muscles. Orthosis weight is 1.4 kg. We performed benchtop isometric tests on the muscles to compare artificial muscle mechanical properties with published data from human muscle. We also recorded kinematic and kinetic data on one healthy subject walking on a treadmill for thirty minutes with the orthosis.

RESULTS AND DISCUSSION

Mechanical properties of the artificial muscles were similar to the properties of human muscle. Single twitch tests with 5 ms pulse stimuli revealed a time to peak tension of 69 ms and a half relaxation time of 69 ms for the artificial muscles. Respective values for human triceps surae are 101 ms and 94 ms (Rice et al. 1988) and for human tibialis anterior are 99 ms and 87 ms (Connelly et al. 1999). During physiological activation, the artificial muscles had an electromechanical delay of 51 ms between EMG burst onset and initial rise in muscle tension. Electromechanical delay values for human soleus and gastrocnemius are 27 ms and 35 ms (Komi et al. 1987), respectively.

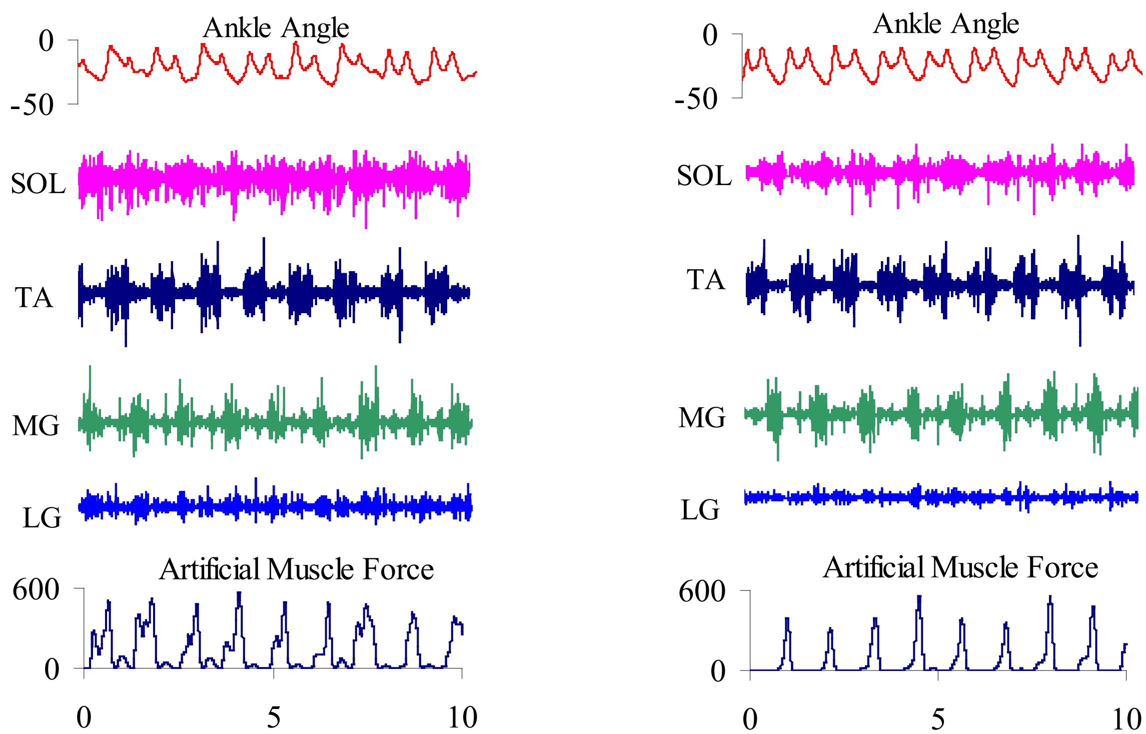

Figure 2: The left graphs show data during the first minute of walking on the treadmill (1.2 m/s) with the orthosis activated. The right graphs show data after thirty minutes of continuous walking. Only the plantarflexor artificial muscle is attached (moment arm = 11 cm). Soleus EMG amplitude decreased by 25% and lateral gastrocnemius EMG amplitude decreased by 41%. Tibialis anterior and medial gastrocnemius EMG amplitudes were unchanged. Mean artificial muscle force decreased by 48%. Notice the decrease in interstride variability in ankle kinematics and artificial muscle force with practice. Ankle angle is in degrees, EMG units are arbitrary but consistent across time, force is in Newtons, and the x-axis is Time in seconds.

SUMMARY

We have built a pneumatically powered ankle-foot orthosis that can be used to assist gait rehabilitation and study human locomotor adaptation. Preliminary data using proportional myoelectric control on a non-disabled subject indicate that the human nervous system selectively modifies muscle activation patterns to control the orthosis with practice. Kinematic or kinetic data (e.g. foot contact or hip angle) could also be used to control the orthosis when sufficient EMG signals are not present due to stroke or spinal cord injury. Future work will extend the concept to a hip-knee-ankle foot orthosis to provide assistance at other joints.

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