MOTOR ADAPTATION DURING DORSIFLEXION-ASSISTED WALKING WITH A POWERED ORTHOSIS

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INTRODUCTION

The ankle dorsiflexors have two main mechanical functions during gait. One function is eccentric control at heel strike and a second function is concentric control during swing. Correspondingly, in healthy persons, the ankle dorsiflexors show two bursts of electromyographic (EMG) activity during walking. A large burst is seen around heel strike to prevent the forefoot from colliding with the floor at a high velocity (i.e. foot slap). A second burst is seen after toe off to prevent the toes from hitting the floor on swing through (i.e. drop foot).

Post-stroke patients often have very weak dorsiflexor recruitment at the stance to swing transition. Rigid AFOs are frequently prescribed to improve walking ability and to prevent stumbling. However, rigid AFOs impede push-off plantar flexion and do not allow the user to make step-to-step changes in motion dynamics for terrain. The purpose of this study was to determine how people adapt their walking patterns to a powered orthosis with dorsiflexion assist. The advantage of a powered orthosis is that it would not impede plantar flexion at the end of stance.

Our preliminary results using a powered orthosis under proportional myoelectric control of the tibialis anterior (TA) revealed different motor adaptation responses for the two mechanical functions of the dorsiflexors (Kao and Ferris 2006). Although TA EMG amplitude decreased for the first burst by about 22% as subjects adapted, TA EMG amplitude did not change for the second burst. In order to clarify the different responses, we tested a second group of subjects that received the active dorsiflexion assist **ONLY** during the swing phase. We hypothesized that subjects would demonstrate similar adaptation as before for the second TA burst (i.e. no change in dorsiflexor EMG amplitude).

METHODS

Ten healthy subjects were fitted with custom-made polypropylene orthosis (Fig. 1). An artificial pneumatic muscle attached on the foot and shank portions powered the orthosis to provide dorsiflexor torque. A real-time computer controller modulated force in the artificial muscle proportional to tibialis anterior EMG amplitude via a pressure regulator. Foot-switches were used for gating control signals.

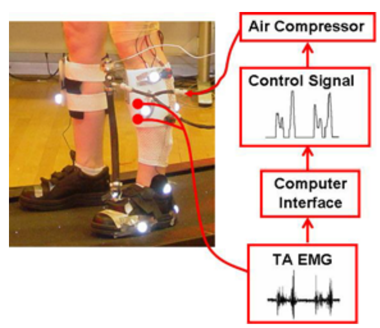

Figure 1: The powered orthosis uses TA EMG to control air pressure in an artificial dorsiflexor muscle.

We collected lower body kinematics, EMG, and artificial muscle force while subjects walked on a treadmill for two training sessions. Subjects walked with the orthosis first without power for 10 minutes (baseline), with power for 30 minutes (active), and without power again for 15 minutes (post-passive). We repeated the same protocol 3 days later. During active orthosis walking, subjects in the first group (n=5) received powered assistance **BOTH** at heel strike and during swing. Subjects in the second group (n=4) received assistance **ONLY** during the stance-to-swing transition (50-90% of step cycle). We calculated root-mean-square EMG for each burst of TA EMG individually and normalized data to the last minute of baseline. We also correlated every minute's ankle angle profile to the baseline trial. We used the method described by Noble and Prentice (2006) to define steady state dynamics

RESULTS AND DISCUSSION

At the end of Day 2, the orthosis provided peak dorsiflexor torques ~0.22 Nm/kg at heel contact for Group One subjects and ~0.10 Nm/kg during swing for both Group One and Two subjects. The assist provided was ~150% of the normal peak dorsiflexor moments (~0.1 Nm/kg at heel contact and ~0.06 Nm/kg during swing (Mills, 2001)).

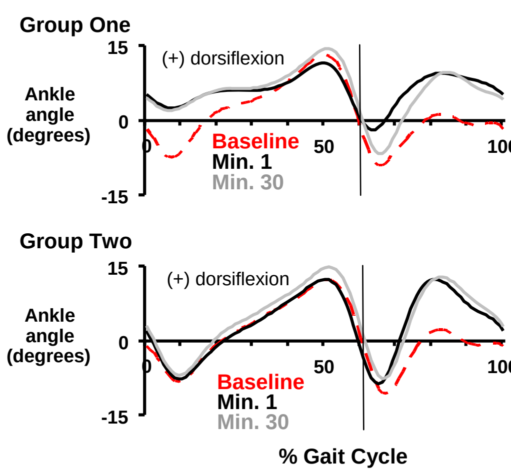

Figure 2: Ankle kinematics on Day 2.

Subjects walked with increased ankle dorsiflexion (Fig. 2) and had tibialis anterior muscle activation patterns similar to passive orthosis trials (Fig. 3). In comparison with the baseline condition, amplitude of the second TA EMG burst did not change during the active condition for Group Two subjects (baseline 1.17 ±0.10 (m±s.d.); active 1.6±0.10; $F_{1,40}$=0.0534, p=0.82). Both Group One and Two subjects reached steady state more rapidly on the second day.

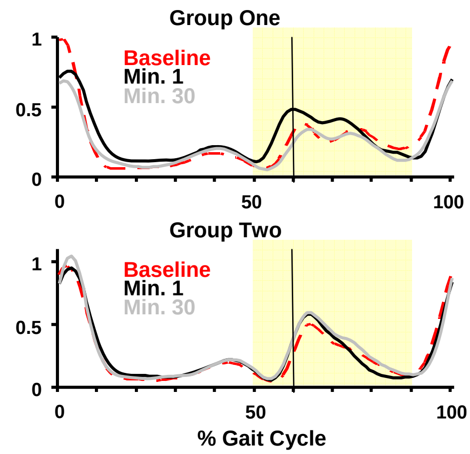

Figure 3: TA EMG on Day 2.

SUMMARY/CONCLUSIONS

When wearing a powered orthosis providing dorsiflexor assist, healthy subjects walked with increased dorsiflexion without altering TA EMG amplitude during the stance to swing transition. In contrast, subjects decreased TA EMG amplitude during the swing to stance transition when dorsiflexor assist was provided at that time. These findings indicate that the nervous system modulates the two TA EMG bursts differently in regards to motor adaptation. In addition, active dorsiflexion assist controlled by EMG has potential for assisting people with drop-foot gait.

REFERENCES

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