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Powered Lower Limb Orthoses: Applications in Motor Adaptation and Rehabilitation

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Abstract—**Task-specific practice can be beneficial for motor rehabilitation after neurological injury. Unfortunately, high labor demands have limited its clinical acceptance, especially for gait rehabilitation. A number of research teams around the world are testing large robotic devices for assisting treadmill stepping as a means for reducing therapist labor. We propose that powered lower limb orthoses may also have a role in assisting gait rehabilitation. Powered orthoses could assist task specific practice of gait with the long-term goal of improving patients' inherent locomotor capabilities. We present data showing that: (1) pneumatically powered lower limb orthoses can provide substantial mechanical assistance to human walking, (2) powered orthoses can lead to motor adaptation of gait in healthy subjects, and (3) powered lower limb orthoses may have positive benefits during gait rehabilitation.**

I. INTRODUCTION

Locomotor training can improve human walking ability following neurological injury [1-5]. Typically locomotor training involves patients practicing stepping with bodyweight support and external assistance as needed [6]. This therapy was developed based on two major principles learned from extensive studies on cats [7-11] and rats [12]. The first principle, task specificity (as applied to locomotor training), states that to improve walking ability patients must practice walking [11]. The second principle, activity-dependent plasticity, states that patients must be active participants in the therapy to drive neural adaptation [13, 14]. The functional benefits of locomotor training with manual assistance are considerable but so are the costs. Providing proper manual assistance is physically demanding and requires a high level of skill and training. Because it is labor intensive, a session of locomotor training with manual assistance can require several therapists. In addition the skill of the therapist is a very important factor in determining the efficacy of the therapy.

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Because of the drawbacks to manual locomotor training, scientists and engineers are developing robotic devices that can assist gait rehabilitation. Most of the currently available devices are designed to guide the legs through pre-programmed physiological gait patterns. The Lokomat® System developed by Hocoma (Switzerland) consists of a position controlled robotic gait orthosis that attaches to a treadmill frame and a body weight support system [15-18]. The AutoAmbulator® [www.autoambulator.com] is a similar device being developed by HealthSouth, a commercial healthcare provider. The Mechanized Gait Trainer is based on a crank and rocker gear system, providing limb motion similar to that of an elliptical trainer [19, 20]. Reinkensmeyer et al. are also working on devices that use pneumatic actuators and high bandwidth force control [21, 22]. All of these robotic devices clearly have potential for assisting gait rehabilitation after neurological injury, especially for patients with little to no walking ability. However, for patients with some but limited walking ability, it may also be helpful to consider other complementary devices.

An alternative approach for robotic gait rehabilitation devices is to make them wearable so that they can function during overground locomotion. This would allow the practice of task specific aspects of walking such as gait initiation and termination, turning, negotiating slopes, dynamic balance control and speed modulation. In addition, it may prove particularly helpful to provide powered plantar flexion during gait practice. In healthy subjects, the ankle joint contributes more mechanical work to the gait cycle than either the hip or the knee [23]. A powered lower limb orthosis could mechanically assist at the ankle joint while allowing subjects more freedom in their gait pattern during rehabilitation. A powered orthosis might be especially useful for patients who are ready to practice more demanding locomotor tasks like turning and obstacle avoidance.

Although lower limb orthoses have traditionally been passive, there have been attempts at providing powered versions. Importantly, the main goal of these previous prototypes has been to create assistive technology. These research teams envisioned replacing lost motor capabilities rather than improving motor capabilities through therapy. In the 1970s, Vukobratovic built pneumatic robotic exoskeletons for human walking [24, 25]. Seireg et al. developed a hydraulic device with a dual axis hip, dual axis ankles, and a single axis knee joint [26]. A more recent attempt was the Powered Gait Orthosis (PGO), a four bar linkage and CAM system [27]. Blaya et al. built an orthosis to assist drop foot gait [28]. In addition, there are other groups developing powered lower limb orthoses to replace lost motor function of patients [29, 30]. All of these prototypes have had difficulty with achieving sufficient energy density. That is, to make the devices truly portable so they can function as assistive technology, the actuators and batteries have to be powerful and lightweight while providing many hours of use.

Powered orthoses for motor rehabilitation do not face as many technical difficulties as those intended for use as assistive technology. Using a powered lower limb orthosis as gait therapy would restrict the device to the clinic. As a result, control hardware and power do not have to be on board the orthosis itself. Electric, hydraulic, or pneumatic energy could be supplied through a tether that includes cables connected to a desktop computer. A therapist could have real-time control over the magnitude and timing of mechanical assistance during gait practice. In addition, sensors could provide feedback to the therapist about the performance of the patient. As rehabilitation progresses, the patient could be weaned by decreasing orthosis assistance. This would enforce active patient participation over the training period. The ultimate goal would be to divorce the patient from the powered orthosis as motor capabilities improved.

The following sections describe our initial attempts at developing powered orthoses for motor rehabilitation and discuss alternative uses for the orthoses in studying motor adaptation during human walking.

II. DESIGN

We have constructed orthoses capable of providing mechanical assistance at the ankle and knee [31-33] (Figure 1). They are comfortable, lightweight and allow unencumbered movement through a normal range of motion during walking. The orthoses are custom built for each subject from a combination of carbon fiber and polypropylene. Steel hinge joints allow sagittal plane movements at ankle and knee joints. The design of the ankle-foot orthoses have been described in detail previously [33-35].

Artificial pneumatic muscles attached to the orthoses provide flexion and extension torque at individual joints. Artificial pneumatic muscles can provide high power outputs, are relatively light-weight, and possess inherent compliance [36-39]. The artificial muscle consists of an expandable internal bladder housed inside a braided polyester shell. When the bladder is inflated, the braided shell constrains its expansion. As the volume of the internal bladder increases with greater air pressure, the pneumatic muscle shortens and/or produces tension if coupled to a mechanical load. The mechanical properties of artificial pneumatic muscles have been described in detail elsewhere [40-42]. We have utilized several different signals (footswitch, proportional myoelectric, push button, etc.) to control artificial pneumatic muscle activation during locomotion.

Figure 1. ABOVE LEFT: An ankle-foot orthosis with an artificial pneumatic plantar flexor muscle. ABOVE RIGHT: A knee-ankle-foot orthosis with artificial pneumatic muscles providing flexion and extension torque at each joint. Plastic tubes provide compressed air to the artificial muscles from an external air source.

III. PRELIMINARY STUDIES

A. Footswitch-Controlled Unilateral Ankle-Foot Orthosis

The goal of this first study was to quantify the mechanical performance of a powered ankle-foot orthosis that provided plantar flexor assistance during human walking. We placed our initial emphasis on this orthosis because ankle plantar flexor power dominates the mechanical work required for walking [23]. We examined three healthy subjects as they walked over a range of speeds. The artificial plantar flexor muscle was activated by a foot switch controller. When the subject's forefoot contacted the ground, the artificial muscle contracted maximally. When the subject's forefoot lost ground contact at toe-off, the artificial muscle relaxed completely. We used this simple bang-bang controller to quantify orthosis performance in a situation that limited effects of motor adaptation by the subject.

We found that net ankle moments during walking were similar when the subjects walked with the orthosis active and passive (i.e. artificial pneumatic muscles were inactive). When the artificial pneumatic muscles were active, the orthosis generated ~57% of the peak ankle plantar flexor torque during stance (Figure 2) and performed ~70% of the plantar flexor positive work done during normal walking [30]. The results of this study demonstrated that our orthosis was able to create substantial plantar flexor torque and performed considerable work during human walking by neurologically intact subjects.

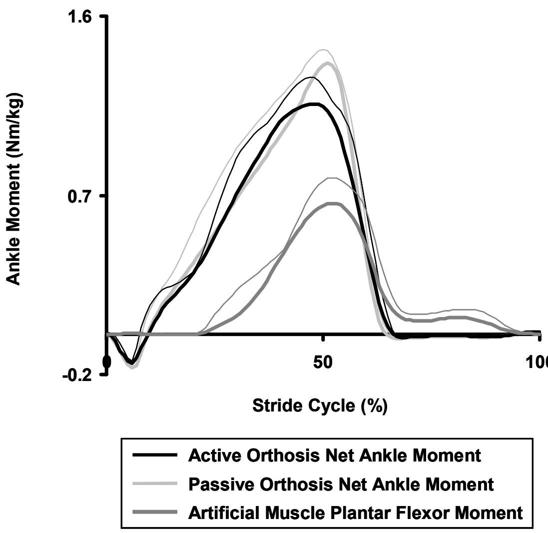

Figure 2. Mean + standard deviation of net ankle moments and artificial pneumatic muscle moments for the three subjects during overground walking at 1.0 m/s. Plantar flexion moments are in the positive direction. Subjects walked with very similar net ankle moments for the passive and active orthosis conditions when using foot-swtitch control.

B. EMG-Controlled Unilateral Ankle-Foot Orthosis

Based on the performance of the ankle-foot orthosis with footswitch control, we felt that the orthosis might also be helpful in studying motor adaptation in healthy subjects. Using proportional myoelectric control, we were able to activate the artificial pneumatic plantar flexor based on the amplitude of a subject's soleus electromyography (EMG). This effectively amplified torque production of soleus activation. Six subjects walked at 1.25 m/s for thirty minutes with the ankle-foot orthosis active.

Initially subjects were not able to fluidly control the supplemental plantar flexor torque. Subjects walked on their toes for the first several minutes (Figure 3). Soleus EMG during these first few minutes was active throughout the gait cycle, resulting in accompanying artificial plantar flexor torque. During thirty minutes of walking with the active orthosis, subjects gradually returned to kinematic patterns similar to normal. This was accomplished by modulating soleus muscle activity. Soleus EMG amplitude was reduced by about 50% and became timed to occur just prior to toe off. Results from this study demonstrated that subjects were able to selectively modulate muscle activity during walking in response to altered musculoskeletal mechanics. Studies of this nature may be helpful in elucidating general principles of motor control and adaptation during human locomotion. The findings also suggest that powered orthoses might be useful for shaping gait patterns in neurologically impaired individuals.

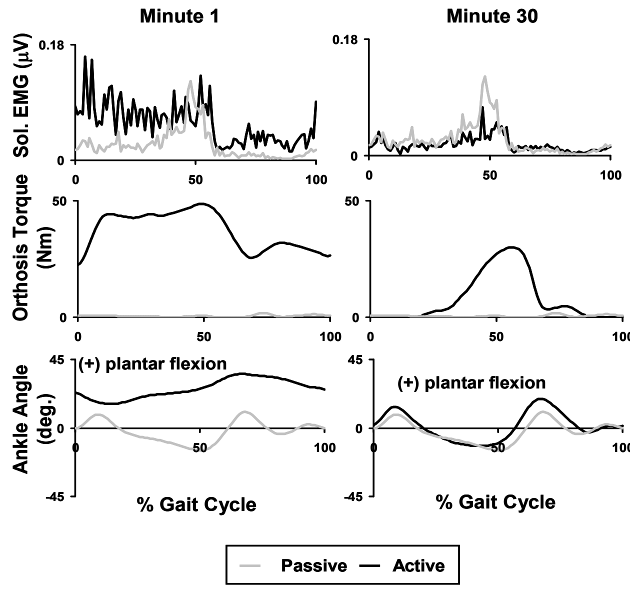

Figure 3. Mean step cycle data from a representative subject during 30 minutes of continuous walking. The subject was wearing a powered ankle-foot orthosis under proportional myoelectric control. Soleus EMG was activating an artificial plantar flexor muscle. Graphs compare data with a passive orthosis to data from the first and thirtieth minutes of active orthosis walking. Early in the trial the orthosis greatly altered joint kinematics, but subjects learned to incorporate the orthosis power into a normal gait pattern with practice.

C. Push Button-Controlled Bilateral Ankle-Foot Orthoses

We tested the ankle-foot orthoses on subjects with partial paralysis using a novel controller. Hand-held push buttons provided proportional activation of artificial plantar flexors on two ankle-foot orthoses (Figure 4). Elastic cords provided limited dorsiflexor torque to assist toe clearance. Six subjects with chronic incomplete spinal cord injury walked at 0.54 m/s under three different conditions: (1) without the powered orthoses, (2) with passive orthoses, and (3) with active orthoses under push button control by a therapist. In addition, three of the subjects completed an additional condition: (4) with active orthoses under push button control by the subject. A harness provided torso support at 30% or 50% body weight depending on the subject's ability. Elastic cords increased lateral stability at the waist. We measured EMG, joint kinematics and orthosis torque assistance. Foremost, we wanted to determine if the added mechanical assistance of the orthosis would decrease neuromuscular recruitment in the soleus and gastrocnemius.

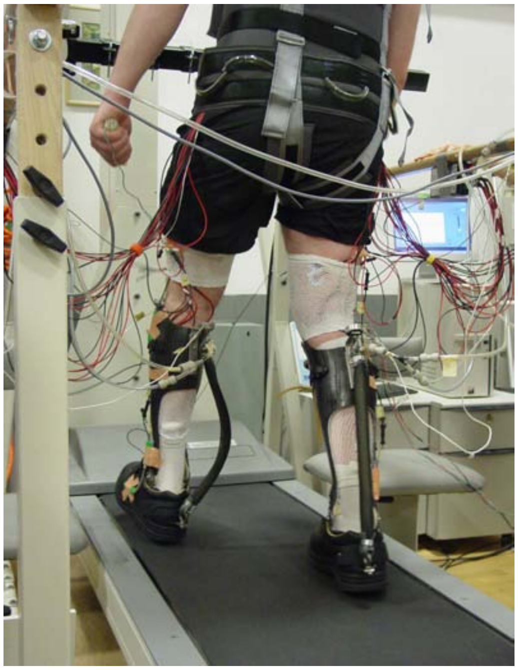

Figure 4. A patient uses bilateral ankle foot orthoses and push button controllers to assist locomotor training.

It was possible that the powered orthoses would decrease soleus and gastrocnemius recruitment because they unloaded the biological muscles during walking. If orthosis assistance reduces muscle activation and promotes passivity in the subject, it could hinder activity-dependent plasticity during rehabilitation.

We found that powered plantar flexion assistance controlled by a therapist or by the subject did not decrease soleus or gastrocnemius recruitment (Figure 5). The added torque at the ankle joint did provide increased plantar flexion at the end of the stance phase however. Increased push-off at the end of stance may have provided more appropriate sensory feedback to locomotor neural networks because of more normal gait dynamics. The enhanced sensory feedback could then evoke greater muscle activation. Future experiments need to examine the changes in gait dynamics and sensory feedback in spinal cord injury subjects more closely.

A few of the subjects could not use the push button controllers during treadmill walking. They felt it required too much mental concentration. The subjects that could complete the "patient-controlled" push button condition commented that it was good to have control over the orthoses.

IV. FUTURE DESIGNS AND FUTURE STUDIES

There are several modifications that could improve the performance of powered lower limb orthoses. We are currently working on an adjustable design that could fit multiple subjects. It may be possible to integrate other actuator technologies to produce higher forces and powers with greater control bandwidth. More advanced control algorithms may be able to detect the user's intent and provide assistance on an 'as needed' basis.

With the current design, we plan on exploring questions related to locomotor adaptation in both healthy and neurologically impaired subjects. The current ankle-foot orthosis presented here may be valuable in probing the relationship between gait mechanics and metabolic cost.

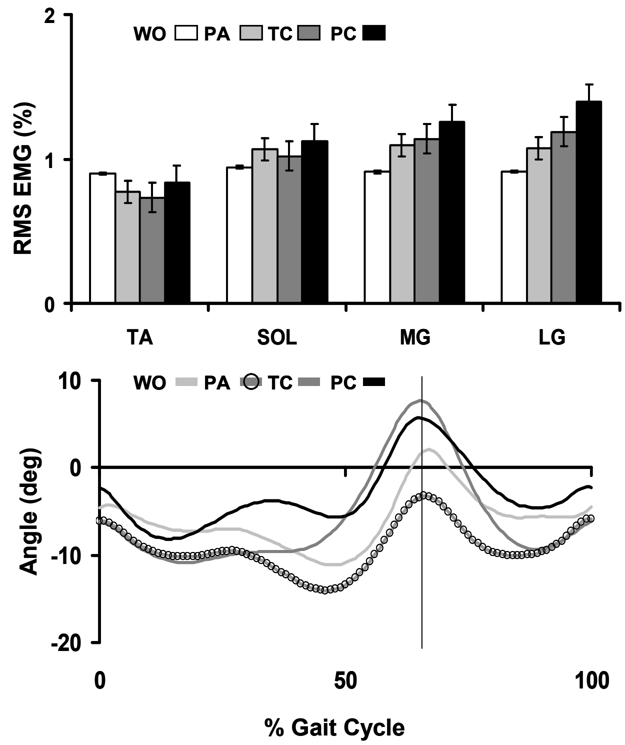

Figure 5. TOP: Normalized root mean square EMG of the tibialis anterior (TA), soleus (SOL), medial gastrocnemius (MG), and lateral gastrocnemius (LG) under four conditions: without orthoses (WO), wearing orthoses passively (PA), wearing orthoses under therapist control (TC), and wearing orthoses under patient control (PC). Bars are means ± SE. BOTTOM: Mean ankle angle during the gait cycle. Plantar flexion is +.

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