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An improved powered ankle–foot orthosis using proportional myoelectric control

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Abstract

We constructed a powered ankle–foot orthosis for human walking with a novel myoelectric controller. The orthosis included a carbon fiber and polypropylene shell, a metal hinge joint, and two artificial pneumatic muscles. Soleus electromyography (EMG) activated the artificial plantar flexor and inhibited the artificial dorsiflexor. Tibialis anterior EMG activated the artificial dorsiflexor. We collected kinematic, kinetic, and electromyographic data for a naive healthy subject walking with the orthosis. The current design improves upon a previous prototype by being easier to don and doff and simpler to use. The novel controller allows naive wearers to quickly adapt to the orthosis without artificial muscle co-contraction. The orthosis may be helpful in studying human walking biomechanics and assisting patients during gait rehabilitation after neurological injury.

Keywords: Locomotion; Exoskeleton; Gait; Rehabilitation

1. Introduction

Our goal was to improve upon a previously designed, powered ankle–foot orthosis [1]. The orthosis is intended as a tool for studying gait biomechanics and rehabilitation after neurological injury. The previous prototype had univalve shank and foot sections made from carbon fiber and connected by a plastic hinge joint [1]. It had two artificial pneumatic muscles to provide dorsiflexor and plantar flexor torques about the ankle. That prototype was effective in producing high external torques to the ankle but had limitations. The univalve designs for shank and foot sections made it difficult to don and doff. Changing artificial pneumatic muscles or force sensors was time consuming due to attachment mechanisms. Lastly, the proportional myoelectric control was functional but produced a high degree of co-activation when the two antagonistic artificial muscles were on the orthosis at the same time. The revised orthosis prototype presented here overcomes limitations in the previous design and utilizes a new control method facilitating walking by new users. We tested the orthosis on a naive subject to determine how much torque the orthosis would produce during walking.

2. Methods

We fabricated the orthosis shell from carbon fiber, polypropylene, and a metal hinge based on a cast of the subject's lower limb (Fig. 1). Instead of using a univalve design for the shank section, we used a bivalve carbon fiber design with plastic buckles to increase ease of donning and doffing. The foot section was made from polypropylene for

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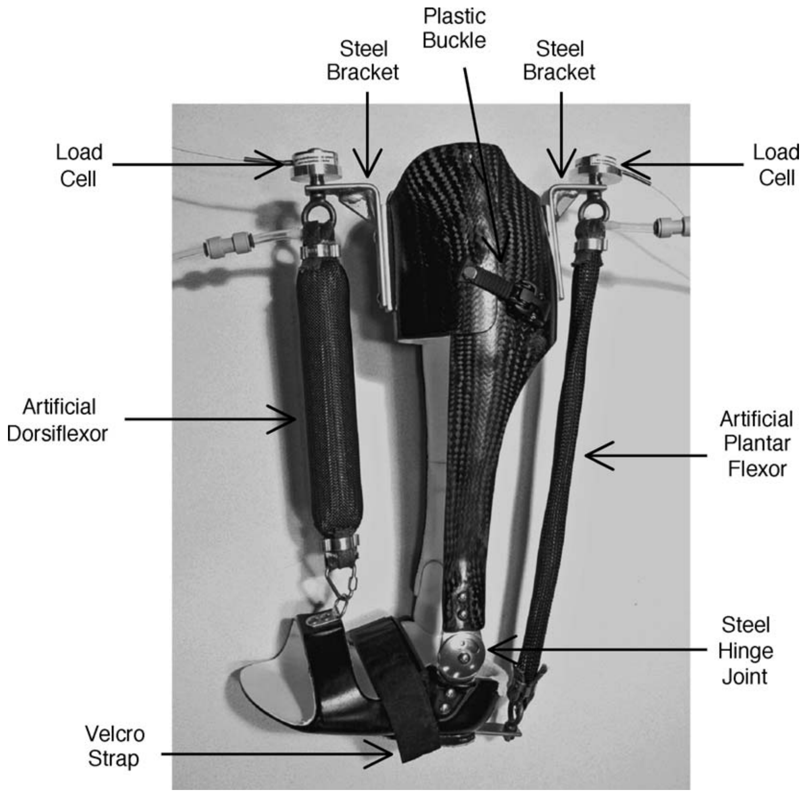

Fig. 1. Pneumatically powered ankle–foot orthosis. The shank section is made from carbon fiber and the foot section is made from polypropylene. The artificial dorsiflexor muscle is maximally inflated and the artificial plantar flexor muscle is relaxed. Load sensors in series with the artificial muscles monitored artificial muscle tension. Both artificial pneumatic muscles had moment arms of 11 cm for the range of motion during walking.

increased pliability. We attached stainless steel brackets to the anterior and posterior shank sections for connecting artificial pneumatic muscles. A stainless steel D-ring on the dorsum of the foot section provided an attachment for the artificial dorsiflexor.

We added two artificial pneumatic muscles to provide actuation of orthosis plantar flexion and dorsiflexion. Artificial pneumatic muscles provide high power outputs with relatively light weights and possess inherent compliance that facilitates their use with human interaction [2–5]. Nylon tubing supplied compressed air to the muscles (0–6.2 bar) from proportional pressure regulators. We attached solenoid valves to the tubing to facilitate exhaust during deactivation of the artificial pneumatic muscles. The ankle–foot orthosis had a total mass, including artificial muscles and force transducers, of 1.7 kg.

We implemented proportional myoelectric control through a desktop computer and real-time control board (dSPACE, Inc., Northville, MI). The program regulated air pressure in the artificial pneumatic muscles proportional to the processed muscle activation pattern. Soleus electromyography (EMG) activated the artificial plantar flexor muscle and tibialis anterior EMG activated the artificial dorsiflexor muscle. EMG signals from the soleus and tibialis anterior were high-pass filtered with a second-order Butterworth filter (cut-off frequency 20 Hz) to remove movement artifact, full wave rectified, and low-pass filtered with a second order Butterworth filter (cut-off frequency 10 Hz) to smooth the signal. Threshold cut-offs eliminated background noise and adjustable gains scaled the control signals. When the low-pass filtered soleus EMG signal was above threshold, the software completely inhibited activation of the artificial dorsiflexor muscle.

One healthy subject (age 30 years, body mass 91 kg) walked with the orthosis to test its performance and comfort during gait. We collected three-dimensional joint kinematics (120 Hz, Motion Analysis Corporation, Santa Rosa, CA), ground reaction forces (1200 Hz, AMTI, Inc. Watertown, MA), and lower limb surface EMG (1200 Hz, Konigsberg Instruments, Inc., Pasadena, CA) as the subject walked overground at 1.25 m/s. For the proportional myoelectric control, we set the threshold to 20 μV for soleus EMG and to 5 μV for tibialis anterior EMG. The subject wore reflective markers during all trials so that we could measure joint kinematics and kinetics. We calculated joint angle displacement after smoothing marker position data with a fourth order Butterworth low-pass filter (cut-off frequency 6 Hz) with zero lag. We used commercial software (Visual3D, C-Motion, Inc., Rockville, MD) to calculate internal muscle moments about the lower limb joints based on kinematic marker and force platform data (http://www.c-motion.com/support/FAQ_Kinetics.php). Lower limb inertial properties were estimated based on anthropometric measurements of the subjects [6]. For the inverse dynamics calculations during passive and active orthosis conditions, we modified segment inertial parameters of the lower limb to take account of orthosis mass and moment of inertia. We placed bipolar surface electrodes on the soleus (SOL) and tibialis anterior (TA). We visually inspected muscle activation during manual muscle tests to minimize crosstalk, moving electrode placements as necessary. EMG amplifier bandwidth was 1000 Hz.

3. Results

The improved powered ankle–foot orthosis (Fig. 1) overcame limitations of the previous design. The present orthosis was easier to don and doff than the previous version. The bivalve design and buckles on the shank section allowed the subject to put on or take off the orthosis in less than a minute. The polypropylene foot section was more flexible than the previous carbon fiber foot section, providing a very comfortable fit.

The subject was able to walk overground immediately after turning on the proportional myoelectric control. The artificial plantar flexor muscle provided a peak torque of 50.7 N m, 36% of the peak plantar flexor torque during the passive orthosis condition (140 N m). The artificial dorsiflexor muscle provided a peak torque of 20.7 N m, 123% of the peak dorsiflexor torque during the passive orthosis condition (16.8 N m).

4. Discussion

Several researchers have built powered lower limb orthoses for human locomotion in the past. Most previous designs have been intended as assistive technology for disabled individuals [7–11] or as human performance augmentation for manual laborers [12–15]. For those uses, portable energy supplies and robust control algorithms are critical to the success of the device.

The powered ankle–foot orthosis presented here is intended for basic science studies on human walking and possibly for rehabilitation after neurological injury. As a result, the orthosis could be used in a gait laboratory or rehabilitation clinic where compressed air and electrical power is easily provided. Recent breakthroughs in clinical neuroscience have revealed that humans with spinal cord injury or stroke can increase their motor capabilities through intense task-specific practice [16–19]. For gait rehabilitation, locomotor training often requires the manual assistance of three or more physical therapists and a harness to provide partial body weight support [20]. Robotic exoskeletons or powered orthoses could reduce therapist manual labor. Researchers are testing several robotic devices for repetitive step training [21,22] but none currently provide plantar flexion torque at the ankle. The lack of plantar flexion torque could be a particularly significant aspect as plantar flexor muscles perform more positive mechanical work than the knee or hip during normal walking [23]. It would be helpful to determine if increasing plantar flexion power output in neurologically impaired subjects helps to improve overall gait dynamics and neuromuscular recruitment. From a basic science perspective, powered robotic devices for the upper limbs have been incredibly useful in understanding the neural control of reaching movements [21,22]. However, there have been relatively few studies that have used powered robotic devices for the lower limbs to study the neural control and biomechanics of locomotion [21]. With this orthosis, it would be possible to study motor adaptation during human locomotion in a very controlled manner. Biomechanists could quantify how fast humans can adjust muscle activation patterns in response to increased strength of the musculoskeletal system (i.e. biological and artificial combined). It would also be possible to determine in humans if motor memories for locomotor control are formed with practice and to quantify how long they last. These types of neural control studies could provide a good a means to study locomotor adaptation in humans without invasive procedures.

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References

[1] Ferris DP, Czerniecki JM, Hannaford B. An ankle–foot orthosis powered by artificial pneumatic muscles. J Appl Biomech 2005;21:189–97.
[2] Klute GK, Czerniecki JM, Hannaford B. Artificial muscles: actuators for biorobotic systems. Int J Robotics Res 2002;21:295–309.
[3] Davis S, Tsagarakis N, Canderle J, Caldwell DG. Enhanced modelling and performance in braided pneumatic muscle actuators. Int J Robotics Res 2003;22:213–27.
[4] Tondu B, Lopez P. Modeling and control of McKibben artificial muscle robot actuators. IEEE Cont Syst Mag 2000;20:15–38.
[5] Reynolds DB, Repperger DW, Phillips CA, Bandry G. Modeling the dynamic characteristics of pneumatic muscle. Ann Biomed Eng 2003;31:310–7.
[6] Zatsiorsky VM. Kinetics of human motion. Champaign, IL: Human Kinetics, 2002.
[7] Ruthenberg BJ, Wasylewski NA, Beard JE. An experimental device for investigating the force and power requirements of a powered gait orthosis. J Rehab Res Dev 1997;34:203–13.
[8] Vukobratovic M, Borovac B, Surla D, Stokic D. Biped locomotion: dynamics, stability, control and application, vol. 7. Berlin: Springer–Verlag, 1990.
[9] Hughes J. Powered lower limb orthotics in paraplegia. Paraplegia 1972;9:191–3.
[10] Seireg A, Grundman JG. Design of a multitask exoskeletal walking device for paraplegics. In: Ghista DNs, Ghista DN, editors. Biome-

chanics of medical devices. New York: Marcel Dekker, Inc.; 1981. p. 569–639.

[11] Blaya JA, Herr H. Adaptive control of a variable-impedance ankle–foot orthosis to assist drop-foot gait. IEEE Trans Neural Systems Rehab Eng 2004;12:24–31.

[12] Pratt JE, Krupp BT, Morse CJ, Collins SH. The RoboKnee: an exoskeleton for enhancing strength and endurance during walking.. In: IEEE International Conference on Robotics and Automation; 2004.p. 2430–5.

[13] Jacobsen SC, Olivier M, Smith FM, Knutti DF, Johnson RT, Colvin GE, Scroggin WB. Research robots for applications in artificial intelligence, teleoperation and entertainment. Int J Robotics Res 2004;23:319–30.

[14] Reinkensmeyer DJ. Rehabilitators. In: Kutz Ms, Kutz M, editors. Standard handbook of biomedical engineering & design, 35. McGraw-Hill; 2003. p. 1–17.

[15] Kazerooni H. The human power amplifier technology at the University of California, Berkeley. Robot Auton Syst 1996;19:179–87.

[16] Wernig A, Muller S, Nanassy A, Cagol E. Laufband therapy based on 'rules of spinal locomotion' is effective in spinal cord injured persons. Eur J Neurosci 1995;7:823–9.

[17] Harkema SJ. Neural plasticity after human spinal cord injury: application of locomotor training to the rehabilitation of walking. Neuroscientist 2001;7:455–68.

[18] Barbeau H, Norman K, Fung J, Visintin M, Ladouceur M. Does neurorehabilitation play a role in the recovery of walking in neurological populations? Ann NY Acad Sci 1998;860:377–92.

[19] Dietz V, Wirz M, Colombo G, Curt A. Locomotor capacity and recovery of spinal cord function in paraplegic patients: a clinical and electrophysiological evaluation. Electroenceph Clin Neurophysiol 1998;109:140–53.

[20] Behrman AL, Harkema SJ. Locomotor training after human spinal cord injury: a series of case studies. Phys Ther 2000;80:688–700.

[21] Reinkensmeyer DJ, Emken JL, Cramer SC. Robotics, motor learning, and neurologic recovery. Annu Rev Biomed Eng 2004;6:497–525.

[22] Hesse S, Schmidt H, Werner C, Bardeleben A. Upper and lower extremity robotic devices for rehabilitation and for studying motor control. Curr Opin Neurol 2003;16:705–10.

[23] Gitter A, Czerniecki JM, DeGroot DM. Biomechanical analysis of the influence of prosthetic feet on below-knee amputee walking. Am J Phys Med Rehab 1991;70:142–8.