34th Annual International Conference of the IEEE EMBS
San Diego, California USA, 28 August - 1 September, 2012

Enhancing Stance Phase Propulsion during Level Walking by Combining FES with a Powered Exoskeleton for Persons with Paraplegia*

Kevin H. Ha, Hugo A. Quintero, Ryan J. Farris, and Michael Goldfarb, *Members, IEEE*

Abstract—**This paper describes the design and implementation of a cooperative controller that combines functional electrical stimulation (FES) with a powered lower limb exoskeleton to provide enhanced hip extension during the stance phase of walking in persons with paraplegia. The controller utilizes two sources of actuation: the electric motors of the powered exoskeleton and the user's hamstrings activated by FES. It consists of a finite-state machine (FSM), a set of proportional-derivative (PD) controllers for the exoskeleton and a cycle-to-cycle adaptive controller for muscle stimulation. Level ground walking is conducted on a single subject with complete T10 paraplegia. Results show a 34% reduction in electrical power requirements at the hip joints during the stance phase of the gait cycle with the cooperative controller compared to using electric motors alone.**

I. Introduction

Spinal cord injury is a devastating condition with no available cure to date. Despite ongoing research in areas such as stem cell therapy and neuronal regeneration, these fields are still in preclinical states and thus far have made little clinical impact at the patient level. On the other hand, functional electrical stimulation (FES) is already in use in areas such as breathing, bowel and bladder function, and hand grasp [1, 2]. In the past few decades, many researchers have developed FES-based gait restoration systems for persons with paraplegia [3-6], including the commercially-available Parastep system [7]. However, these devices have shown limited everyday usability due to some key limitations such as muscle fatigue, uneven muscle response, and lack of sensing for control purposes, resulting in limited run time and inconsistent gait motions [1, 8]. Some researchers have addressed these problems using hybrid systems combining FES with computer-controlled passive lower limb braces [9-14]. Others have developed fully-powered, electric motor-actuated lower limb exoskeletons without the use of FES [15-22].

The authors have recently developed a powered lower limb exoskeleton, shown in Fig. 1 [23, 24]. Whereas prior publications by the authors have described the design and demonstrated the efficacy of the powered exoskeleton [23, 24], this paper describes the use of supplemental FES of the hamstring muscles to enhance stance phase propulsion during level walking. This hybrid approach has a few advantages compared to using either FES or electric actuators alone. Compared to using FES alone, the powered exoskeleton provides joint motions that are otherwise difficult to achieve consistently (e.g. hip flexion). Even for motions that can be achieved using FES, the exoskeleton ensures that the joint trajectories stay consistent in the presence of time-varying muscle behavior, providing consistent and repeatable gait. Compared to using a powered exoskeleton alone, the addition of FES reduces electrical power consumption while providing additional joint torques. Furthermore, FES provides physiological benefits to persons with paraplegia [1].

In this work, the authors present a cooperative controller combining the powered exoskeleton and FES of the hamstring muscles during the stance phase of the gait cycle (i.e. during hip extension of the weight-bearing leg).

II. Hamstrings as Hip Extensors

The hamstrings consist of three heads: semitendinosus, semimembranosus, and the long head of biceps femoris. They are biarticular muscles spanning both the hip and knee joints with their two main actions being knee flexion and hip extension [25]. Hamstring contraction generates an extension torque at the hip when the knee joint is immobilized, as it is in this application by the normally-locked knee joints of the exoskeleton.

III. Hardware

A. Vanderbilt Exoskeleton

The Vanderbilt exoskeleton is a powered exoskeleton described in [23, 24] for gait restoration in persons with paraplegia. It includes actuated hip and knee joints and has a mass of 12 kg (26.4 lb) including the 29.6 V, 3.9 A·hr lithium polymer battery. The knee joints are normally-locked, so that the unit does not collapse in the event of a power failure. Also, because the brakes are normally-locked, they require no additional electrical power to immobilize the knees during hamstring contractions (so that hip extension occurs). The device does not have foot and ankle sections, as it is designed

Figure 1. Vanderbilt exoskeleton

*Research supported by the U.S. Department of Health and Human Services under National Institutes of Health Grant 1R01HD059832-01A1.

K. H. Ha, H. A. Quintero, R. J. Farris, and M. Goldfarb are with Vanderbilt University, Nashville, TN 37235 USA (e-mail: kevin.h.ha@vanderbilt.edu, hugo.a.quintero@vanderbilt.edu, ryan.farris@vanderbilt.edu, michael.goldfarb@vanderbilt.edu).

to be used with standard ankle foot orthoses (AFOs). An 80 MHz PIC32 microprocessor on the electronic board in each thigh segment provides low level control of the joints. High level control runs on MATLAB Real-Time Workshop and communicates with the exoskeleton via an RS-232 interface.

B. *Stimulator*

A self-contained stimulator board located in each thigh segment produces current-controlled symmetrical biphasic stimulation waves at 50 Hz and at a 2% duty cycle (i.e., a 200 ms pulsewidth each for the upwave and the downwave for a combined pulsewidth of 400 ms). Commercially-available TENS surface electrodes are used.

IV. CONTROLLER

A. *Exoskeleton Control Architecture*

The exoskeleton controller consists of a finite-state machine (FSM) at high level, and a set of proportional-derivative (PD) controllers at low level, as described in [23] and also shown in part in Fig. 2. The high level FSM uses sensory data from the exoskeleton to determine the state, followed by the low level PD controller commanding appropriate trajectories for the joints. For example, if the subject leans forward while in the right forward state (i.e. standing with right foot forward), the FSM switches the state from right forward to right stance. Once the state goes into right stance, the low level PD controller extends the right hip and flexes the left hip, while flexing and extending the left knee (to take a left step).

B. *Cooperative Controller Combining FES with Exoskeleton*

The goal of the cooperative controller is to maximize the use of muscle power while preventing the muscles and the exoskeleton from working against each other. That is, the hamstrings should help the exoskeleton extend the hip when extension torques are needed while not affecting the system when flexion torques are needed. To achieve this, the controller uses a constant stimulation level and varies the timing of stimulation based on both the FSM and hip joint torques from previous steps.

Fig. 2 illustrates the states in which hamstring contractions occur. Because the hip extends during the stance phase of the corresponding side, right hamstrings are stimulated in right stance (when taking a left step) and left hamstrings in left

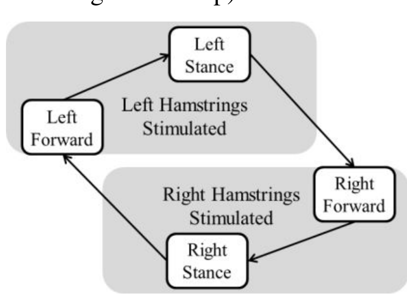

Figure 2. State-flow diagram showing the states involved in walking and when hamstring stimulation occurs. Left hamstrings are stimulated in left forward (standing with left foot forward) and left stance (taking a right step). Right hamstrings are stimulated in right forward (standing with right foot forward) and right stance (taking a left step).

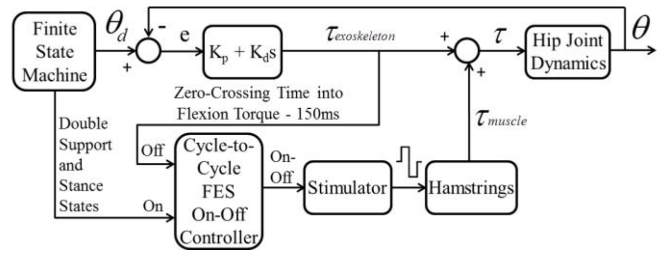

Figure 3. Cooperative controller for the Vanderbilt exoskeleton with cycle-to-cycle adaptive FES timing. Hamstrings are stimulated in double support and stance states. The controller adapts the timing of stimulation within each state based on hip torque data from previous steps.

stance (when taking a right step). The hamstrings are also stimulated in double support (standing in right forward or left forward states) because the exoskeleton requires a significant amount of hip extension torques for torso stability. Only the forward leg is stimulated to achieve smoother transition into the stance phase and to reduce muscle fatigue. A 2-second time limit on stimulation is imposed in these states in case users choose to stay standing in double support without taking a step.

Within the right stance and left stance states, the controller adapts the timing of stimulation based on the zero-crossing time of hip torques from previous steps (Fig. 3). That is, the controller monitors when the exoskeleton shifts from providing extension torques to flexion torques and updates the stimulation off time for the following step. This is done to ensure that the muscles do not exert extension torques when the system needs flexion torques (towards the end of the step). The off time is updated based on the weighted moving average value. Although a preset stimulation off time at around 0.8 s would provide a reasonable performance for the subject in this paper, a cycle-to-cycle adaptive controller is used to account for step-to-step variations and to reduce the need for manual tuning between subjects. Stimulation is turned off a set duration before the zero-crossing time to account for delays from muscle physiology and dynamics [26]. The delay was observed to be approximately 150 ms for the subject in this paper. Therefore, the controller turns off the stimulation 150 ms before the zero-crossing time.

V. EXPERIMENTAL IMPLEMENTATION

A. *Subject*

The controller was implemented on a single paraplegic subject with a complete sensory and motor T10 injury (American Spinal Injury Association, ASIA, A classification). The subject was a 36-year-old male (1.85 m, 75 kg) and was already familiar with walking with the Vanderbilt exoskeleton (without FES).

B. *Experimental Setup*

Commercially available TENS surface electrodes were used to stimulate the hamstring muscles. With the subject lying prone on a mat, surface electrodes were applied to the posterior thigh. Before donning the exoskeleton, hamstring contractions and resulting hip extensions were visually confirmed with the subject on his side (Fig. 4a). After donning the exoskeleton, the subject was instructed to walk around at

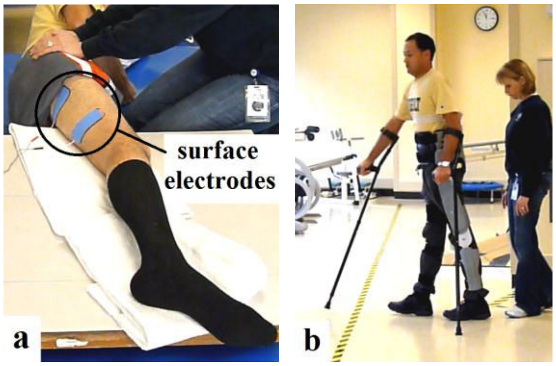

Figure 4. Experimental setup. a) Surface electrodes placed over the posterior thigh. b) Subject walking with the exoskeleton.

his own pace (Fig. 4b). The subject walked in two 10-minute sessions (20 minutes total), during which FES was turned on and off every few minutes.

VI. Results

During 20 minutes of periodic walking, the subject took 156 steps total: 76 with FES and 80 without FES. Fig. 5 shows the exoskeleton hip joint torque averaged over all steps taken during the stance phase with FES (solid red) and without FES (dashed blue). Fig. 6 shows the exoskeleton hip power for the same interval. The hip joint torque was lower with FES during the first 0.85 s (FES was assisting the exoskeleton), and the torque requirements were similar towards the end of the step with or without FES (FES was not working against the exoskeleton when flexion torques were needed), indicating that the cycle-to-cycle adaptive controller turned off the stimulation at appropriate times.

The RMS hip torque the exoskeleton provided throughout the stance phase was 15.9 N·m with FES compared to 22.6 N·m without FES for a reduction of 30% (Table I). The RMS power requirement during the stance phase was 4.1 W with FES compared to 6.2 W without FES for a reduction of 34% (Table I). That is, the subject's hamstrings provided 30% of the hip torque and 34% of the total power during hip extension while the exoskeleton provided the remaining required torque and power, respectively.

VII. Discussion

A. Overall Power Consumption

The use of hamstrings for hip extension lowered power consumption during stance phases. Although not explicitly presented in this paper, power consumption during double support phases was also lowered, and other phases of the gait

TABLE I
RMS Exoskeleton Hip Torque and Power With and Without FES

	Without FES	**With FES**	**Contribution from FES**
RMS Hip Torque (SD)	22.6 N·m (7.6)	15.9 N·m (7.4)	30%
RMS Hip Power (SD)	6.2 W (3.1)	4.1 W (2.5)	34%

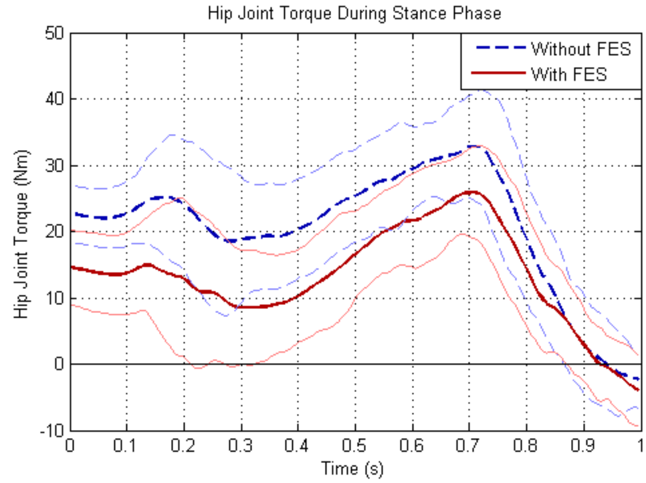

Figure 5. Exoskeleton hip joint torque during extension without FES (dashed blue) and with FES (solid red). Thick lines are joint torques averaged over all steps taken. Thin lines indicate one standard deviation.

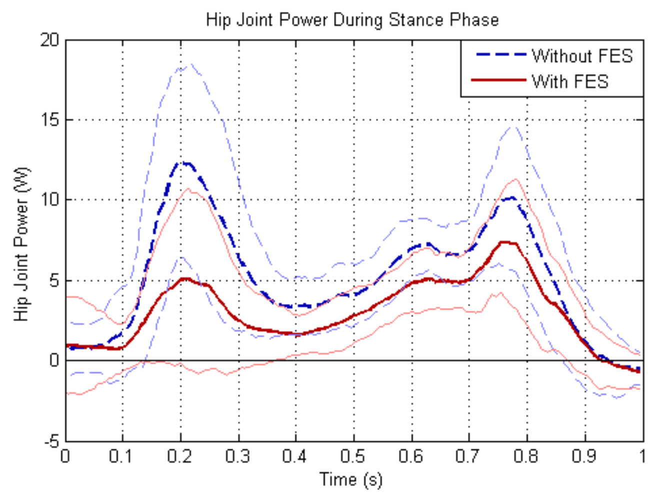

Figure 6. Exoskeleton hip joint power during extension without FES (dashed blue) and with FES (solid red). Thick lines are joint power averaged over all steps taken. Thin lines indicate one standard deviation.

cycle remained unaffected. Power consumption by the stimulator is minimal, as it provides 120 mA of current across approximately 1 kΩ of impedance at 2% duty cycle (i.e., approximately 300 mW on average). Because the knee brakes are normally locked and only require power to unlock them, the device does not require any additional electrical power to ensure that hamstring contractions result in hip extension rather than knee flexion. In short, one would expect that the overall (battery) power consumption of the exoskeleton would be substantially lower with FES relative to the electrical power consumption without.

B. Stimulation Level

Initially, the stimulation levels were varied between steps using a cycle-to-cycle adaptive controller, but the subject's hamstrings were incapable even at maximum stimulation to provide greater amounts of the hip torque than those required during the stance phase of walking. That is, the adaptive controller consistently saturated at the maximum stimulation amplitude of 120 mA. Consequently, the authors used a preset level of 120 mA for the subject in this paper. In the future, adaptive stimulation level may be implemented on a different subject with stronger hamstrings or could be used on the subject in this paper, assuming he builds hamstring muscle strength over time.

C. Muscle Fatigue

Within the data presented in this paper, signs of muscle fatigue were observed. Although not shown in this paper, hamstring contribution for hip extension declined towards the end of the session compared to the beginning. Nevertheless, muscle fatigue did not affect the overall system performance (since the exoskeleton supplies the torque that the hamstrings cannot), and the subject was able to continue walking with consistent and repeatable gait behavior. This demonstrates that combining FES with the exoskeleton provides safe and reliable gait even in the presence of muscle fatigue.

VIII. Conclusion

The authors have developed a cooperative controller combining FES of the hamstrings with a powered exoskeleton for hip extension during walking. The controller was implemented on a single subject with T10 complete sensory and motor paraplegia. Experimental results indicate that the hamstrings and the exoskeleton cooperatively generated hip extension torques without working against each other, resulting in reduced torque requirements and electrical power consumption during level walking.

Acknowledgment

The authors would like to thank C. Hartigan, MPT for her assistance.

References

[1] K. T. Ragnarsson, "Functional electrical stimulation after spinal cord injury: current use, therapeutic effects and future directions," *Spinal Cord,* vol. 46, pp. 255-74, Apr 2008.

[2] P. H. Peckham and J. S. Knutson, "Functional electrical stimulation for neuromuscular applications," *Annual Review of Biomedical Engineering,* vol. 7, pp. 327-360, 2005.

[3] G. R. Cybulski, R. D. Penn, and R. J. Jaeger, "Lower extremity functional neuromuscular stimulation in cases of spinal cord injury," *Neurosurgery,* vol. 15, pp. 132-46, Jul 1984.

[4] D. Graupe and K. H. Kohn, "Functional neuromuscular stimulator for short-distance ambulation by certain thoracic-level spinal-cord-injured paraplegics," *Surg Neurol,* vol. 50, pp. 202-7, Sep 1998.

[5] A. R. Kralj and T. Bajd, *Functional electrical stimulation : standing and walking after spinal cord injury*. Boca Raton, Fla.: CRC Press, 1989.

[6] E. B. Marsolais and R. Kobetic, "Functional walking in paralyzed patients by means of electrical stimulation," *Clin Orthop Relat Res,* pp. 30-6, May 1983.

[7] P. Gallien, R. Brissot, M. Eyssette, L. Tell, M. Barat, L. Wiart, and H. Petit, "Restoration of Gait by Functional Electrical-Stimulation for Spinal-Cord Injured Patients," *Paraplegia,* vol. 33, pp. 660-664, Nov 1995.

[8] P. H. Peckham and P. H. Gorman, "Functional Electrical Stimulation in the 21st Century," *Topics in Spinal Cord Injury Rehabilitation,* vol. 10, pp. 126-150, 2004.

[9] M. Goldfarb, K. Korkowski, B. Harrold, and W. Durfee, "Preliminary evaluation of a controlled-brake orthosis for FES-aided gait," *IEEE Trans Neural Syst Rehabil Eng,* vol. 11, pp. 241-8, Sep 2003.

[10] M. Solomonow, R. Baratta, S. Hirokawa, N. Rightor, W. Walker, P. Beaudette, H. Shoji, and R. Dambrosia, "The Rgo Generation-II - Muscle Stimulation Powered Orthosis as a Practical Walking System for Thoracic Paraplegics," *Orthopedics,* vol. 12, pp. 1309-1315, Oct 1989.

[11] W. K. Durfee and A. Rivard, "Design and simulation of a pneumatic, stored-energy, hybrid orthosis for gait restoration," *J Biomech Eng,* vol. 127, pp. 1014-9, Nov 2005.

[12] C. S. To, R. Kobetic, J. R. Schnellenberger, M. L. Audu, and R. J. Triolo, "Design of a variable constraint hip mechanism for a hybrid neuroprosthesis to restore gait after spinal cord injury," *Ieee-Asme Transactions on Mechatronics,* vol. 13, pp. 197-205, Apr 2008.

[13] R. Kobetic, C. S. To, J. R. Schnellenberger, M. L. Audu, T. C. Bulea, R. Gaudio, G. Pinault, S. Tashman, and R. J. Triolo, "Development of hybrid orthosis for standing, walking, and stair climbing after spinal cord injury," *J Rehabil Res Dev,* vol. 46, pp. 447-62, 2009.

[14] M. L. Audu, C. S. To, R. Kobetic, and R. J. Triolo, "Gait evaluation of a novel hip constraint orthosis with implication for walking in paraplegia," *IEEE Trans Neural Syst Rehabil Eng,* vol. 18, pp. 610-8, Dec 2010.

[15] Y. Ohta, H. Yano, R. Suzuki, M. Yoshida, N. Kawashima, and K. Nakazawa, "A two-degree-of-freedom motor-powered gait orthosis for spinal cord injury patients," *Proceedings of the Institution of Mechanical Engineers Part H-Journal of Engineering in Medicine,* vol. 221, pp. 629-639, Aug 2007.

[16] K. Suzuki, G. Mito, H. Kawamoto, Y. Hasegawa, and Y. Sankai, "Intention-based walking support for paraplegia patients with Robot Suit HAL," *Advanced Robotics,* vol. 21, pp. 1441-1469, Dec 2007.

[17] Y. Hasegawa, J. Junho, and Y. Sankai, "Cooperative walk control of paraplegia patient and assistive system," in *Intelligent Robots and Systems, 2009. IROS 2009. IEEE/RSJ International Conference on*, 2009, pp. 4481-4486.

[18] A. Tsukahara, Y. Hasegawa, and Y. Sankai, "Standing-up motion support for paraplegic patient with Robot Suit HAL," in *Rehabilitation Robotics, 2009. ICORR 2009. IEEE International Conference on*, 2009, pp. 211-217.

[19] K. Hian Kai, J. H. Noorden, M. Missel, T. Craig, J. E. Pratt, and P. D. Neuhaus, "Development of the IHMC Mobility Assist Exoskeleton," in *Robotics and Automation, 2009. ICRA '09. IEEE International Conference on*, 2009, pp. 2556-2562.

[20] A. Tsukahara, R. Kawanishi, Y. Hasegawa, and Y. Sankai, "Sit-to-Stand and Stand-to-Sit Transfer Support for Complete Paraplegic Patients with Robot Suit HAL," *Advanced Robotics,* vol. 24, pp. 1615-1638, 2010.

[21] P. D. Neuhaus, J. H. Noorden, T. J. Craig, T. Torres, J. Kirschbaum, and J. E. Pratt, "Design and evaluation of Mina: A robotic orthosis for paraplegics," in *Rehabilitation Robotics (ICORR), 2011 IEEE International Conference on*, 2011, pp. 1-8.

[22] A. M. Dollar and H. Herr, "Lower Extremity Exoskeletons and Active Orthoses: Challenges and State-of-the-Art," *Robotics, IEEE Transactions on,* vol. 24, pp. 144-158, 2008.

[23] H. A. Quintero, R. J. Farris, and M. Goldfarb, "Control and implementation of a powered lower limb orthosis to aid walking in paraplegic individuals," *IEEE Int Conf Rehabil Robot,* vol. 2011, pp. 1-6, Jun 2011.

[24] R. J. Farris, H. A. Quintero, and M. Goldfarb, "Preliminary Evaluation of a Powered Lower Limb Orthosis to Aid Walking in Paraplegic Individuals," *Neural Systems and Rehabilitation Engineering, IEEE Transactions on,* vol. 19, pp. 652-659, 2011.

[25] K. L. Moore, A. F. Dalley, and A. M. R. Agur, "Posterior thigh muscles," in *Clinically oriented anatomy*, 5th ed Philadelphia: Lippincott Williams & Wilkins, 2006, pp. 616-619.

[26] G. E. Loeb and C. Ghez, "The motor unit and muscle action," in *Principles of neural science*, E. R. Kandel, J. H. Schwartz, and T. M. Jessell, Eds., 4th ed New York: McGraw-Hill, Health Professions Division, 2000, pp. 674-694.