J. Biomed. Eng. Res : 344-350, 2010

Identifications of Reflex Muscle Activities and Joint Moments Triggered by Electrical Stimulation to Sole of the Foot during Lokomat Treadmill Walking

Yongchul Kim

Department of Rehabilitation Science and Technology, Daegu University, Gyeongsan-si, 712-714, Korea

(Received May 26, 2010. Accepted September 29, 2010)

Abstract

The aim of this study was to investigate the characteristics of the flexion withdrawal reflex modulated during Lokomat treadmill walking in people with spinal cord injury. The influence of the limb position and movement were tested in 5 subjects with chronic spinal cord injury. EMG activities from tibialis anterior and moments of the hip joint elicited by the foot stimulation were examined during Lokomat treadmill walking. To trigger the flexion withdrawal reflex during Lokomat treadmill walking, a train of 10 stimulus pulses was applied at the skin of the medial arch. The TA EMG activity was modulated during gait phase and the largest TA reflex was obtained after heel-off and initial swing phase. During swing phase, TA EMG was 40.9% greater for the extended hip position (phase 6), compared with flexed hip position (phase 8). The measured reflex moment of the hip joint was also modulated during gait phase. In order to characterize the neural contribution of flexion reflex at the hip joint, we compared estimated moments consisted of the static and dynamic components with measured moment of the hip joint. The mean static gains of reflex hip moments for swing and stance phase are -0.1, -0.8, respectively. The mean dynamic gains of reflex hip moments are 0.25 for swing, 0.75 for stance phase. From this study, we postulate that the joint moment and muscle response of flexion withdrawal reflex have the phase-dependent modulation and linear relationship with hip angle and angular velocity for swing phase during Lokomat treadmill walking.

Key words : Reflexive muscle activity, Joint moment, Lokomat system, Phase dependent modulation

Ⅰ. INTRODUCTION

Over the past thirty years, manual-assisted treadmill training with body weight support has been used in many research and rehabilitation settings to improve locomotor function in individuals with spinal cord injury (SCI) and stroke [1-3]. It is postulated that treadmill training is a therapeutic paradigm that effectively activates afferent receptors in the lower limb, generating the necessary sensory feedback needed to train central pattern generators in the spinal cord [4,5]. However, the primary limitation with manual-assisted treadmill training with body weight support is that a training session relies on several physical therapists to manually assist the patient's leg movement and training sessions tend to be short because of the physical demands on the therapists.

To improve these limitations of manual-assisted treadmill training, the Lokomat System, robot-assisted gait training device, was developed recently [6]. This robotic gait training system is a computer-controlled, exoskeletal device that is secured to a person's legs while he or she is supported over a motorized treadmill using a counterweight unloading system similar to therapist-assisted treadmill training. The potential advantages of using robotic devices in rehabilitation fields include patient safety, repeatability, unlimited duration of training and hands-free operation by a single therapist. The Lokomat System also has advantages in respect to research purpose for rehabilitation [7,8]: It will become possible to measure different parameters of gait, change loading condition for body weight support and the degree of assistance

Corresponding Author : Yongchul Kim
Department of Rehabilitation Science and Technology
Daegu University
15 Naeri-ri, Jillyang-eup, Gyeongsan-si, 712-714, South Korea
Tel : +82-53-850-4342 / Fax : +82-53-850-4349
E-mail : nakim@daegu.ac.kr

The author would like to acknowledge the assistance and support of Dr. B. D. Schmit, T. H. George and M. Wu.

of leg movements. This will facilitate investigation of the influence of these parameters on the effects of training.

The swing phase of gait can be assisted with functional electrical stimulation (FES) by using the flexion withdrawal reflex [9-11]. The advantage of the flexion reflex is that it provides balanced hip flexion that is difficult to attain with stimulation of individual hip flexors. Therefore, nociceptive flexor reflex has been used for producing or assisting the swing phase of gait cycle in SCI patients through FES system. The combined use of FES and treadmill training (with Lokomat System) is a recent innovation, and it has been shown to increase walking speed in people with SCI [12]. Clinically applied FES systems have been based on both open-loop control through a trial-error approach and closed-loop control to improve the performance of neuroprosthetic devices. However, it is difficult to determine the optimal stimulation pattern that is required to obtain the desired limb motion because the human body consists of a complex multi-joint system having highly nonlinear neuromuscular properties [13].

For several decades, researchers have studied the identification of the input-output properties of electrically stimulated muscle responses [14] and have developed mathematical models for lower limb dynamics in paraplegic patients [15,16]. Unfortunately, a mathematical model of the flexion withdrawal reflex, which can play an important role in producing flexion of lower limb during the swing-phase in simple reciprocal ambulation of paraplegic persons, has not been studied. Therefore, it is important to investigate the relationship between hip conditions (position and angular velocity) and the reflex response elicited by nociceptive electrical stimulation of the foot in order to restore motor function in patients with upper motor neuron lesions. In our previous studies [17,18], we showed that the flexion reflex elicited by nociceptive electrical stimulation of the foot was modulated by the hip conditions (hip position and velocity) in a supine position of subjects with SCI. The flexion reflex might also be influenced by the subject's body weight under upright posture [19].

The aim of this study is to investigate the influence of hip conditions on the flexor reflex elicited by nociceptive electrical stimulation of the skin afferents of the foot in SCI subjects during bodyweight supported treadmill walking with Lokomat System. The joint moment responses of the hip and the EMG (electromyogram) of the tibialis anterior (TA) were measured in order to characterize the flexor reflex response during gait cycle. The influence of hip position and movements was identified using simultaneous electrical stimulation and imposed movements of the leg using a motorized apparatus.

Ⅱ. METHODS

A. Subjects

Five subjects were recruited into this study through the inpatient and outpatient physical therapy clinics of the Rehabilitation Institute of Chicago. The patients included were sensory and motor complete (American Spinal Injury Association (ASIA) Scale A) or sensory and motor incomplete (ASIA C, D) at levels ranging from C5 to C7, as summarized in Table I. Subject consent was obtained and all procedures were conducted in accordance with the Helsinki Declaration of 1975 and approved by the Institutional Review Boards of Northwestern University, Chicago, Illinois.

B. Experimental setup

The primary equipment used in this study was the Lokomat System developed by Hocoma AG (Volkestswil, Switzerland). This system consists of the Lokomat (robotic gait orthosis), body weight support system and a treadmill (Woodway Inc., Waukesha, WI, USA) (Fig. 1). The Lokomat is composed of high quality computer controlled drives with a precision ball screw and a rotatable parallelogram structure that allows the upward and downward movements of the body, but prevents lateral balance problems. Two drives on each leg were moving

Table 1. Clinical features of subjects

Subject	Level of Injury	ASIA Impairment Scale	Age(years)	Time post injury (years)l
A	C5	C	37	4
B	C5	C	27	2
C	C7	A	33	1
D	C5-6	D	42	8
E	C6-7	D	35	13

the hip and knee joints of the orthosis and consequently the legs of the subject. The motors were controlled by a real time system and synchronized with the speed of the treadmill, so that the legs became moved by a normal physiological gait pattern on the moving treadmill. This sensitive system assures a precise match between the speed of the gait orthosis and the treadmill. In this study, speed of gait was set between 2.0 and 2.5 km/h as determined by subject tolerance and comfort. The amount of body-weight support provided was minimized to allow maximum lower-extremity loading without evidence of excessive knee flexion during stance or toe drag during swing.

In order to elicit the flexor reflex during Lokomat treadmill walking, the electrical stimuli by constant current stimulator (Model DS-7A, Digitimer Stimulator, Hertfordshire, UK) were applied at the medial arch of the foot in one of the following phases of the gait cycle: heel contact, foot flat, mid stance, heel off; toe off, initial swing, mid swing, terminal swing (Fig. 2). The electrical stimulus train consisted of a 50mA, 200-Hz pulse train composed of 10 monophasic pulses (each pulse of 1-ms duration).

To measure the reflex activity of monoarticular flexor muscle, surface electrode was applied to lightly abraded, degreased skin over the muscle belly of the TA. The signal was amplified (x 1,000) and filtered (10-500 Hz; Myosystem 1400A, Noraxon USA, Inc, Scottsdale, AZ). The interaction forces exerted by the subject on the Lokomat were also measured to identify reflexive joint moments using six degrees of freedom load cells (JR3 Inc., Woodland, CA, USA) mounted between the Lokomat and the leg cuffs attached to the subject. The signals were sampling at 1,000 Hz using a data acquisition card (PCI-6031E, National Instruments, Austin, TX, USA). Custom Matlab software (Mathworks Inc., Natick, MA, USA) was used for controlling data acquisition and electrical stimulation.

C. Data analysis

TA EMG was band-stop filtered at 57.5-62.5 Hz (8th order Butterworth filter), rectified and smoothed using a low pass filter (5 Hz, 8th order Butterworth filter). The rectified, smoothed TA EMG data were integrated for each of the three trials at each condition to obtain a measure of the EMG area associated with the flexor reflex response. The muscle activation (EMG) pattern for each stride was then time normalized, expressed as a percentage of the total gait cycle (e.g., 0-100%).

The standard leg attachment cuffs that come with the Lokomat were modified to contain 6-degrees of freedom load cells (JR3 Inc., Fig. 3(a)). The calibrated forces and moments were obtained from multiplying the six by six calibration matrix by the six-element output voltage vector (Eq. 1).

$$\begin{bmatrix} A_{11} & A_{12} & A_{13} & A_{14} & A_{15} & A_{16} \\ A_{21} & A_{22} & A_{23} & A_{24} & A_{25} & A_{26} \\ A_{31} & A_{32} & A_{33} & A_{34} & A_{35} & A_{36} \\ A_{41} & A_{42} & A_{43} & A_{44} & A_{45} & A_{46} \\ A_{51} & A_{52} & A_{53} & A_{54} & A_{55} & A_{56} \\ A_{61} & A_{62} & A_{63} & A_{64} & A_{65} & A_{66} \end{bmatrix} \begin{bmatrix} V_{FX} \\ V_{FY} \\ V_{FZ} \\ V_{MX} \\ V_{MY} \\ V_{MZ} \end{bmatrix} = \begin{bmatrix} Load_{FX} \\ Load_{FY} \\ Load_{FZ} \\ Load_{MX} \\ Load_{MY} \\ Load_{MZ} \end{bmatrix} \tag{1}$$

where, [A]=6 × 6 calibration matrix, [V]=output voltage vector and [Load]=calibrated load vector.

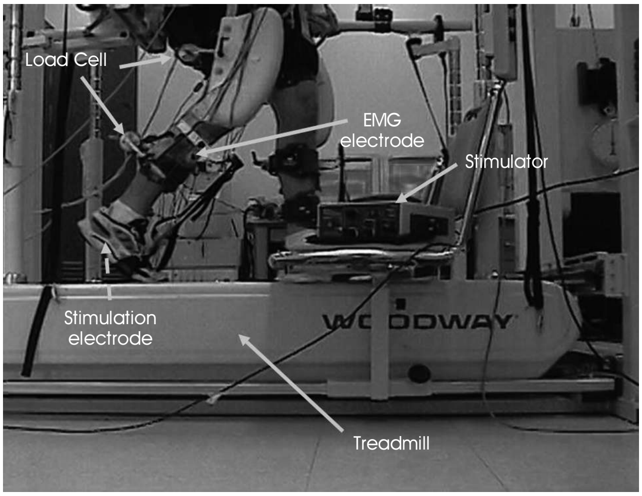


Fig. 1. Experimental setup

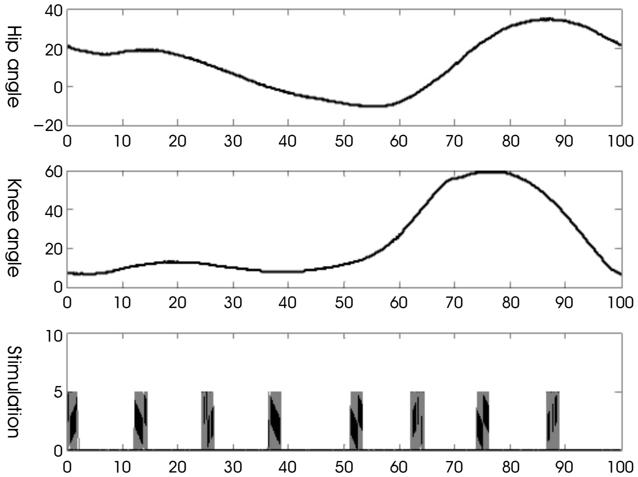


Fig. 2. The eight stimulation phase during the gait cycle

(a) (b)

Fig. 3. Calculation of joint torque from force sensor

In fig. 3(b), the horizontal and vertical distances of load cells from each joint were used for calculating the joint moments of subject (Eq. 2).

$$\begin{bmatrix} M_{hip} \\ M_{knee} \end{bmatrix} = \begin{bmatrix} M_1 \\ M_2 \end{bmatrix} + \begin{bmatrix} H_1 & V_1 & 0 & 0 \\ 0 & 0 & H_2 & V_2 \end{bmatrix} \begin{bmatrix} R_1 \\ R_2 \\ R_3 \\ R_4 \end{bmatrix} \tag{2}$$

where, H, V are horizontal and vertical distances from each joint of subject; R, M are forces and moments from load cell.

The net reflex moments of the hip and knee joints induced by foot stimulation during Lokomat treadmill walking can be determined by subtracting passive hip and knee moments without stimulation.

D. Identification of reflex moment using linear model

In our previous studies [17,18], we postulated that the hip proprioceptors, including the hip joint afferents and hip muscle afferents, converged with the cutaneous afferents from the noxious foot stimulation onto common spinal neuronal pathways to result in a flexion withdrawal reflex that was modulated by hip joint angle and angular velocity. Therefore, we assumed the reflex moment of the hip joint as combination of linear model of hip angle (θ_{hip}) and angular velocity $(\dot{\theta}_{hip})$.

$$M_{hip}^{reflex} = B_0 + B_1\theta_{hip} + B_2\dot{\theta}_{hip} \tag{3}$$

where B_1, B_2 are static and dynamic gains of reflex moment of hip joint, respectively.

Ⅲ. RESULTS

A. Characteristics of TA EMG Response during Lokomat Treadmill Walking

Fig. 4 shows a typical example of TA EMG response in each of the eight stimulation phases during Lokomat treadmill walking. The TA EMG activity was very small in the SCI subjects during Lokomat treadmill walking without stimulation of the foot. In contrast, a distinct flexor reflex after stimulation of the foot appeared in the TA in this subject. Fig. 5 shows the integrated TA EMG of the flexor reflex for all subjects. The TA reflex response was modulated during gait

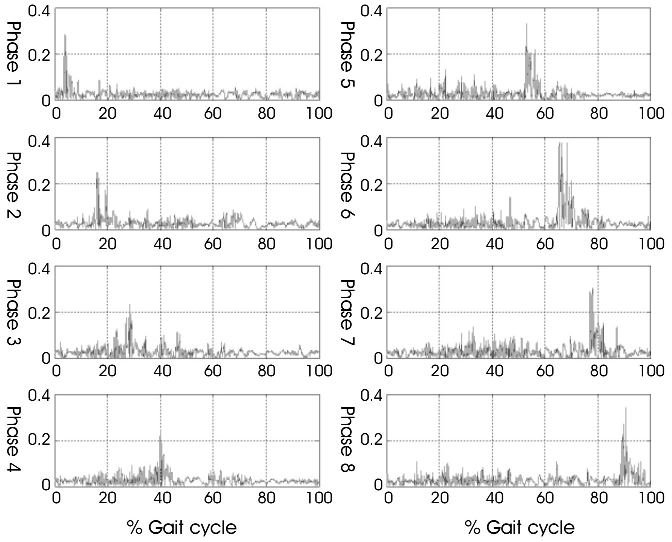


Fig. 4. Typical TA EMG activity of spinal cord injured patients during Lokomat treadmill walking with foot stimulation (subject B)

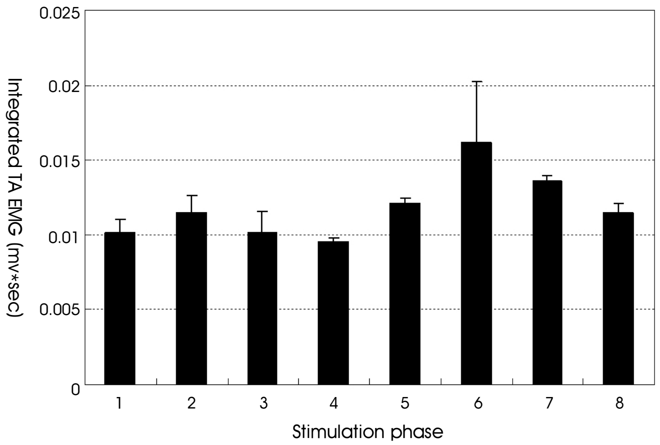
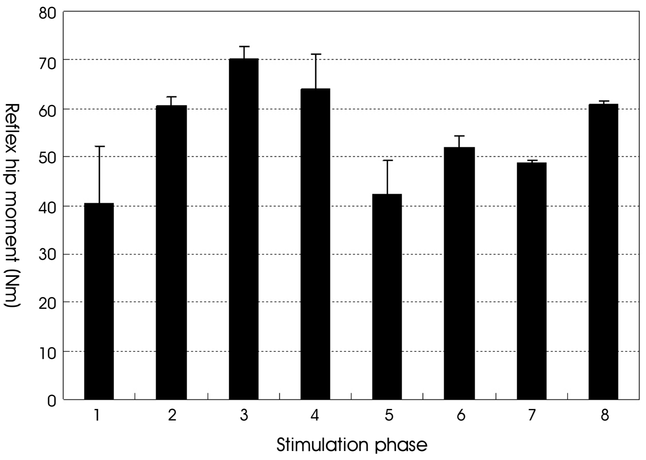


Fig. 5. Integrated TA EMG response induced by stimulation of the foot sole at eight different phases of the gait cycle

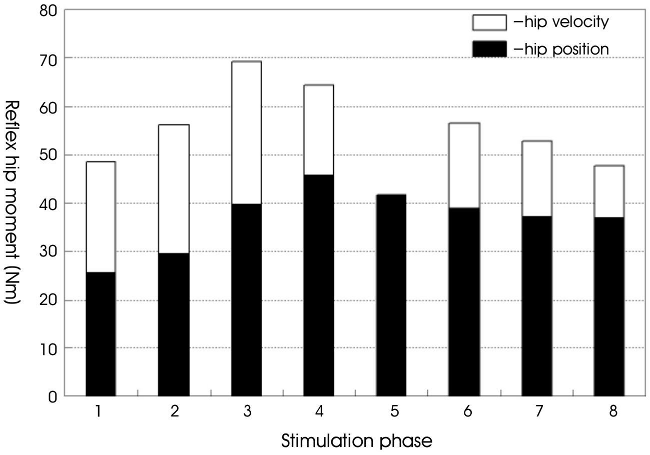
phase and the largest reflex responses were obtained after heel-off and initial swing phase. During swing phase, TA EMG was 40.9% greater for the extended hip position (phase 6), compared with flexed hip position (phase 8). These data support the conclusion that an extended hip posture increases the reflex excitability of the flexor reflex.

B. Characteristics of Reflex Moment during Lokomat Treadmill Walking

Fig. 6 shows a typical reflex moment of the hip joint at stimulation phase 3 and hip moment without stimulation during Lokomat treadmill walking. The net reflex moment recorded after stimulation was subtracted from the response



(a) Measured reflex moment



(b) Static and dynamic contributions of reflex moment

Fig. 7. Reflex hip moments at the hip during gait cycle.

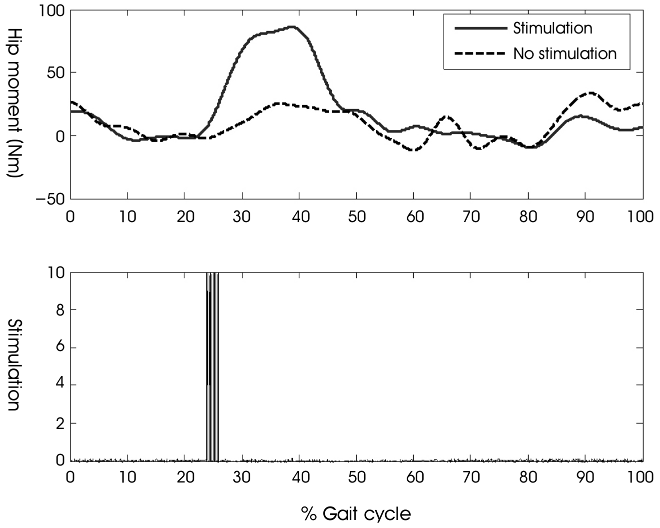


Fig. 6. Raw data of reflex moments at the hip during gait cycle at phase 3

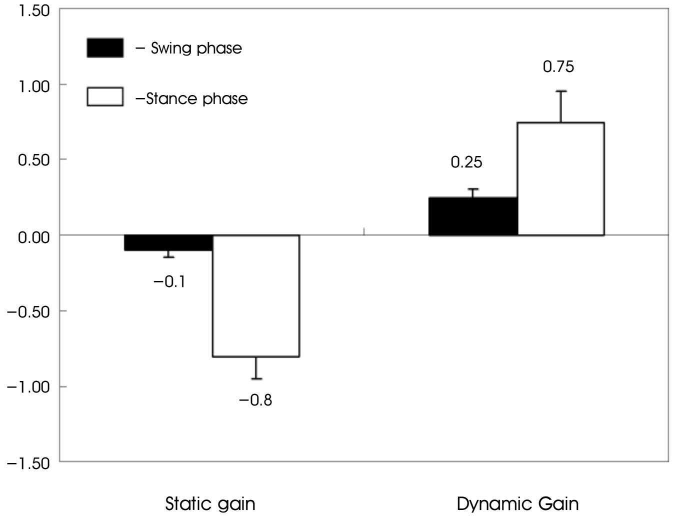


Fig. 8. Mean static and dynamic gains of reflex hip moments using linear model

without stimulation. In Fig. 7 (a), reflex moment of the hip joint was modulated during gait phase and the reflex hip moment observed after heel-contact and during foot-flat was larger than the responses after heel-off and during swing-phase. Fig. 7 (b) shows the contributions of reflex hip moment that is divided into static and dynamic components by assuming the reflex moment as linear function of hip conditions (hip position and hip angular velocity). The mean static and dynamic contributions of the reflex hip moment using linear model are shown in Fig. 8. The mean static gains of reflex hip moments for swing and stance phase are -0.1, -0.8, respectively. The mean dynamic gains of reflex hip moments are 0.25 for swing, 0.75 for stance phase.

Ⅳ. DISCUSSION

Our study showed that the flexion reflex during Lokomat treadmill walking was modulated in a phase-dependent manner.

1) TA EMG response

TA EMG response was modulated during Lokomat treadmill walking. Specifically, TA EMG was greater for the extended hip position (phase 6), compared with flexed hip position (phase 8) during swing phase. Our current observations are consistent with the modulation pattern of the flexion reflex in SCI subjects by hip positions, which is facilitated during hip extension and depressed during hip flexion [17,18]. In this study, flexion reflexes were facilitated at the step cycle phases at which the hip was extended and depressed when the hip was flexed as occurs during the swing-to-stance transition. The nociceptive flexion reflex was also modulated in a phase dependent manner in non-injured subjects [20,21], similar to our observations.

2) Reflex moment response of the hip joint

The hip moment response was probably modulated by a combination of differences in the neural drive and differences in the muscle mechanics involving muscle length and moment arm changes of the hip musculature during Lokomat treadmill walking. For example, the difference between the muscle length of the iliapsoas (a hip flexor) in the flexed and extended positions was approximately 63 and 170% of the optimal fiber length, respectively (based on biomechanical estimates using SIMM, MusculoGraphics, Santa Rosa, CA, USA). Thus, the muscle force output would be modulated by the length-tension properties of the hip muscles. The moment arms of the hip muscles are also changed by hip posture. As a result, the hip joint moments generated in the extended hip position might be higher for the same amount of muscle activation. In addition, a stretch reflex excitation of the hip flexors would also be expected to increase the neural drive to the hip flexors in the extended hip posture, further enhancing the neural excitation of these muscles. Although the dependence of flexor reflexes on hip position may reflect an effect of hip flexor afferents on locomotion circuits, as observed in the cat, a central drive from the locomotor generator is also likely to play a unique role in the modulation of flexor reflexes during walking. During walking in humans, the flexor reflex pattern evoked by a painful cutaneous stimulus is modulated by the step cycle [20,22].

Ⅴ. CONCLUSION

We concluded that static and dynamic components from hip proprioceptors enhanced the flexor reflex response during Lokomat treadmill walking. The input-output model of withdrawal reflex can be applied to the design and control strategies of neuroprosthetic devices with functional electrical stimulation for spinal cord injured patients.

REFERENCES

[1] M. Visintin and H. Barbeau, “The effects of body weight support on the locomotor pattern of spastic paretic patients,” *Can. J. Neurol. Sci.*, vol. 16, no. 3, pp.315-325, 1989.

[2] H. Barbeau, K. Norman, J. Fung, M. Visintin and M. Ladouceur, “Does neurorehabilitation play a role in the recovery of walking in neurological populations? ,” *Ann. NY Acad. Sci.*, vol. 860, pp. 377-392, 1998.

[3] H. Barbeau, and J. Fung, “The role of rehabilitation in the recovery of walking in the neurological population,” *Curr. Opin. Neurol.*, vol. 14, no. 6, pp. 735-740, 2001.

[4] A. Wernig, S. Muller, A. Nanassy and E. Cagol, “Laufband therapy based on “rules of spinal locomotion” is effective in spinal cord injured persons,” *Eur. J. Neurosci.*, vol. 7, no. 4, pp. 823-829, 1995.

[5] M. Visintin, H. Barbeau, N. Korner-Bitensky and N. E. Mayo, “A new approach to retrain gait in stroke patients through body weight support and treadmill stimulation,” *Stroke*, vol. 29, no. 6, pp. 1122-1128, 1998.

[6] G. Colombo, M. Joerg, R. Schreier and V. Dietz, “Treadmill training of paraplegic patients using a robotic orthosis,” *J. Rehabil. Res. Dev.*, vol. 37, no. 6, pp. 693-700, 2000.

[7] V. Dietz, R. Muller and G. Colombo, “Locomotor activity in spinal man: significance of afferent input from joint and load receptors,” *Brain*, vol. 125, no. 12, pp. 2626-2634, 2002.

[8] R. Muller and V. Dietz, “Neuronal function in chronic spinal cord injury: Divergence between locomotor and flexion- and H-reflex activity,” *Clin. Neurophysiolo*., vol. 117, no. 7, pp.1499-1507, 2006.

[9] W. T. Liberson, H.J. Holmquest, D. Scot and M. Dow, “Functional electrotherapy: stimulation of the peroneal nerve synchronized with the swing phase of the gait of hemiplegic patients,” *Arch. Phys. Med. Rehabil*., vol. 42, pp.101-105, 1961.

[10] K.H. Lee and R. Johnston, “Electrically induced flexion reflex in gait training of hemiplegic patients,” *Arch. Phys. Med. Rehabil*., vol. 57, no. 1, pp.311-314, 1976.

[11] T. Bajd, A. Kralj, R. Turk, H. Benko and J. Sega, “Use of functional electrical stimulation in the rehabilitation of patients with incomplete spinal cord injuries,” *J. Biomed. Eng*., vol. 11, no. 2, pp.96-102, 1989.

[12] E.C. Field-Fote, “Combined use of body weight support, functional electrical stimulation, and treadmill training to improve walking ability in individuals with chronic incomplete spinal cord injury,” *Arch. Phys. Med. Rehabil*., vol. 82, no. 6, pp. 818-824, 2001.

[13] R. Riener, “Model-based development of neuroprostheses for paraplegic patients,” *Philos. Trans. R. Soc. Lond. B. Biol. Sci*., vol. 354, pp. 877-894, 1999.

[14] M. Ferrarin and A. Pedotti, “The relationship between electrical stimulus and joint torque: a dynamic model,” *IEEE Trans. Rehabil. Eng*., vol. 8, no. 3, pp. 342-352, 2000.

[15] R. Riener and T. Fuhr, “Patient-driven control of FES-supported standing up: a simulation study,” *IEEE Trans. Rehabil. Eng*., vol. 6, no. 2, pp.113-124, 1998.

[16] D.B. Popovic, R.B. Stein, M.N. Oguztoreli, M. Lebiedowska, S. Jonic, “Optimal control of walking with functional electrical stimulation: a computer simulation study,” *IEEE Trans. Rehabil. Eng*., vol. 7, no. 1, pp. 69-79, 1999.

[17] Y. Kim, B.D. Schmit and Y. Youm, “Stimulation parameter optimization for functional electrical stimulation assisted gait in human spinal cord injury using response surface methodology,” *Clin. Biomech*., vol. 21, no. 5, pp. 485-494, 2006.

[18] Y. Kim, Y. Youm, M. Wu, B.D. Schmit, “Modulation of flexor reflexes by static and dynamic hip proprioceptors in chronic human spinal cord injury,” *J. Clin. Neurosci*., vol. 14, no. 11, pp.1078-1088, 2007.

[19] A. Rossi and B. Decchi, “Flexibility of lower limb reflex responses to painful cutaneous stimulation in standing humans: evidence of load-dependent modulation,” *J. Physiol*., vol. 481, no. 2, pp. 521-532, 1994.

[20] P. Crenna and C. Frigo, “Evidence of phase-dependent nociceptive reflexes during locomotion in human,” *Exp. Neurol*., vol. 85, no. 2, pp.336-345, 1984.

[21] E.G. Spaich, L. Arendt-Nielsen and O.K. Andersen, “Modulation of lower limb withdrawal reflexes during gait: A topological study,” *J. Neurophysiol*., vol. 91, no. 1, pp.258-266, 2004.

[22] M. Belanger and A.E. Patla, “Corrective responses to perturbation applied during walking in humans,” *Neurosci. Lett*., vol. 49, no. 3, pp.291-295, 1984.