The Role of Ankle Plantar Flexors in Walking Performance during Closed-Loop Control of Walker-Assisted FES-Activated Paraplegic Walking

V. Nekoukar, and A. Erfanian

***Abstract*— In this paper, we investigate the role of the ankle plantar flexors in walking performance and handle reaction force (HRF) during closed-loop control of walker-assisted FES-activated walking. For this purpose, we employed a decentralized control scheme for control of FES-activated walking. The control scheme has a modular structure. For each muscle-joint complex (i.e., flexor or extensor), an independent local controller is designed and the dynamic interactions between the joints are considered as the external disturbances for the local controllers. Activation and deactivation times of the muscles as well as the stimulation intensity are automatically determined by the controllers. The results of experiments on two paraplegic subjects show that the 10-channel closed-loop control of walking including plantar flexors can improve walking performance and reduce HRF when compared to the 8-channel closed-loop FES system (excluded plantar flexors).**

I. INTRODUCTION

For over three decades, many research groups have shown that limited crutch- or walker-assisted walking can be restored in subjects with spinal cord injuries by means of functional neuromuscular stimulation (FNS) systems [1].

One important issue in restoring motor function using FES is to determine the stimulation patterns for a specific number of muscles. To address these problems, Popović et al. [2] employed the musculoskeletal model developed in [3] and used optimal control methods to determine the muscle activation patterns and then applied to paraplegic subjects. In previous work [4], we introduced a two-dimensional (2-D), 11-degree-of-freedom (DOF) dynamic model of walker-assisted FES-activating paraplegic walking. The model consists of feet, shanks, thighs, trunk, arms, forearms, and a walker. The goal of the simulation was to determine the muscle stimulation profiles. The performance criterion was the minimization of the sum of the squares of tracking errors from desired trajectories including a penalty function on the total muscle effort acting around the hip, knee and ankle joints and the handle reaction force (HRF) (the reaction force at the hand-handle interface). The results of experiments on paraplegic subjects indicated that the proposed methodology can improve walking performance in terms of walking pattern, reduced HRF, and increased walking speed. However, optimal control is not suitable for online control of walking. To implement the optimal control for walking, an accurate model of multilink neuromusculoskeletal system is required. Due to time-varying property of neuromusculoskeletal system, subject-to-subject variations, unmodeled dynamics, model uncertainties, and disturbances, determining the muscle activation patterns using optimal control is much less efficient for online control of walking in paraplegic subjects.

A useful and powerful control scheme to deal with the uncertainties, nonlinearities, and bounded external disturbances is the sliding mode control (SMC) [5]-[7]. We have already developed an adaptive fuzzy terminal sliding mode (AFTSM) for control of a class of nonlinear uncertain systems without relying on a priori knowledge about the dynamics of the system to be controlled [8]. The method was successfully applied on paraplegic subjects to control walker-assisted FES-activated walking [9]. In the current study, we employed the AFTSM control scheme proposed in [9] for closed-loop control of FES-activated walking.

In addition, the quality and stability of the resulting gait produced by walking stimulation patterns depends on the number of stimulation channels and restoring the function of individual muscles contributed to the acceleration/deceleration of the body. In the current study, we investigated the role of plantar flexor in quality of generated gait patterns in paraplegic subjects and upper body effort during closed-loop control of FES-activated walking.

II. METHOD

The proposed control strategy for walking has a modular structure. For each muscle group (i.e., extensor or flexor), an independent controller was designed; AFTSM for the hip flexor and extensor and the knee extensor and fuzzy logic control for the ankle extensor and flexor. The structure of the control framework used for control of paraplegic walking is shown in Fig. 1.

To implement the proposed TSM controller, each muscle-joint complex should be presented in the canonical form as

$$\theta(t) = f(\theta,\dot{\theta},t) + b(\theta,\dot{\theta},t)\cdot u(t) + d(t) \qquad (1)$$

where $\theta(t)$ is the joint angle, $d(t)$ is the lumped disturbance, and $u(t)$ is the control input (i.e. stimulation signal). $f(\theta,\dot{\theta},t)$ and $b(\theta,\dot{\theta},t)$ are unknown continuous functions but are estimated as known nominal dynamics $\hat{f}(\theta,\dot{\theta},t)$ and $\hat{b}(\theta,\dot{\theta},t)$, respectively, with the bounded estimation errors.

The objective of the control framework is to determine the muscle stimulation patterns online to force the joint angle to track a desired trajectory in the presence of model uncertainties and external disturbances. It can be proved that by choosing the following control law, the estimation errors converge to the zero asymptotically, and tracking error and its first derivative converges to the neighborhood of zero in finite time [8].

$$u = u_{eq} + u_c \qquad (2)$$

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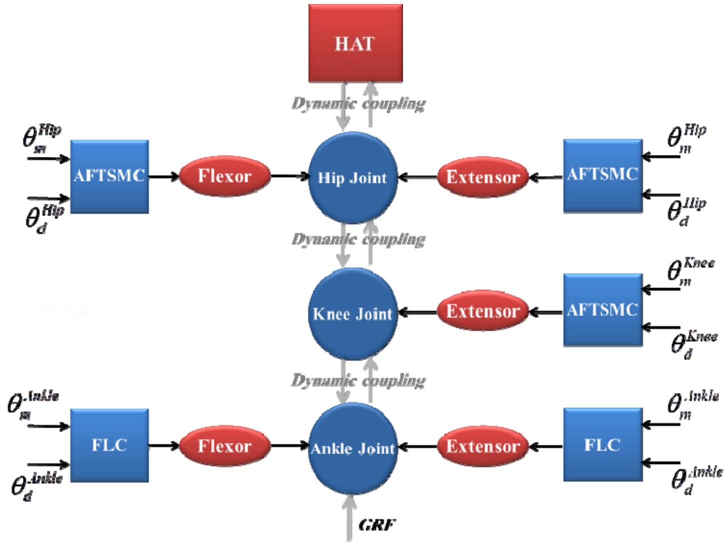

Figure1. Block diagram of the proposed decentralized AFTSM control system for control of walking.

$$u_{eq} = \hat{b}^{-1}(\theta,\dot{\theta},t)\Big(-\hat{f}(\theta,\dot{\theta},t) - d(t) + \ddot{\theta}_d(t) + \beta^{-1}\gamma^{-1} sig(\dot{e})^{2-\gamma} + k_1 s + k_2 sig(s)^p\Big) \tag{3}$$

$$u_c = \bar{\varepsilon}_f + \bar{\varepsilon}_b\left|u_{eq}\right| + \left|u_0\right| \tag{4}$$

$$u_0 = \hat{b}^{-1}(\theta,\dot{\theta},t)\Big(-\hat{f}(\theta,\dot{\theta},t) + \ddot{\theta}_d(t) + \beta^{-1}\gamma^{-1} sig(\dot{e})^{2-\gamma} + k_1 s + k_2 sig(s)^p\Big) \tag{5}$$

$$s(t) = e(t) + \beta\left|\dot{e}(t)\right|^{\gamma} sign(\dot{e}(t)) = 0 \tag{6}$$

where $e(t) = \theta_d(t) - \theta(t)$ is the tracking error, $\theta_d(t)$ is the desired trajectory, $\beta > 0$, $1 < \gamma < 2$, $k_1 > 0$, $k_2 > 0$, $0 < p < 1$, s is the terminal sliding variable, $sig(s)^p = |s|^p sign(s)$ and $\bar{\varepsilon}$ is the upper bound of the estimation error. The details can be found in [8].

The nonlinear functions $f(\theta,\dot{\theta},t)$ and $b(\theta,\dot{\theta},t)$ are approximated using fuzzy logic system as follows

$$\hat{f}(\theta,\dot{\theta},\nu_f) = \nu_f^T \psi_f(\theta,\dot{\theta}) \tag{7}$$

$$\hat{b}(\theta,\dot{\theta},\nu_b) = \nu_b^T \psi_b(\theta,\dot{\theta}) \tag{8}$$

where ν^T is the adjustable parameters and ψ is the fuzzy basis vector [8]. The parameters are adjusted online with the following adaptation laws

$$\dot{\nu}_f = -\eta_f \beta\gamma\left|\dot{e}\right|^{\gamma-1} \psi_f(x,\dot{x}) s \tag{9}$$

$$\dot{\nu}_b = -\eta_b \beta\gamma\left|\dot{e}\right|^{\gamma-1} \psi_b(x,\dot{x}) s u_{eq} \tag{10}$$

where $\eta_f > 0, \eta_b > 0$. Further details was given in [8].

The parameters of the AFTSM controllers were chosen heuristically to achieve the best controller performance during the first session of the experiment and then used for subsequent experiments on different sessions.

III. Experimental Procedure

The experiments were conducted on two thoracic-level complete spinal cord injury subjects with injury at T7 (RR: 35 years old, H = 174 cm, and M = 76 kg) and T12 (MS: 23 years old, H = 191 cm and M = 65 kg). The paraplegic subjects are active participants in a rehabilitation research program involving daily electrically stimulated exercise of their lower limbs. The 20 reusable self-adhesive skin electrodes were placed as follows: two pairs of electrodes (1 right, 1 left) over the quadriceps muscle, two pairs over the common peroneal nerve, two pairs over the gluteus maximus/minimus muscle, two pairs over the soleus and gastrocnemius muscles, and two pairs over the iliacus muscle. The iliacus muscles were stimulated to assist flexing the thigh and pulling the knee upward during swing phase.

A computer-based FES system was used to stimulate the muscles using pulse width (PW) modulation (from 0 to 700 µsec) with balanced bipolar stimulation pulses at constant pulse amplitude and a constant frequency (25 Hz). The experimental sessions for each subject were conducted once a week. Each session consisted of six trials with an inter-trial resting interval of at least 5 min. Different walking trials were conducted, each with approximately 8 strides.

The joint angles were measured with motion tracker system MTx (Xsens Technologies, Netherlands). The ground reaction force (GRF) was measured with the pedar-x system (Novel, Germany). Paraplegic subjects were allowed moving the walker without lifting. The HRF was measured by a 3-component piezoelectric force sensor (9602, Kistler, Switzerland) placed under the front-wheeled walker handle.

IV. Results and Discussion

The walking performance was assessed using average and standard deviations of root-mean-square (RMS) error and normalized RMS (NRMS) error for trajectory tracking. Typical results of the closed-loop control of paraplegic walking for subject RR are shown in Fig. 2.

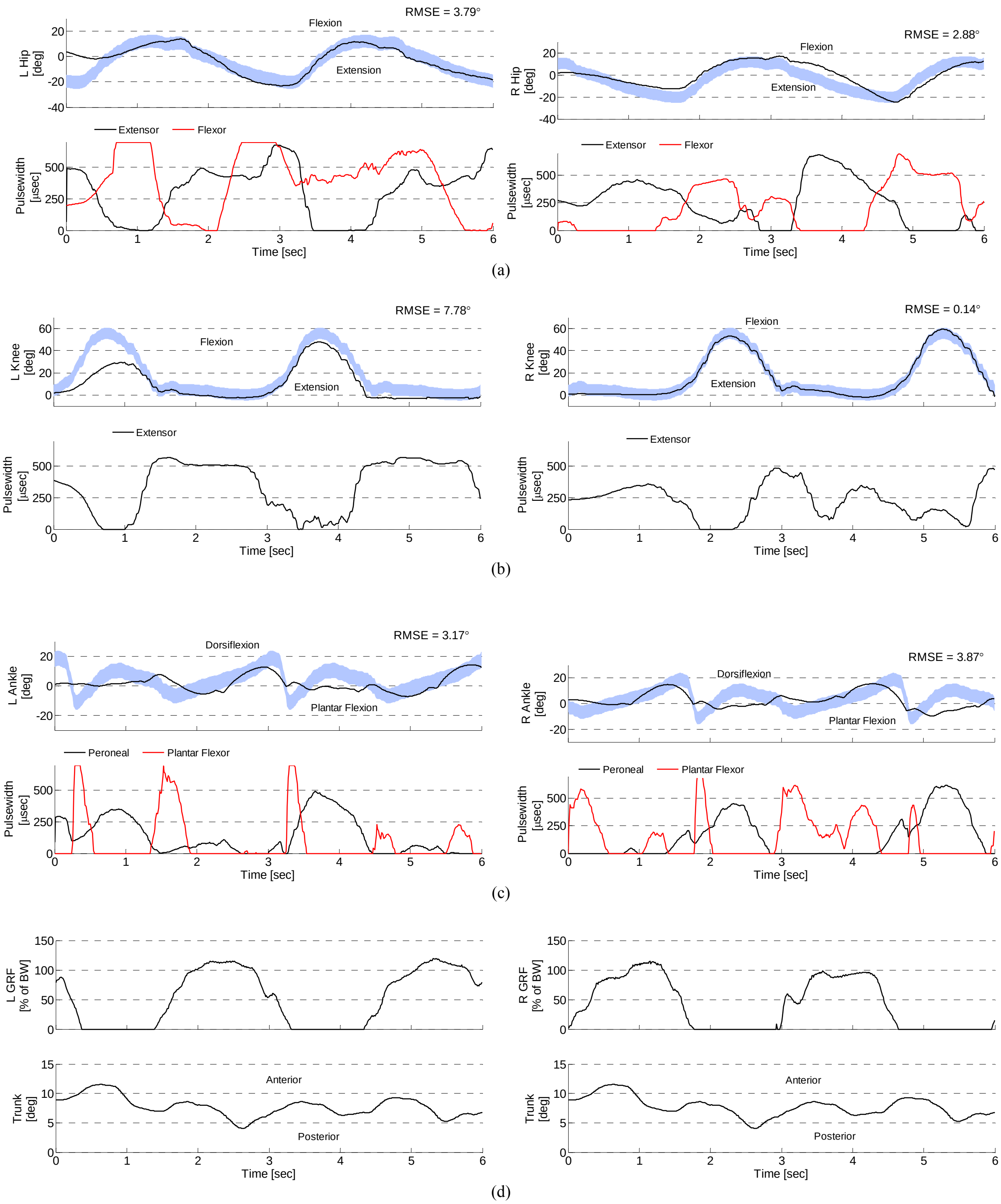

Figure 2. Typical results of the 10-channel closed-loop walker-assisted FES-activated paraplegic walking (subject RR). (a)-(c) The upper graph shows the boundary of the desired trajectory (blue shading) and the measured joint angle (solid line). The lower plots show the stimulation PW. (d) The upper graph shows the measured GRF, the lower graph the trunk joint angle. The left column is for the left leg and vice versa.

The RMS tracking errors was 2.88° (6.77%) for the hip joint, 0.14° (0.22%) for the knee, and 3.87 ° (9.70%) for the ankle joint. It was observed that a good tracking performance was obtained for the hip and the knee joints. During loading response, the ankle flexor was activated (heel rocker). This is in agreement with what has been reported in studies of normal gait [10]. Activation of the pretibial muscles causes the rate of foot drop to decelerate and causes the tibia to draw forward. During mid-stance and terminal stance, the plantar flexor was also observed to be activated. Activation of the plantar flexor is needed to decelerate the effects of the passive dorsiflexion torque and stabilize the dorsiflexion ankle during mid-stance and terminal stance.

The control objective of the ankle joint was maintaining the foot clearance during the swing phase of walking. It is interesting to note that although the knee flexor muscles were not stimulated, the knee flexed during the swing phase. The reason is that the controller stimulated the common peroneal nerve to produce the swing phase. Stimulation of the peroneal nerve produced a withdrawal reflex that induced simultaneous excitation of the hip, knee and ankle flexors. Fig. 2 (c) shows that the plantar flexion was not produced during swing phase properly. This is because the plantar flexor controller was switched off and the peroneal controller was switched on to flex the knee and ankle joints. Nevertheless it was observed that the foot clearance was perfectly achieved during the swing. The GRF (Fig. 2(d)) shows that the body weight of the subject was transferred to the feet. The vertical force exerted by the ground showed the familiar double-hump pattern**.**

Table I summarizes the average RMS errors over all trials for three experiment days using the proposed closed-loop control. The averages of tracking error were 10.4%, 4.4%, and 5.5%, for the ankle, knee, and hip, respectively. Average of the vertical HRF during closed-loop control of walking using 10-channel and 8-channel FES systems are summarized in Table II. The results showed that the average of the vertical HRF were 26.36% and 33.29% of the body weight using the 10-channel and 8-channel closed-loop FES systems, respectively. Using10-channel stimulation, the vertical HRF decreased approximately 20% with respect to the 8-channel stimulation.

V. Conclusion

In this paper, we investigated the role of plantar flexors during closed-loop control of FES-activated paraplegic walking. In normal walking, during loading response, the plantar flexor is activated to reduce the heel rocker effect. Throughout mid-stance an ever-increasing dorsiflexion torque is created in response to momentum from the swing limb and the forward fall of body weight. The ankle plantar flexor muscles react to restrain the passive dorsiflexion that follows forefoot floor contact. During terminal stance, the plantar flexor is also activated to decelerate the effects of the passive dorsiflexion torque and stabilize the dorsiflexion ankle [10]. Therefore, restoring the function of plantar flexor during FRS-activated walking can improve the walking performance. The results of this study show that walking performance improved when 10-channel closed-loop FES system is used to control walking (included plantar flexor) compared to 8-channel closed-loop FES system (excluded plantar flexor).

TABLE I. Average Root-Mean-Square Tracking Error (±Standard Deviation) Obtained Using the 8-Channel and 10-Channel Closed-Loop FES System for Two Paraplegic Subjects

Subject			Day 1	Day 2	Day 3	Mean
RR	Ankle	8-channel	5.2°±0.7°	5.0°±0.9°	5.4°±0.9°	5.2° (13.0%)
		10-channel	3.7°±0.8°	3.9°±0.6°	4.0°±0.5°	3.9° (9.8%)
	Knee	8-channel	3.5°±0.6°	3.2°±0.3°	2.9°±0.5°	3.2° (4.9%)
		10-channel	3.1°±0.5°	2.8°±0.4°	2.6°±0.4°	2.8° (4.3%)
	Hip	8-channel	2.4°±0.3°	2.6°±0.4°	2.1°±0.2°	2.4° (5.7%)
		10-channel	2.2°±0.3°	2.3°±0.3°	2.0°±0.2°	2.2° (5.2%)
MS	Ankle	8-channel	6.3°±0.6°	6.9°±0.9°	7.1°±0.9°	6.8° (17.0%)
		10-channel	4.7°±0.6°	4.4°±0.8°	4.2°±0.7°	4.4° (11.0%)
	Knee	8-channel	3.3°±0.4°	3.7°±0.5°	3.2°±0.5°	3.4° (5.2%)
		10-channel	2.9°±0.4°	3.1°±0.3°	2.8°±0.3°	2.9° (4.4%)
	Hip	8-channel	2.6°±0.5°	2.7°±0.3°	2.7°±0.3°	2.7° (6.4%)
		10-channel	2.5°±0.4°	2.4°±0.4°	2.5°±0.3°	2.5° (5.9%)

TABLE II. Mean Values (±Standard Deviation) of the Vertical HRF (% Of Body Weight) Using the 8-Channel and 10-Channel Closed-Loop FES Systems for Two Paraplegic Subjects

Subject	8-channel closed-loop FES System (% of BW)	10-channel closed-loop FES System (% of BW)
RR	32.45 ± 1.68	24.04 ± 1.55
MS	34.12 ± 1.89	28.67 ± 1.73

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