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A decentralized adaptive fuzzy robust strategy for control of upright standing posture in paraplegia using functional electrical stimulation

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ABSTRACT

In this paper, we present a novel decentralized robust methodology for control of quiet upright posture during arm-free paraplegic standing using functional electrical stimulation (FES). Each muscle–joint complex is considered as a subsystem and individual controllers are designed for each one. Each controller operates solely on its associated subsystem, with no exchange of information between them, and the interaction between the subsystems are taken as external disturbances. In order to achieve robustness with respect to external disturbances, unmodeled dynamics, model uncertainty and time-varying properties of muscle–joint dynamics, a robust control framework is proposed. The method is based on the synergistic combination of an adaptive nonlinear compensator with sliding mode control (SMC). Fuzzy logic system is used to represent unknown system dynamics for implementing SMC and an adaptive updating law is designed for online estimating the system parameters such that the global stability and asymptotic convergence to zero of tracking errors is guaranteed. The proposed controller requires no prior knowledge about the dynamics of system to be controlled and no offline learning phase. The results of experiments on three paraplegic subjects show that the proposed control strategy is able to maintain the vertical standing posture using only FES control of ankle dorsiflexion and plantarflexion without using upper limbs for support and to compensate the effect of external disturbances and muscle fatigue.

1. Introduction

For over three decades, many research groups have shown that arm-supported standing can be restored in subjects with spinal cord injuries by means of functional electrical stimulation (FES) systems [1–5]. During recent years, much attention has been directed to arm-free paraplegic standing via FES [6–14]. Hunt et al. [8], Munih et al. [9], Holderbaum et al. [10], and Gollee et al. [11] have proposed a nested-loop control structure in which the total ankle moment was regulated by an inner loop controller and the inclination angle by an outer loop controller. For the inner controller the plant was the muscle, while for the outer the plant was the combination of inner-loop dynamics and the biomechanical dynamics. In their experimental setup, all joints above the ankle were locked using a special body brace, allowing isolating the effects of the artificial control system from the remaining motor control actions of the intact upper body. The control design approach involved first designing the inner loop controller based upon fixed-parameters muscle models empirically derived under *isometric conditions*. The muscle model described by a linear-dynamic structure in [10,11] and a Hammerstein structure [8,9] consisted of a static recruitment nonlinearity followed by a linear transfer function. In [8,9], the recruitment nonlinearity was cancelled using the inverse of the estimated nonlinearity. They used linear quadratic Gaussian (LQG) design approach for both the inner and the outer loop controllers in [8,9], standard pole-placement design for both controllers in [11], and pole-placement design for the inner and H_∞ optimal control for the outer loop in [10].

In all the above-mentioned unsupported standing strategies, the subject was modeled as a single-link inverted pendulum. The only degree of freedom was the ankle joint while all joints above the ankles were braced. The muscle activation was limited to the ankle plantarflexor group and an anterior posture was maintained during feedback control of unsupported standing.

Control strategies for unsupported paraplegic standing, utilizing the voluntary and reflex activity of the non-paralyzed upper body and closed-loop FES control of ankle plantar and dorsal flexors have been also proposed by Matjačić et al. [12] and Mihelj and Munih [13]. The subject was modeled as a double two-link inverted pendulum while the knees and hips were maintained in the extended position by long leg braces, the upper body was free to move voluntary, and the paralyzed ankle muscles were controlled via FES to achieve a desired level of ankle stiffness. Matjačić et al. [12] used the method proposed in [11] with FES-controlled ankle stiffness and voluntary motor control action from the upper body, for control of arm-free standing in a complete T5 paraplegic

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subject. They demonstrated that the subject not only was able to maintain upright posture during unperturbed standing for periods of 1 min, but that he has also managed to progressively decrease the amplitude of the sway inclination angle, indicating the effect of the voluntary upper motor control. The result showed that the amplitude of the sway angle was more than $\pm 7^\circ$ at the start of trial experiment. It should be noted that these joint displacements generate CoP displacements which are within undesirable preference zone discussed in [15], i.e., the area where the subject is forced to change posture in order to maintain balance.

To produce a more robust two-link arm-free standing, Mihelj and Munih [13] performed experiments on a complete T6 paraplegic subject under free disturbance while an *anterior posture* was maintained by adding offset value to the ankle torque operating point. They showed that by setting the torque operating point to -40 N m, the stance position leaded to anterior posture that required sustained plantarflexor muscle effort.

The current study focuses on developing a robust control scheme for posture control during upright quiet standing without involving the upper body in paraplegic subjects using FES control of agonist–antagonist muscles.

2. Problem statement

Despite the exciting approaches to FES-assisted free-support standing discussed above [8–14], control of the upright posture (i.e., erect position) during quiet standing using only ankle strategy, still remains as an open problem. Upright postural control during unperturbed stance has recently been achieved in a paraplegic subject by utilizing his voluntary motor actions from the intact upper body [12,13]. However, it should be noted that involving the upper body in posture stabilization causes the arms and hands not to be free to concentrate on some functional tasks (one can imagine the performance of a rope walker to stabilize the standing posture). Moreover, experimental observations by others [16–20] have shown that, for small disturbances, human subjects prefer to move primary about the ankles to maintain erect stance and the ankle plantarflexors/dorsiflexors alone act to control the posture. In more perturbed situations or when the ankle muscles cannot act, the hip strategy must be employed.

The main motivation for this work is to develop a robust control strategy for control of upright posture during quiet standing in paraplegic subjects using only ankle strategy by coordinately activating the dorsiflexor and plantarflexor muscles.

All the earlier mentioned works used control schemes (i.e., pole placement, linear quadratic Gaussian, linear quadratic regulator, and H_∞ optimal control design) [8–13] which concern with linear time-invariant systems. Pole placement which is an important control design method for linear time-invariant systems is strongly dependent on the availability of an accurate linear model of the system. Also, the underlying assumptions of the traditional linear quadratic Gaussian control are that the plant has a known linear description, and that the exogenous noises and disturbances impinging on the feedback system are stochastic in nature, but have known statistical properties. Moreover, LQG optimality does not automatically guarantee the stability robustness and must be analyzed a posteriori [21]. Mihelj and Munih [13], who used a linear quadratic controller for control of standing posture, modeled the body as a linearized double inverted pendulum dynamics and plantarflexor and dorsiflexor muscle groups as a set of second order linear models. However, it should be noted that during quiet standing, joint angles' displacements are not small enough to assume a system's linearity [22]. When the required operation range is large, a linear controller is likely to perform very poorly or to be unstable, since the system nonlinearities cannot be properly compensated for [23].

A useful and powerful control scheme to deal with the uncertainties, nonlinearities, and bounded external disturbances is the sliding mode control (SMC) [23]. In robust control designs, a fixed control law based on *a priori* information on the uncertainties is designed to compensate for their effects, and *exponential* convergence of the tracking error to a (small) ball centered at the origin is obtained. Robust control has some advantages over the adaptive control, such as its ability to deal with disturbances and quickly varying parameters [23]. Nevertheless, the SMC suffers from the high frequency oscillations in the control input, which is called "chattering" [24]. In order to limit the chattering phenomena and to preserve the main advantages of the original SMC, we have already proposed a robust control strategy which is based on synergistic combination of adaptive nonlinear compensator with SMC [25,26]. However, the methods requires offline identification of the muscle–joint dynamics, which implies a fundamental limit on its application to inherently unstable (e.g., unsupported standing) and time-varying system.

To overcome this problem, we extend previous work and design an adaptive updating law for online identification of the muscle–joint dynamics without requiring any offline identification or offline calibration during different experiment sessions.

3. Controller design

In order to limit the chattering phenomena and to preserve the main advantages of the original SMC, we design an adaptive fuzzy robust control (AFRC), by combination of an adaptive nonlinear compensator [25,26] with an adaptive fuzzy SMC (AFSMC). To implement the SMC (see Appendix A for details), the nonlinear system being controlled should be first presented in a standard canonical form. We use standard fuzzy systems to approximate the nonlinear function of the canonical model. Then, an adaptive updating law is designed for online estimating the model parameters in such a way that the global asymptotic stability of the control system can be guaranteed.

Over the past decade, there has been much research on the design of fuzzy sliding-mode controller based on an adaptive updating law for online estimating the model parameters [27–31]. However, all these works replaced the discontinuous term $k \cdot \mathrm{sign}(s)$ in the sliding mode control law with a continuous term to reduce the chattering. This violates the η-reachability condition which insures finite-time *convergence* to sliding surface [25–32]. In contrast, we use an AFSMC with η-reachability condition ((A.7), Appendix A), in combination with an adaptive nonlinear compensator [25,26] to reduce the chattering.

3.1. Structure of AFRC

The configuration of the proposed control strategy is similar to fuzzy-based SMC proposed in [25] and is schematically depicted in Fig. 1, where u_1 is the output of an adaptive fuzzy SMC (AFSMC) (defined in Section 3.2), u_2 is the output of an adaptive nonlinear compensator as the auxiliary control input (details about u_2 can be

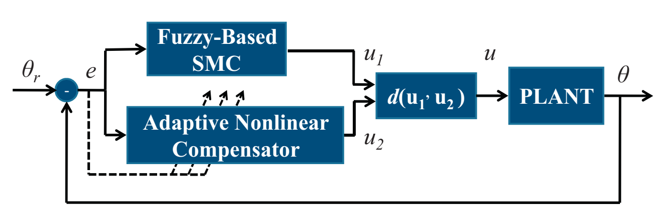

Fig. 1. Structure of the proposed adaptive fuzzy robust control (AFRC).

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found in [25,26]), and $u = d(u_1, u_2)$ is a function of u_1 and u_2 defined by

$$u = d(u_1, u_2) = \begin{cases} u_1 & \text{if } \ |s(e)| > \phi \\ \alpha(e)u_1 + (1-\alpha(e))u_2 & \text{if } \ |s(e)| \le \phi \end{cases} \quad (1)$$

where $s(e)$ is a scalar function described in (A.5) (Appendix A), ϕ is the boundary layer thicknesses, and $\alpha(e)$ is a function of error and is adapted by $\alpha(e) = |s(e)|/\phi$. The original SMC structure is retained in the proposed scheme, but the role of adaptive nonlinear compensator becomes more significant as the state trajectory is nearing the sliding surface.

3.2. Adaptive fuzzy sliding mode control

To implement the SMC (see Appendix A for details), the nonlinear function $f(\mathbf{x},t)$ and the control gain $g(\mathbf{x},t)$ in (A.1) (Appendix A) should be estimated. In this work, we use the fuzzy logic system to approximate the nonlinear function $f(\mathbf{x},t)$. The fuzzy system uses the fuzzy IF–THEN rules to perform a mapping from an input vector $\mathbf{x} = [x_1, x_2, \ldots, x_n]^T \in \Re^n$ to an output $\hat{f}(\mathbf{x}, t) \in \Re$. Using the product inference, singleton fuzzifier, and center-average defuzzifier, the output of the fuzzy system is given as

$$\hat{f}(\mathbf{x}, t) = \hat{\vartheta}_t^T \psi(\mathbf{x}) \quad (2)$$

where $\hat{\vartheta}_t$ is an adjustable parameter vector, and ψ is a fuzzy basis vector (see details in [25]). To approximate the nonlinear function (2), the parameter vector $\hat{\vartheta}$ need to be estimated. In previous work [25], we used the standard recursive least-squares (RLS) algorithm to estimate the parameters using the data obtained during an experiment session. In this work, an adaptive law is designed based on the Lyapunov stability theory for online estimating the parameters ϑ, such that the tracking error converges to zero asymptotically and stability of control system can be guaranteed.

A fuzzy system can be also used to approximate $g(\mathbf{x},t)$. However, there is no guarantee that the value of estimate $\hat{g}(\mathbf{x}, t)$ to be nonzero during on-line operations (i.e., controller singularity problem). To avoid this problem, in this work, $g(\mathbf{x},t)$ is assumed to be constant and set to one.

Lemma 1. *Consider the nonlinear system (A.1) (Appendix A) with uncertain nonlinear function $f(\mathbf{x},t)$, which is approximated as (2). Suppose control input is chosen as (A.8) (Appendix A)*

$$u_1(t) = \frac{1}{\hat{g}(\mathbf{x}, t)} \cdot [-\hat{\vartheta}_t^T \psi(\mathbf{x}) + \ddot{x}_d(t) - 2\lambda \dot{e}(t) - \lambda^2 e(t) - k \cdot sgn(s)] \quad (3)$$

where $\hat{\vartheta}$ is the estimate of ϑ^* (ϑ^* is the optimal parameters of ϑ) and the update laws are chosen as

$$\dot{\hat{\vartheta}} = \Gamma s \psi \quad (4)$$

where $\Gamma > 0$ is the adaptation gain matrix and s is the sliding variable defined in (A.5) (Appendix A). Then $s \to 0$ as $t \to \infty$. In the following derivation, $g(\mathbf{x},t)$ can be unknown but it is assumed to be constant and set to one.

Proof of Lemma 1. Consider the Lyapunov function

$$v = \frac{1}{2}s^2 + \frac{1}{2}\tilde{\vartheta}^T \Gamma^{-1} \tilde{\vartheta} \quad (5)$$

where $\tilde{\vartheta} = \vartheta^* - \hat{\vartheta}$ is the estimation error. The time derivative of (5) is

$$\dot{v} = s\dot{s} + \tilde{\vartheta}^T \Gamma^{-1} \dot{\tilde{\vartheta}} \quad (6)$$

Using (A.1) (Appendix A), differentiation of (A.5) (Appendix A) with respect to time can be written as

$$\begin{aligned} \dot{s} &= \ddot{e} + 2\lambda \cdot \dot{e} + \lambda^2 \cdot e = \ddot{x} - \ddot{x}_d + 2\lambda \cdot \dot{e} + \lambda^2 \cdot e \\ &= f^* + u + w - \ddot{x}_d + 2\lambda \cdot \dot{e} + \lambda^2 \cdot e. \end{aligned} \quad (7)$$

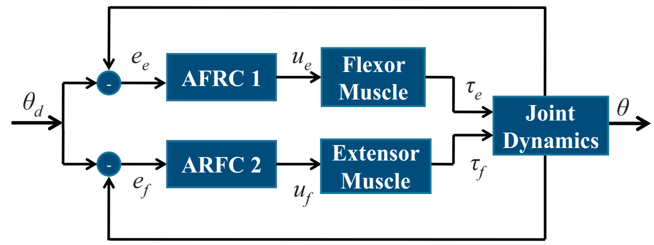

Fig. 2. Block diagram of the proposed decentralized adaptive fuzzy robust control system for control of joint movement using functional electrical stimulation of agonist–antagonist muscles.

where $f^*(\mathbf{x}, t) = \vartheta_t^{*T} \psi(\mathbf{x})$. Apply (7) to (6), we have the following relationship:

$$\dot{v} = s[f^* + u_1 + w - \ddot{x}_d(t) + 2\lambda \cdot \dot{e} + \lambda^2 \cdot e] + \tilde{\vartheta}^T \Gamma^{-1} \dot{\tilde{\vartheta}} \quad (8)$$

Apply (A.8) (Appendix A) and (2)–(8), the following relationship is obtained:

$$\begin{aligned} \dot{v} &= -k \cdot s \cdot sgn(s) + w \cdot s + s \cdot (\vartheta^{*T} - \hat{\vartheta}^T) \cdot \psi - \tilde{\vartheta}^T \Gamma^{-1} \dot{\hat{\vartheta}} \\ &= -k \cdot |s| + w \cdot s + \tilde{\vartheta}^T \cdot (s\psi - \Gamma^{-1} \dot{\hat{\vartheta}}) \end{aligned} \quad (9)$$

Substituting (4) into (9), we obtain

$$\dot{v} = -k \cdot |s| + w \cdot s \quad (10)$$

Let $k = k_2 + \rho$, where $k_2 > 0$ and $|w| \le \rho$. From (10), one obtains

$$\dot{v} = -k_2 \cdot |s| + w \cdot s - \rho \cdot |s| \le -k_2 \cdot |s| \quad (11)$$

Using Barbalat's lemma and Lyapunov theory [33], (11) implies $s \to 0$ as $t \to \infty$. This completes the proof. □

4. Decentralized robust control of joint movement

A representative diagram of the proposed control system of joint movement is shown in Fig. 2. Each muscle–joint complex has its own controller developed in Section 3. To implement the AFRC, the musculoskeletal system should be presented in a standard canonical form as

$$\ddot{\theta} = f_f(\theta, \dot{\theta}) + g_f(\theta, \dot{\theta}) \cdot u_f(t) + w_f(t) \quad (12)$$

$$\ddot{\theta} = f_e(\theta, \dot{\theta}) + g_e(\theta, \dot{\theta}) \cdot u_e(t) + w_e(t) \quad (13)$$

where (12) and (13) present muscle–joint dynamics for flexion and extension movements, respectively. Parameter θ denotes the joint angle, and u_f and u_e are the input commands to the flexor and extensor muscles, respectively. w_f and w_e represent dynamic coupling, parameter uncertainty, unmodeled dynamics, gravity loading of musculoskeletal system, and external disturbances for flexion and extension, respectively. The error signals used for controllers of two muscles are calculated as

$$\begin{bmatrix} e_e \\ e_f \end{bmatrix} = \begin{bmatrix} +1 \\ -1 \end{bmatrix} [\theta - \theta_d] \quad (14)$$

where e_f and e_e are the error signals for controllers of the flexor and extensor, respectively; θ is the measured joint angle, and θ_d the desired trajectory. Fuzzy modeling approach is used to approximate the nonlinear functions $f_f(\theta, \dot{\theta})$ and $f_e(\theta, \dot{\theta})$, and online identification of fuzzy models is performed using adaptation laws derived in Section 3 (4). The gains g_f and g_e, are assumed to be constant and set to one.

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Table 1
Clinical characteristics of paraplegic subjects.

Subject	Sex	Age	Weight (kg)	Height (cm)	Injury level	Feet length (cm)	ASIA[a] class	Date of injury	FNS training
RR	Male	35	76	164	T7	24.5	A	1997	5 Years
MR	Male	28	90	180	T6-T7	28	A	2003	6 Months
EA	Male	25	72	167	T7-T8	25	A	2006	6 Months

[a] American Spinal Injury Association Classification.

5. Experimental studies

5.1. Apparatus and experimental procedure

The experiments were conducted on three thoracic-level complete spinal cord injury subjects (Table 1) using an eight-channel computer-based closed-loop FES system [34]. The paraplegic subjects are active participants in a rehabilitation research program involving daily electrically stimulated exercise of their lower limbs (either seated or during standing and walking) using ParaWalk neuroprosthesis [35]. The experiment sessions for each subject were conducted once a week and each session consisted of different trials with inter-trial resting interval at least 2 min. Totally, 341 experimental trials were conducted on all subjects. At the start of each experimental trial, the subjects were held in a specified standing position by the experimenter. At the instant of activating the controller ($t = 0$ s), the experimenter released the subject to allow the body leans forward and backward. All experimental procedures were approved by the local ethics committee and the subject gave informed consent.

All joints above the ankles were locked using a special body brace (Fig. 3), allowing us to use only ankle strategy to control posture. For safety, the subject wore a fall arrest harness which attached by a pair of lanyards and locking carabiners to a 2-point spreader bar of a ceiling lift system (Maxi Sky 600, ArjoHuntleigh, Sweden) (Fig. 3). The spreader bar height can be adjusted from a handset. During quiet standing experiments, the rope of spreader bar is slackened sufficiently to allow the body to move back and forth in the sagittal plane.

Inclination angle of the body was measured using inertial motion tracking sensor (MTx Motion Tracker, Xsens Technologies, The Netherlands) that was positioned around the third lumbar vertebra (L3) on the subject's back. The subject stood on a force plate (Kistler, Switzerland) that recorded the fluctuation of the center of pressure (CoP). Both inclination angle and CoP were used for stability analyses, while inclination angle was used as sensory feedback.

During the experimental sessions, the dorsiflexor and plantarflexor muscle groups of each leg were stimulated by two pairs of self-adhesive surface elliptical electrodes (10 cm × 5 cm) which were placed over the midlines of the soleus and gastrocnemius muscles for the plantarflexor stimulation and over the tibialis anterior muscles for the dorsiflexor stimulation. Pulse width modulation (from 0 to 700 μs) with balanced bipolar stimulation pulses, at a constant frequency (25 Hz) and constant amplitude, was used. The controller adjusted the pulsewidths and pulsewidths with a negative value were then set to zero. Prior to the experimental session, the best positions of electrodes were determined and the suitable current levels for the flexor and extensor muscles were adjusted. The current levels for flexor (extensor) muscle were 40, 70, and 32 mA (32, 80, 35 mA), in subjects RR, MR, and EA, respectively.

The computer-based closed-loop FES system used Matlab Simulink (The Mathworks, R2007b), Real-Time Workshop and Real-Time Windows Target under Windows 2000/XP for online data acquisition, processing and controlling. The proposed control strategy was implemented by the S-functions using C++.

The fuzzy system used to describe $f(\mathbf{x},t)$ has θ and $\dot{\theta}$ as inputs. For each state variable $\mathbf{x} = [\theta, \dot{\theta}]^T$, five Gaussian-type membership functions are defined as

$$\mu_{A_j^r}(x_j) = \exp\left[-\left(\frac{(x_j - c_j^r)}{\delta_j}\right)^2\right]$$

where $c^1 = -60°$; $c_1^2 = -30°$; $c_1^3 = -0°$; $c_1^4 = -30°$; $c_1^5 = -60°$; $\delta_1 = 30$; $c_2^1 = 16°/s$; $c_2^2 = 132°/s$; $c_2^3 = 248°/s$; $c_2^4 = 364°/s$; $c_2^5 = -480°/s$; and $\delta_2 = 116$. These values are chosen to

Fig. 3. Experimental setup with paraplegic subject RR.

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cover the full possible range of the motion. This fuzzy logic system is used for both $f_e(\mathbf{x},t)$ and $f_f(\mathbf{x},t)$.

The parameters of the controllers were chosen heuristically to achieve the best controller performance during the first session of experiment. There are some guidelines for choosing the SMC parameters [26]. The SMC would require using a high switching gain, k, in order to compensate for the effects of uncertainties. This would also lead to the use of a thicker boundary layer in order to eliminate the higher chattering effect resulting from the use of a high switching gain. Although the chattering behavior can be reduced by increasing the boundary layer thickness, the control system is actually changing to a system without a sliding mode when a thick boundary layer is used. According to these guidelines, the parameters of the proposed SMC were selected as follows:

$$K_{\text{flexor}} = 5, \quad \lambda_{\text{flexor}} = 200, \quad K_{\text{extensor}} = 300, \quad \lambda_{\text{extensor}} = 50,$$
$$\phi = 2, \quad \Gamma = I \times 0.01,$$

and then used for subsequent experiments on different days and for all subjects.

The root-mean-square (RMS) error and maximum peak displacement (MPD) of CoP were calculated as a measure of postural sway and stability during quiet standing as follows:

$$\text{RMS} = \left(\sqrt{\frac{1}{T}\sum_{t=1}^{T}\theta(t)^2}\right)^{\circ},$$

$$\text{MPD} = \underset{1\le t\le T}{Max}(\text{CoP}(t)) - \underset{1\le t\le T}{Min}(\text{CoP}(t))$$

$$\text{CoP}(t) = \text{CoP}_m(t) - \text{CoP}_m(0)$$

where θ denotes the inclination angle, T is the duration of experimental trial, $\text{CoP}_m(t)$ is the measured center of pressure at time t, and $\text{CoP}_m(0)$ is the measured CoP at the vertical position (i.e., inclination angle at zero) at the start of experiment. The MPD is a measure of the CoP deviation in the anterior–posterior direction and can be used to observe stability of subjects during quiet and quasi-quiet standing. Popović et al. [15] proposed a method to observe stability in subjects during quiet and quasi-quiet standing by measuring the CoP position relative to the *base of support*. Four CoP stability zones were identified: high preference, low preference, undesirable and unstable zones. The high preference zone is defined as the area where the CoP is found 99% of the time during quiet standing. The boundary of this zone in anterior/posterior direction is $d_1 = 0.16 \times$ *feet size*. The area where the CoP is found during the remaining 1% of the time is called the low preference zone ($d_1 = 0.57 \times$ *feet size*). To stand safely and comfortably, the subject must have the CoP in the high or low preference zones. The undesirable zone ($d_1 = 0.97 \times$ *feet size*) is defined as the CoP area where the subject is forced to change posture in order to maintain balance, and the unstable zone is defined as the CoP area in which the subject is forced to step forward, backward or sideways to maintain stability.

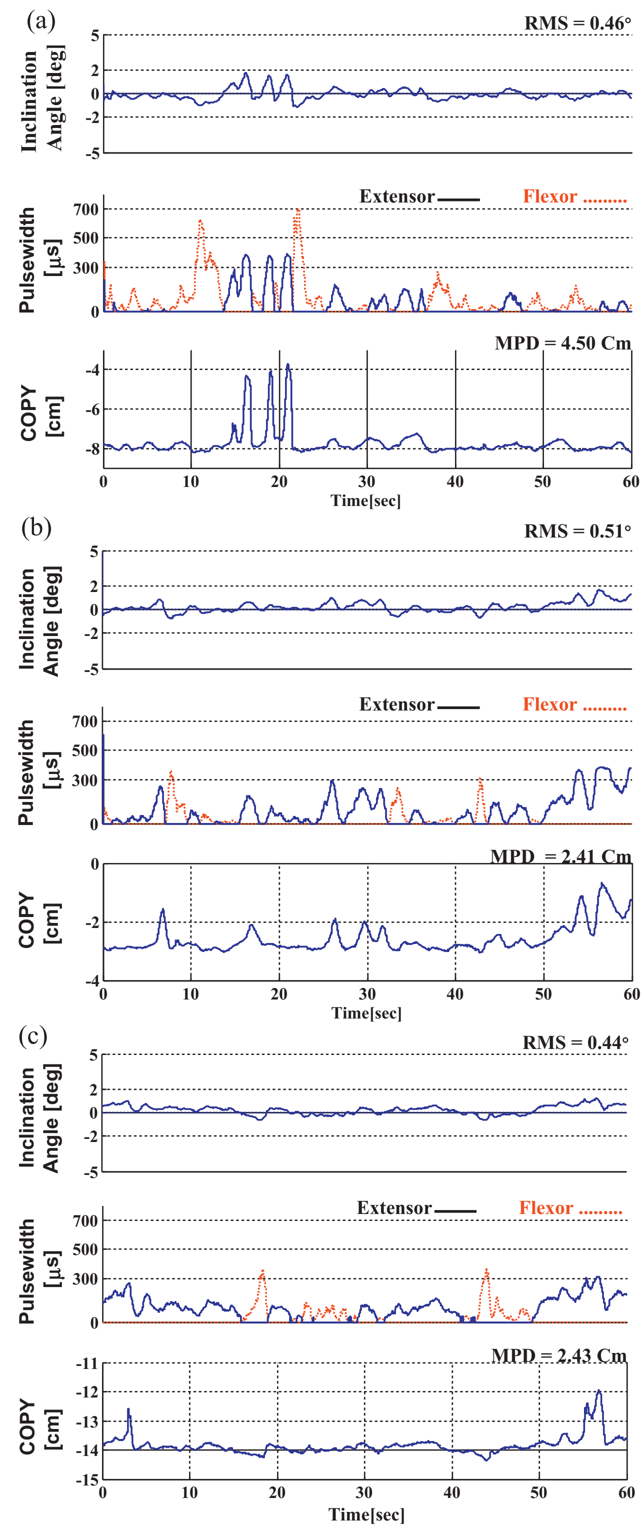

Fig. 4. Typical results of postural stabilization during upright quiet standing on three paraplegic subjects RR (a), MR (b), and EA (c): measured inclination angle, control signals, and measured center of pressure (CoP).

5.2. Results

5.2.1. Free-disturbance quiet standing

During these tests, the subjects were held in an upright standing position at the start of each experimental trial by the experimenter. At the instant of activating the controller ($t = 0$ s), the experimenter released the subject to allow the body leans forward and backward.

Fig. 4 shows typical results of posture control during free-disturbance quiet standing when the paraplegics were at the vertical position (inclination angle at zero) at the start of experiments. It is observed that in response to forward and backward lean, the ankle extensor and flexor were activated, respectively. The controller robustly and rapidly switched the activation between extensor and flexor to correct the body sway. Interesting observation is the fast response of the controller while the controller requires no off-line training phase. This observation is the direct

Table 2
Root mean square of inclination angle, average of maximum peak displacement (MPD) of CoP, and average percent of the time that the subject is in high preference zone (HPZ), low preference zone (LPZ), and undesirable zone (UZ) for three paraplegic subjects during disturbance-free upright quiet standing.

Subject	Day 1	Day 2	Day 3	Day 4	Day 5	Day 6	Day 7	Day 8	Day 9	Mean
RR										
RMS°	2.5 ± 0.6	2.2 ± 1.2	1.7 ± 0.6	2.3 ± 1.5	1.1 ± 0.3	1.6 ± 0.2	1.7 ± 0.2	1.4 ± 0	1.2 ± 0	1.8 ± 0.5
MPD^{cm}	17.3 ± 2	15.0 ± 4	7.0 ± 0.9	6.5 ± 0.9	7.4 ± 2.2	7.1 ± 0.1	7.4 ± 0.4	8.9 ± 0	7.4 ± 0	9.3 ± 3.8
HPZ	88.5%	88.3%	98.0%	98.0%	91.0%	83.0%	87.2%	92.5%	96.7%	91.5%
LPZ	11.1%	11.2%	2.0%	2.0%	9.0%	17.0%	12.8%	7.5%	3.3%	8.4%
UZ	0.4%	0.5%	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%	0.1%
MR										
RMS°	2.0 ± 0.6	1.2 ± 0.7	0.7 ± 0.1	1.2 ± 0.7	0.6 ± 0.3	1.0 ± 0.2	1.6 ± 0.4	4.5 ± 0	0.8 ± 02	1.7 ± 1.5
MPD^{cm}	9.3 ± 2.3	4.8 ± 2.5	3.0 ± 0.3	4.3 ± 1.2	3.3 ± 0.6	5.6 ± 0.9	4.2 ± 1	4.9 ± 0.1	4.9 ± 0.2	5.0 ± 1.8
HPZ	96.2%	85.9%	100.0%	99.0%	99.9%	95.4%	97.8%	100.0%	95.8%	96.7%
LPZ	3.8%	14.1%	0.0%	1.0%	0.1%	4.6%	2.2%	0.0%	4.2%	3.3%
UZ	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%
EA										
RMS°	1.2 ± 0.6	2.0 ± 0.8	1.7 ± 0.6	0.9 ± 0.5	1.0 ± 0.3	1.4 ± 0	1.4 ± 0.1	1.0 ± 0	1.4 ± 0.4	1.3 ± 0.4
MPD^{cm}	9.6 ± 1.9	8.0 ± 0.7	5.8 ± 2.9	4.9 ± 1.7	7.8 ± 2.2	8.3 ± 0.4	5.2 ± 1.7	8.4 ± 0	8.1 ± 0.1	7.4 ± 1.6
HPZ	99.9%	99.7%	99.5%	99.6%	99.9%	99.7%	97.4%	100.0%	100.0%	99.5%
LPZ	0.1%	0.3%	0.5%	0.4%	0.1%	0.3%	2.6%	0.0%	0.0%	0.5%
UZ	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%	0.0%

consequence of the exponential convergence of the output error and its derivative to zero in the proposed control strategy. During this trial of experiment, maximum peak displacements of CoP are 4.5 cm, 2.41 cm, 2.43 cm, in subjects RR, MR, and EA, respectively. The results indicate that CoP was maintained inside the high preference zone [15].

Table 2 summarizes the results of posture control during free-disturbance quiet standing over 106 successful trials for three subjects on different days. Each trial lasted 60 s. It is observed that the average values of MPD over different experiment sessions are 9.3 ± 3.8 cm, 5.0 ± 1.8 cm, and 7.4 ± 1.6 cm for subjects RR, MR, and EA, respectively.

The results indicate that in subject RR, the CoP can be found 91.5% and 8.4% of the time within the high and low preference zones, respectively, and only 0.1% of the time is in the undesirable zone. In subjects MR and EA, the CoP was always maintained inside the high and low preference zones. The average of RMS values of inclination angle is very small and is less than 2.0° for all subjects.

It is very interesting to observe that the standing performance suddenly improved after the first 2 days of the experiment. This improvement can be explained by the muscle training which was done over the experimental sessions and adjustment of the electrode positions during the first sessions of experiments. The positions of electrodes play a key role in standing performance. The electrode positions were determined by stimulating the muscles and observing the movement while the subject sat on his wheelchair before wearing the brace and were adjusted after the first sessions of standing experiments.

Fig. 5 shows a typical successful test of posture control during free-disturbance quiet standing when the subjects were held at the forward or backward lean of about 5° at the start of experiment. It is observed that the controller was capable of returning the body to normal quiet standing posture. Interesting observation is the fully activation of ankle flexors in response to backward sway in all subjects. In spite of saturating the muscle activation, the controller was able to correct the backward body sway. It can be seen from Fig. 5(c) that CoP position was maintained inside the undesirable zone near low preference zone; however, the controller could stabilize the posture and return the body to vertical standing position. The results presented here are typical of 11 successful trials (out of a total of 20 experimental trials) with all subjects.

5.2.2. *Effect of external disturbances*

To evaluate the capabilities of the proposed control strategy to external disturbance rejection during quiet standing, a number of experiments were conducted on the paraplegic subjects holding a 2.5 kg hand weight in each hand (Fig. 3). The tests were carried out while the subjects were at the vertical position at the start of experiments. At 22 s, the subject was asked to raise one hand weight from hanging at arms' length to shoulder height with the arm in front of trunk (Fig. 3) and to hold for 10 s. At 42 s, the subject was informed to raise one hand weight from hanging in backward direction as high as possible and to hold for 10 s. Fig. 6 shows typical results of disturbance rejection using proposed control strategy during quiet standing for three subjects. It is observed that forward and backward lifting caused the body swayed forward and backward, respectively. However, the controller activated the extensor and flexor muscles to prevent the body to fall. The values of MPD are 6.5, 3.7, and 7.0 cm for the subjects RR, MR, and EA, respectively, and the RMS values of inclination angle are less than 1.4°.

Table 3 summarizes the results of disturbance rejection for 76 experimental trials which were conducted on three subjects during

Table 3
Root mean square of inclination angle and average of maximum peak displacement (MPD) of CoP for three paraplegic subjects during perturbed standing.

Subject	Day 1	Day 2	Day 3	Day 4	Day 5	Day 6	Day 7	Day 8	Mean
RR									
RMS°	1.3 ± 0.5	1.3 ± 0.3	1.0 ± 0.2	1.0 ± 0.2	1.5 ± 0.4	1.2 ± 0.5	1.5 ± 0.4	–	1.3 ± 0.4
MPD^{cm}	6.8 ± 1.8	6.3 ± 0.1	4.4 ± 0.3	4.4 ± 0.3	7.2 ± 0.1	6.1 ± 0.9	7.8 ± 1.5	–	6.4 ± 1.5
MR									
RMS°	0.8 ± 0.1	0.9 ± 0.2	1.4 ± 0.2	1.6 ± 0.4	1.5 ± 0.4	–	–	–	1.0 ± 0.3
MPD^{cm}	3.3 ± 0.4	3.6 ± 0.5	4.9 ± 0.1	4.2 ± 1.0	3.4 ± 0.5	–	–	–	3.7 ± 1.0
EA									
RMS°	3.0 ± 0.3	1.2 ± 0.1	1.4 ± 0.3	1.2 ± 0.1	1.8 ± 0.2	1.7 ± 0.1	1.8 ± 0.5	2.8 ± 0.0	1.7 ± 0.7
MPD^{cm}	15.2 ± 0.9	3.6 ± 0.9	3.6 ± 0.4	3.7 ± 0.3	3.4 ± 0.9	4.5 ± 1.1	9.0 ± 1.5	8.3 ± 0.0	6.5 ± 3.7

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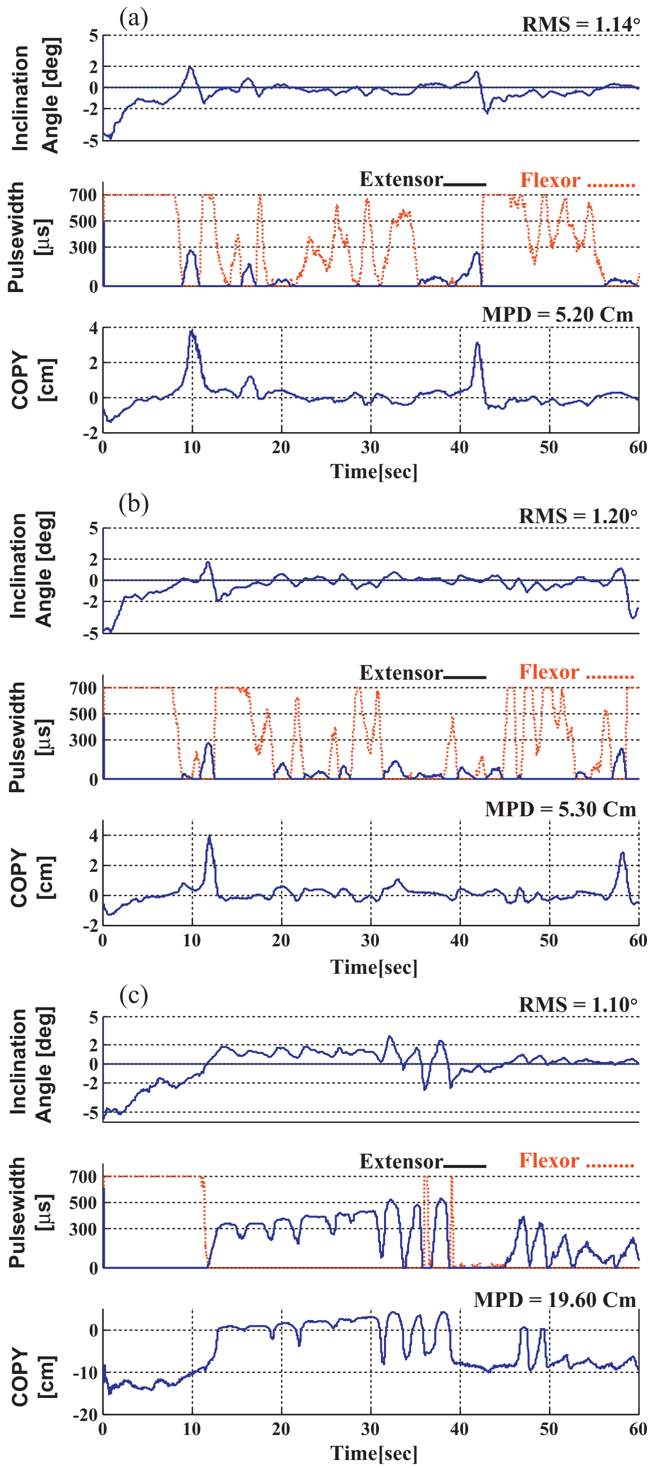

Fig. 5. Typical results of upright standing stabilization on three paraplegic subjects RR (a), MR (b), and EA (c) when the subjects were held at the forward or backward lean of about 5° at the start of experiment.

different days. The average of maximum peak CoP displacements are 6.4 ± 1.5, 3.7 ± 1.0, and 6.5 ± 3.7 cm for the subjects RR, MR, and EA, respectively. The results indicate that during perturbed standing, the control strategy could stabilize the postural system in such a way that the CoP was maintained near the high preference zone.

The results indicate that the proposed adaptive robust control design method can achieve favorable disturbance rejection. It is interesting to note that the average peak CoP displacements

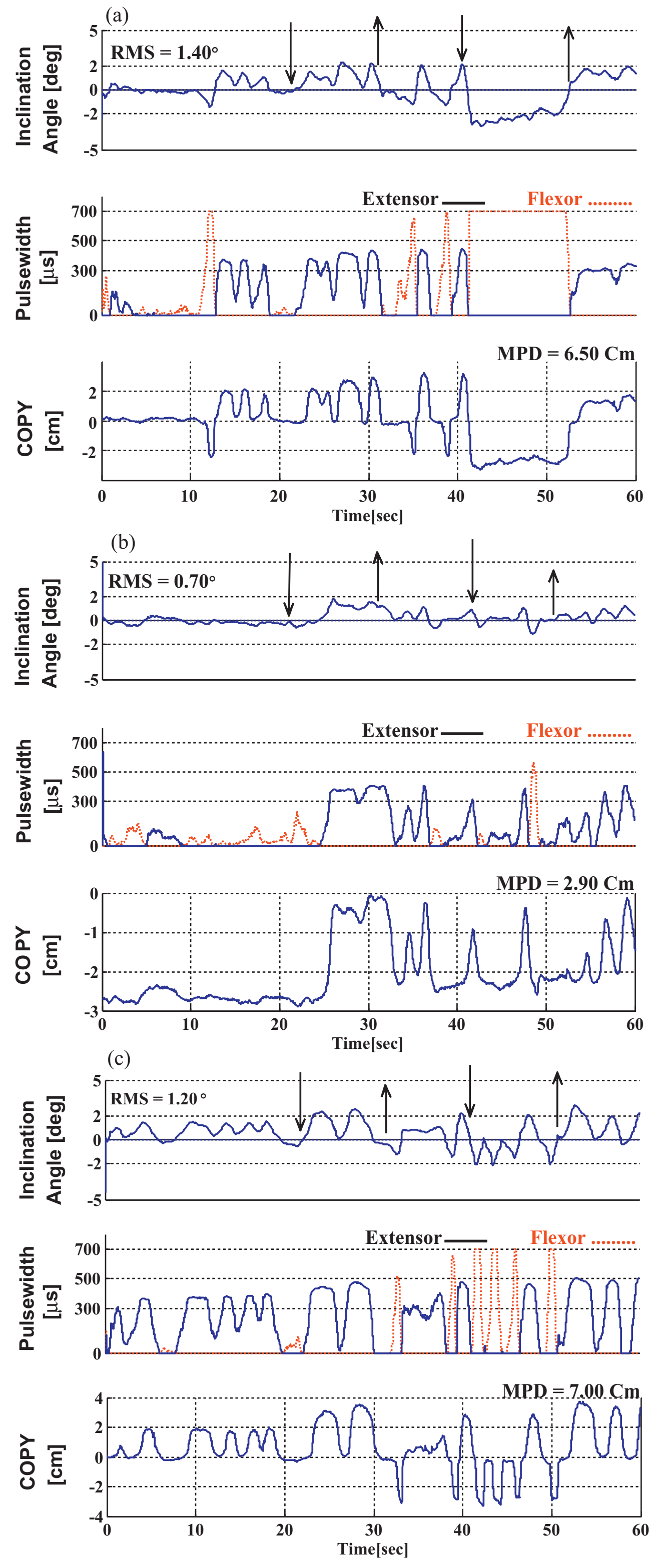

Fig. 6. Typical results of disturbance rejection during upright quiet standing on three paraplegic subjects RR (a), MR (b), and EA (c). At 22 s, the subject was asked to raise one hand weight from hanging at arms' length to shoulder height with the arm in front of trunk and to hold for 10 s. At 42 s, the subject was informed to raise one hand weight from hanging in backward direction as high as possible and to hold for 10 s.

obtained during perturbed standing (Table 3) are somewhat less than that obtained during disturbance-free quiet standing (Table 2). A possible suggestion for this finding is the increasing the control input which causes increasing the ankle stiffness during raising the hand weight.

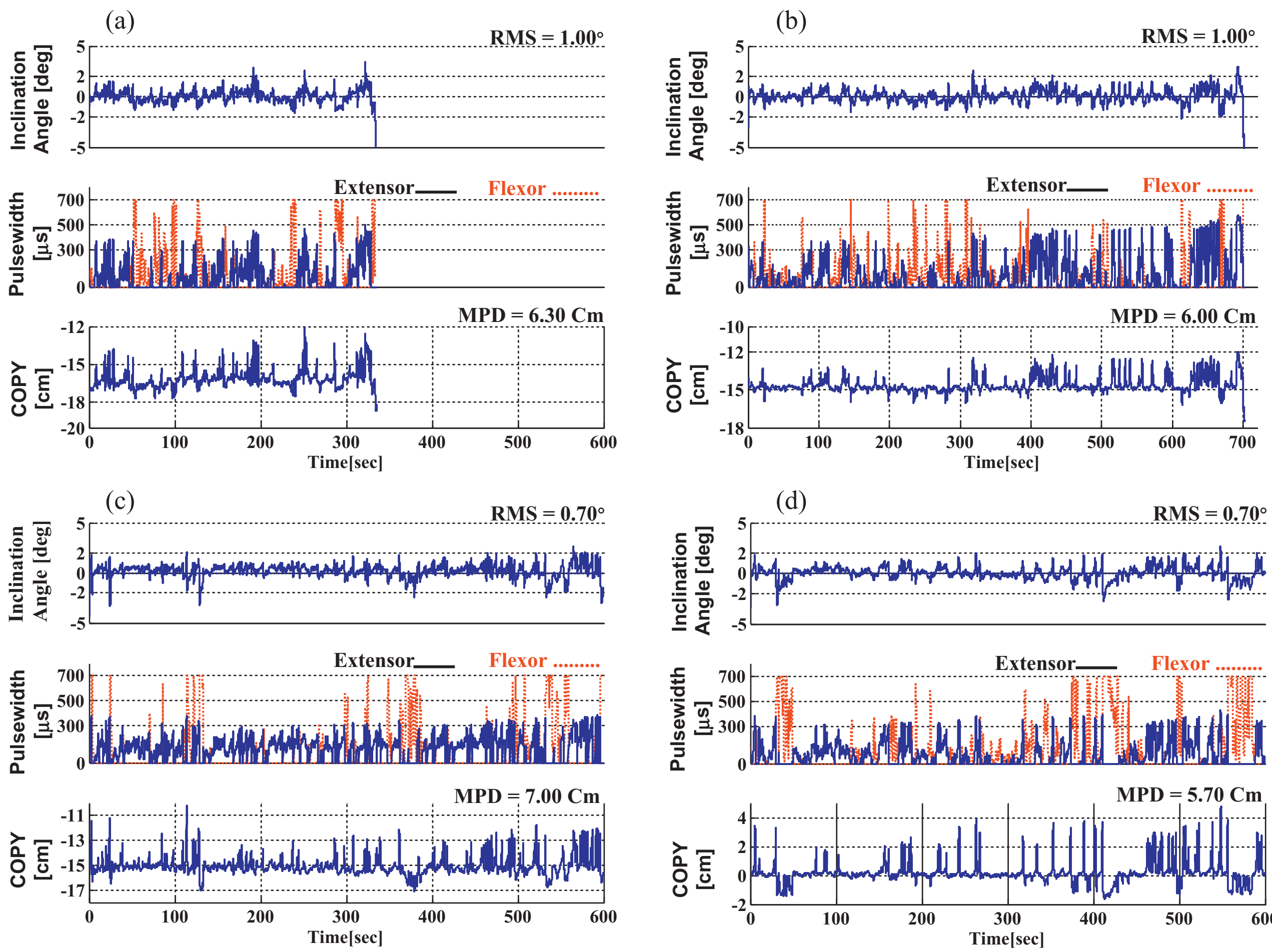

Fig. 7. Typical results of muscle fatigue compensation during long period standing on three paraplegic subjects MR (a, b), RR (c), and EA (d).

5.2.3. Muscle fatigue compensation

In human subjects, it has been shown that muscle fatigue of the limbs affects the postural control by increasing the body sway [36,37]. It is, therefore, important to investigate the ability of the controller to account for muscle fatigue during long period standing. To evaluate the ability of proposed control strategy to compensate the muscle fatigue during free-support quiet standing, a number of experimental trials with long periods of time were conducted on paraplegic subjects. The experiments were carried out while the subjects were at the vertical position at the start of tests and were allowed to run until the muscles fatigued and caused the subject lost the stability and fell down. Note that the subject was protected by the fall arrest system. During the first sessions of experiment, duration of upright standing that can be maintained was short, but as the number of experiments which were conducted on the subject increased, the subject could stand for a longer period.

Fig. 7 shows a typical result of fatigue compensation during quiet standing. Fig. 7(a) shows the result of the first fatigue test on the subject MR. The subject was standing up for 5 min and 40 s, after which he fell backward and was caught by the fall arrest system. Interesting observation is the activation of dorsiflexor muscle group which was saturated occasionally for a short period of time ($t \approx 1$ s) to correct the backward postural lean. In spite of saturation, the controller is able to maintain the upright standing posture at least for the first 5 min and 40 s. Fig. 7(b) shows the result of posture control on the same subject following three sessions of experiment. The subject could stand quietly for 11 min and 40 s. Fig. 7(c) and (d) shows the results of long quiet standing for two other subjects RR and EA, while the experiments were stopped at 10 min.

The results show that the method could adjust the stimulation pattern to compensate the effect of the muscle fatigue and standing performance remains fairly constant throughout the trial. In all subjects, it is observed that the body sway and peak displacement of CoP were increased by passing the time. These results are in agreement with experimental findings in healthy human subjects [36,37]. Interesting observation is the discontinuous activation of the muscle groups during upright quiet standing. The control input was switched between extensor and flexor for control of posture. This causes standing to achieve with a longer periods of time. We found that the fatigue test can be repeated after resting periods of up to 5 min. The results presented here are typical of 12 trials with all subjects.

6. Discussion and conclusion

We have developed a novel decentralized robust methodology for posture control during upright quiet standing. Extensive experiments on three paraplegic subjects demonstrate the exceptional performance and robustness of the proposed control strategy during long period standing and external disturbances.

The current methods for closed-loop control of unsupported standing require offline identification before they could be used to control the limbs [8–13]. The burdens of pre-training may hinder the clinical applications of these methods. A major contribution of our work in this study is that the proposed control scheme requires no prior knowledge about the dynamics of *neuromusculoskeletal* system and no offline learning phase. Compared to the previous work [25], the proposed ARFC has the advantage of online adaptation ability to handle the plant variations or changing environments. The results clearly indicate that the proposed method makes system motion robust with respect to the time-varying properties of musculoskeletal dynamics, day-to-day and subject-to-subject variations. This result is achieved without using any subject-specific model information. The parameters of the controller were adjusted during the first session of experiment and used for subsequent sessions on different days for all subjects. These properties of the controller have a great implication for developing practical FES systems.

In contrast to previous works [12,13], which used voluntary motor control skills of the upper body integrated within the overall control scheme for vertical standing stabilization, the current method stabilizes the posture during vertical quiet standing by alternative activation of dorsiflexor and plantarflexor muscle groups without contributing the upper body. This is an important issue for developing arm-free FES supported standing system in paraplegia.

The mechanisms of postural stabilization during *upright quiet standing* in normal subjects were investigated by many researchers [16–20]. Experiments observations on normal human subjects by others have shown that for small disturbances, *healthy subjects prefer to move primarily about the ankles to maintain erect stance.* When the body is at the extremes of forward or backward lean, the hip strategy must be employed [16]. In the current experimental studies, we fixed the knee and hip joints and used the closed-loop FES control of ankle plantar and dorsal flexors for standing posture stabilization The results indicate that postural stabilization can be achieved, for small disturbance, using only ankle strategy, however, involving both the ankle and the hip joints in postural stabilization using proposed strategy remains as an open problem.

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Appendix A. Brief introduction of sliding mode control (SMC)

Consider the following nonlinear system:

$$\ddot{x} = f(\mathbf{x}, t) + g(\mathbf{x}, t) \cdot u_1(t) + d(t) \tag{A.1}$$

where $x(t)$ is the state to be controlled so that it follows a desired trajectory $x_d(t)$, $d(t)$ the external disturbances which is unknown but bounded by the known function, i.e., $|d(t)| \le D$, and $u_1(t)$ the control input. The nonlinear dynamics $f(\mathbf{x},t)$ and control gain $g(\mathbf{x},t)$ are not known exactly, but are estimated as the known nominal dynamics $\hat{f}(\mathbf{x}, t)$ and $\hat{g}(\mathbf{x}, t)$, respectively, with the bounded estimation errors. With uncertainties, the dynamic equation of the system (A.1) can be modified as

$$\begin{aligned}\ddot{x} &= (\hat{f}(\mathbf{x}, t) + \Delta f(\mathbf{x}, t)) + (\hat{g}(\mathbf{x}, t) + \Delta g(\mathbf{x}, t)) \cdot u_1(t) + d(t) \\ &= \hat{f}(\mathbf{x}, t) + \hat{g}(\mathbf{x}, t) \cdot u_1(t) + w(\mathbf{x}, t)\end{aligned} \tag{A.2}$$

where $\Delta f(\mathbf{x},t)$ and $\Delta g(\mathbf{x},t)$ denote unmodeled dynamics and parameter uncertainties; $w(\mathbf{x},t)$ is the lumped uncertainty and defined as

$$w(\mathbf{x}, t) = \Delta f(\mathbf{x}, t) + \Delta g(\mathbf{x}, t) \cdot u_1(t) + d(t) \tag{A.3}$$

Here the bound of the lumped uncertainty is assumed to be given; that is,

$$|w(\mathbf{x}, t)| < \rho \tag{A.4}$$

The objective of the controller is to design a control law to force the system state vector to track a desired state vector in the presence of model uncertainties and external disturbances. We first define a sliding surface as follows:

$$s(e, t) = \left(\frac{d}{dt} + \lambda\right)^2 \left(\int_0^t e(r)dr\right) = 0 \tag{A.5}$$

where $e(t)$ is the state error and λ is a positive constant. By solving the above equation for the control input using (A.1), we obtain the following expression for $u_1(t)$ which is called equivalent control:

$$\begin{aligned}u_{eq}(t) &= \frac{1}{\hat{g}(\mathbf{x}, t)} \cdot (-\hat{f}(\mathbf{x}, t) + \ddot{x}_d(t) - 2\lambda\dot{e}(t) - \lambda^2 e(t)) \\ &= \frac{1}{\hat{g}(\mathbf{x}, t)} \cdot \hat{u}(t)\end{aligned} \tag{A.6}$$

The equivalent control keeps the system states in the sliding surface $s = 0$ if the dynamics were exactly known. Hence, if the state is outside the sliding surface, to drive the state to the sliding surface, we choose the control law such that

$$\frac{1}{2}\frac{d}{dt}s^2 \le -\eta|s| \tag{A.7}$$

where η is a strictly positive constant and (A.7) is called reaching condition. The control objective is to guarantee that the state trajectory can converge to the sliding surface. It can be proved that the control law

$$u_1(t) = \frac{1}{\hat{g}(\mathbf{x}, t)} \cdot [\hat{u}(t) - k \cdot sgn(s)] \tag{A.8}$$

with $k \ge \rho + \eta$ satisfies the sliding condition (A.7) [23].

Conflict of interest statement

There are no conflicts of interest for the authors of this study.

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