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Identification of ankle plantar-flexors dynamics in response to electrical stimulation

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A B S T R A C T

Modeling the muscle response to functional electrical stimulation (FES) is an essential step in the design of closed-loop controlled neuroprostheses. This study was aimed at identifying the dynamic response of ankle plantar-flexors to FES during quiet standing. Thirteen healthy subjects stood in a standing frame that locked the knee and hip joints. The ankle plantar-flexors were stimulated bilaterally through surface electrodes and the generated ankle torque was measured. The pulse amplitude was sinusoidally modulated at five different frequencies. The pulse amplitude and the measured ankle torque fitted by a sine function were considered as input and output, respectively. First-order and critically-damped second-order linear models were fitted to the experimental data. Both models fitted similarly well to the experimental data. The coefficient of variation of the time constant among subjects was smaller in the case of the second-order model compared to the first-order model (18.1% vs. 79.9%, $p < 0.001$). We concluded that the critically-damped second-order model more consistently described the relationship between isometric ankle torque and surface FES pulse amplitude, which was applied to the ankle plantar-flexors during quiet standing.

1. Introduction

Functional electrical stimulation (FES) has been largely utilized to improve or restore the lost motor function in individuals with spinal cord injury, traumatic brain injury, stroke or other neuromuscular impairments [1–3]. FES is referred to applying patterned electrical pulses to intact motor neurons of paralyzed (or paretic) muscle to artificially induce muscle contractions, and to restore functional motor tasks [4,5]. Over the years various efforts have been made to develop closed-loop controlled FES system for standing, i.e., neuroprosthesis for standing [6–9]. In these systems the ankle flexor muscles were the primary targets for FES, as the ankle joint primarily controls the body equilibrium during quiet standing. The dynamic response of these muscles to FES is an integral component of a closed-loop controlled neuroprosthesis system and precise modeling of this response is critical in the design of the neuroprosthesis system.

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Several models have been suggested to represent muscle response to FES. Linear [10] and nonlinear [8] static models were considered to represent relationship between the ankle torque and the FES pulse amplitude. A first-order linear model [11] and Hammerstein model [12,13] were used to represent the ankle torque as a function of FES pulse duration where FES was applied on ankle flexors in quiet standing posture. Hunt et al. [12] used pseudorandom binary sequences (PRBS) as input for model parameters estimation and showed that multiple locally linear second-order models, as applied elsewhere [7,14,15], are more accurate than a second-order Hammerstein model. Other studies compared several models (e.g., linear models of various orders, Hammerstein model and Hill-Huxley model) for representing the torque generated by plantar-flexors [16,17] or dorsi-flexors [18] as a function of FES pulse intensity while subjects were seated. These studies applied a dozen of single FES pulses (twitches) with various frequency modulation schemes and found that the fitting error obtained with the second-order linear model was only around 4% [18] or up to 8% [16,17] larger than that obtained with the most accurate nonlinear model.

Although many complex models have been used to represent muscle dynamic responses to FES, yet there is no consensus in the

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literature which model is the most optimal [18]. The model of the muscle dynamic response to FES was often driven by the choice of a closed-loop control strategy that researchers investigated. Linear control strategies were frequently used to facilitate the linear controller design because of their straightforward design and implementation [7,11,19–21] and their similarity to the physiological control strategies [6,10], even if their estimation may be slightly less accurate than nonlinear models. Typically, first- and second-order models have been considered, while higher-order linear models were seldom recommended in the literature. Yet, a comprehensive comparison of the performance of first- and second-order models is lacking. Determining the most appropriate order for such linear models is also beneficial in identification of nonlinear models that have linear components (e.g., Hammerstein model or multiple locally linear models).

On the other hand, estimating the linear model parameters and comparing their performance based on transient response of the muscle can be challenging. Former studies, which compared muscle dynamic response models, typically analyzed the transient response of muscles to short input sequences (either a short PRBS with modulating pulse duration or a dozen of single pulses with modulating time interval among them). The estimation of model parameter in this approach can be considerably affected by involuntary (automatic) reflexes or voluntary activation of the muscle [18,22], as evident by difference between the estimated model parameters obtained with different short-term sequences as input [12,22]. At the same time, reflexes are less likely to be triggered by slowly changing stimulation signals, compared to a train of twitches. Therefore, we assumed that identification of first- or second-order linear models based on the steady-state response of the muscle to sinusoidal excitation would reduce the likelihood of voluntary activation or reflexes of muscles and would be more accurate than evaluating the noisy transient response of muscles to a train of twitches or short-term PRBS. As such, the present study was aimed at developing a novel methodology to identify dynamic responses of the ankle plantar-flexors to surface FES in standing posture, with harmonically changing pulse amplitudes. We represented these relationships as first- or second-order linear transfer functions between FES pulse amplitude (input) and the ankle joint torque (output).

2. Methods

2.1. Measurement setup

We developed a measurement setup to isolate ankle plantar-flexors from the physiological control system of standing, while maintaining the standing posture [8,23]. In this measurements setup, voluntary activation and involuntary (automatic) reflexes of the plantar- and dorsi-flexors was minimized [24]. The subject was supported in an upright, stationary standing position by a mechanical frame (Ottobock, Germany) that locked the knees and hips extended (Fig. 2). Subject's feet, positioned horizontally, were fixed firmly to the foot-plates via foot straps over the toes (phalanges) and midfoot distal to the ankle joint (navicular and cuneiform bones). The foot-plates were attached to a shaft, orthogonal to the subject's sagittal plane. A torque transducer (TS11-200 Flange Style Reaction Torque Transducer, Interface, USA) was mounted on this shaft to record the isometric torque applied by the plantar-flexors on the foot-plates. A programmable FES system, Compex Motion II (Compex SA, Switzerland) was utilized to bilaterally stimulate to the plantar-flexors through surface electrodes. 5.5 cm × 9 cm surface electrodes (Stimtrode ST5090, Nidd Valley Medical Ltd, UK) were placed bilaterally along the midline of the posterior calf, approximately 2 cm below the popliteal fossa over the gastrocnemius and soleus muscles' motor points, and above the Achilles tendon and over the lower end of the gastrocnemius muscle belly. These electrodes were used to stimulate the soleus and both gastrocnemius heads. Gel was applied beneath electrodes to increase skin conductivity. A data acquisition device (DAQ Multifunction NI USB-6211, National Instruments, USA) was utilized for regulating the stimulator in real-time and to record the measurement data at the sampling frequency of 100 Hz.

2.2. Experimental protocol

13 subjects (5 female, 24 ± 5 years old, 170 ± 6 cm, 65 ± 10 kg) were recruited. All subjects were able-bodied individuals with no known neurological or musculoskeletal disorders. Each subject gave written informed consent to participate in the study, which prior to the experiment was approved by the local Research Ethics Board. The apparatus introduced in Section 2.1 locked each subject in the neutral quiet standing posture. Trains of FES pulses were bilaterally applied to the plantar-flexors and the exerted isometric ankle torque was recorded. The stimulation pulses were rectangular, balanced, biphasic and asymmetric, and were applied at frequency of 20 Hz (this frequency was chosen to minimize the muscle fatigue [25]). Thus, the FES pulse amplitude could be updated every 50 ms. The pulse amplitude was modulated to produce sinusoids with frequencies of 0.07, 0.15, 0.3, 0.75 and 1.2 Hz. Sinusoidal varying FES amplitudes were between 20 and 60 mA. In our measurement setup, the generated torque with FES amplitudes below 20 mA was negligible and with amplitudes above 60 mA saturated. Therefore, linear relationship between FES amplitudes and torque was not necessarily observed out of this range of amplitudes. Out of 13 subjects, four reported discomfort with FES amplitude of 60 mA in initial testing and thus maximum amplitude was reduced to 55 mA for three and 50 mA for one of them. Notably, this reduced range does not affect the applied linear system identification. The pulse duration was set to 300 µsec to allow for a sufficiently gradual rate of change of torque generation at the minimum FES amplitude step size of 1 mA. The durations of each stimulation trial 10, 10, 11, 16, and 30 s for frequencies of 1.2, 0.75, 0.3, 0.15, 0.07 Hz with a rest time between each two consecutive stimulation trials (see Fig. 1). This duration was chosen to last at least 10 s and long enough to record two complete periods of pulse amplitude modulation sinusoids.

2.3. Data analysis

In order to avoid the influence of transient response of muscle, the first 5 s of each trial was disregarded when fitting the torque output curves to sinusoids. We assumed that the ankles respond bilaterally symmetrically to the synchronously applied FES pulses. The sinusoid of FES pulse amplitude and the fitted sinusoid on the exerted isometric ankle torque were considered as input and output, respectively, to model the muscle dynamic response (Fig. 2). The gain amplitude (A) and phase lag (ϕ) between the input and output were calculated for each frequency of the applied input sinusoid and plotted the in Bode diagrams, as a total of five points (Fig. 3). We previously applied step-wise FES pulses with amplitude from 20 to 60 mA, with the same measurement configuration and observed that, within this range of FES pulse amplitudes, plantar-flexors respond virtually linearly to variation of FES pulse amplitude [8]. Therefore, linear model for muscle responses was an acceptable assumption in the present study. We fitted first-order and critically-damped second-order models on the five measured gain amplitude (A_1 to A_5) and phase difference (ϕ_1 to ϕ_5) between input and output to identify the muscle transfer functions, as follows:

$$M_1(s) = e^{-\tau_1 . s} \times \frac{K_1}{1 + \alpha_1 . s} \quad (1)$$