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

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ABSTRACT

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.

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:

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

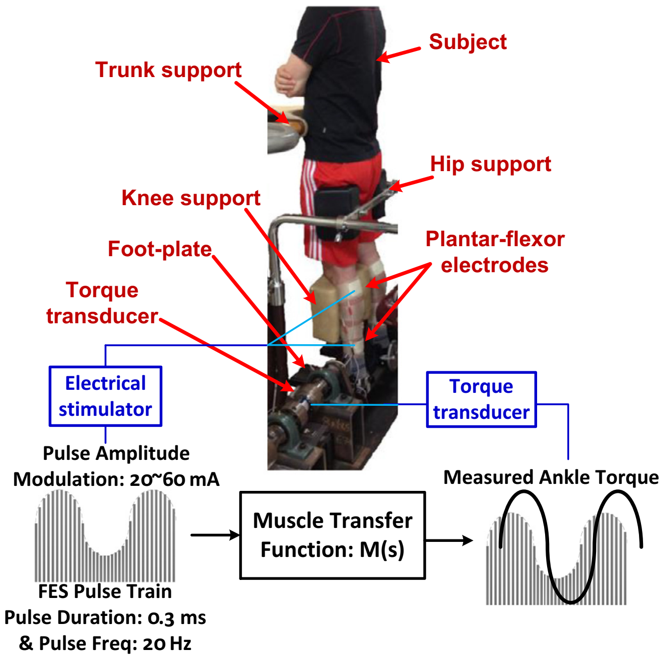

Fig. 1. Measurement setup: Subject stood on a standing frame that locked the knee and hip joints in an extended position (see [23] for more details on this figure). The ankle plantar-flexors were stimulated bilaterally through surface electrodes and the generated ankle torque was measured. The stimulus waveform was rectangular with fixed pulse frequency and pulse duration. The pulse amplitude was modulated to change as a sinusoidal function with different frequencies. Amplitude gain and phase lag were calculated for each frequency of the sinusoidal function.

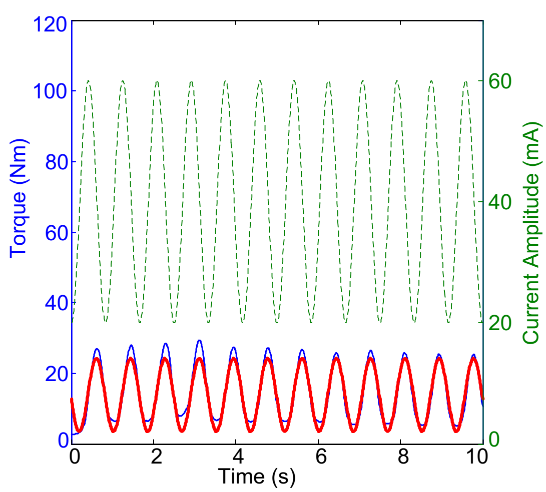

Fig. 2. The dynamics response of ankle plantar-flexors for a representative subject: In the plot the green dashed curve represents inputted sinusoidally-varying FES pulse amplitude (0.3 Hz), whilst the blue curve represents the measured ankle torque as output. The red curve depicts sinusoidal waveform of best fit to the torque output curve. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

$$M_2(s) = e^{-\tau_2 . s} \times \frac{K_2}{(1+\alpha_2 . s)^2} \tag{2}$$

where K_1 and K_2 are static (DC) gains, α_1 and α_2 are time constants, and τ_1 and τ_2 are time delays. $M_1(s)$ and $M_2(s)$ are the first- and critically-damped second-order transfer functions, respectively, for plantar-flexors. A custom-written program (Matlab, Mathworks, USA) was used to estimate the parameters of each model (K_1, K_2, α_1, α_2, τ_1 and τ_2) for an optimized fit for the each subject. Then, the estimated gain amplitude (A'_1 to A'_5) and phase lag (ϕ'_1 to ϕ'_5) for the five measured frequencies based on $M_1(s)$ and $M_2(s)$ were calculated for each subject. The root mean square (RMS) difference and correlation coefficients between the experimentally measured and estimated gain amplitude and phase lag were calculated for each subject and muscle group in order to investigate the model fitting accuracy. Finally, the mean and coefficient of variation (CV%=100 × SD/mean) for static (DC) gains, time constants, and time delays, as well as RMS difference and correlation coefficients of fitting among all subjects were calculated for each model (first- and second-order) in order to assess the consistency (inter-subject repeatability) of model fitting among the subjects.

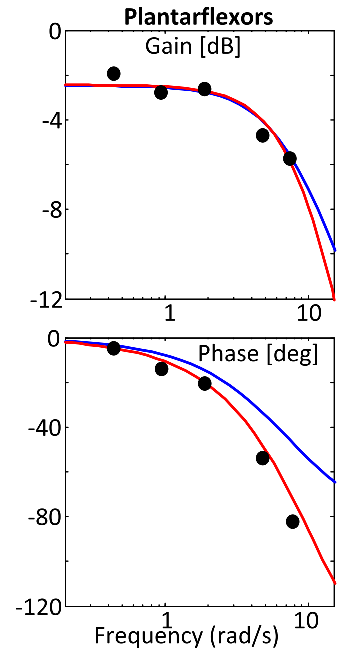

Fig. 3. Bode diagrams of the muscle response to FES for a representative subject: Bode diagrams of gain amplitude and phase for the ankle plantar-flexors obtained experimentally (black dots), and the fitted first-order (blue) and critically-damped second-order (red) models. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

3. Results

The measured gain amplitude and phase lag for the five frequencies, and the fitted first- and second-order models were plotted in Bode diagrams (see Fig. 3, for a representative subject). According to Table 1, both models fitted well the experimental data of plantar-flexors: correlation coefficient around 0.90 for amplitude gain and 0.88 for phase lag (average among all subjects). One-way ANOVA revealed that both first- or second-order models were not significantly different in correlation coefficient and RMS difference for both amplitude gain and phase lag.

The inter-subject variability of the model parameters was expressed as CV% (Table 1). The CV% of the time constants obtained by the second-order model (18.1%) was significantly smaller than that obtained by the first-order model (79.9%) (F-test, $p < 0.001$). Therefore, the time constant obtained by the critically-damped second-order model was more consistent among subjects for plantar-flexors. Table 2 shows the correlation coefficients between the fitted sinusoidal curves and measured torques.

Table 1
(a) Correlation coefficients and RMS differences between experimental and fitted amplitude gains and phase lags in the transfer function between the FES pulse amplitude and the generated ankle torque by plantar-flexors. (b) Estimated static (DC) gain, time constant and time delay in the fitted models for the transfer function of plantar-flexors response to FES. In (a) and (b) the results are presented as mean ± SD (standard deviation) among the 13 subjects, for the first-order and critically-damped second-order models.

	1st order	2nd order
(a) Fitting quality		
Correlation coefficient		
Amplitude gain (A)	0.92 ± 0.14	0.90 ± 0.19
Phase lag (ϕ)	0.88 ± 0.18	0.89 ± 0.30
RMS difference		
Amplitude gain (A)	0.21 ± 0.29	0.20 ± 0.21
Phase lag (ϕ)	14.6 ± 18.8	16.4 ± 22.7
(b) Estimated model parameters		
Static (DC) gain (K) (N m/mA)	2.90 ± 1.04	2.75 ± 0.98
Time constant (α) (s)	0.26 ± 0.21	0.11 ± 0.02
Zero-frequency time delay (τ) (s)	0.07 ± 0.04	0.03 ± 0.03

Table 2
Fitting accuracy of sinusoidal curves to the measured torques generated by plantar-flexors. Correlation coefficients are presented between measured and fitted torque time-series at five frequencies. Maximum and minimum of the measured torque (in 20 and 60 mA) are also presented. The results are presented as mean ± SD (standard deviation) among the 13 subjects.

	$f = 0.07$ Hz	$f = 0.15$ Hz	$f = 0.3$ Hz	$f = 0.75$ Hz	$f = 1.2$ Hz
Correlation coefficient for fitting	0.96 ± 0.03	0.96 ± 0.02	0.97 ± 0.02	0.97 ± 0.02	0.97 ± 0.02
Max. measured torque [N m]	122.7 ± 39.8	119.9 ± 38.0	111.3 ± 39.9	101.9 ± 35.4	90.1 ± 32.6
Min. measured torque [N m]	13.8 ± 16.0	15.1 ± 15.2	15.0 ± 18.7	20.2 ± 22.4	25.3 ± 22.7

4. Discussion

Identification of muscle dynamic response to FES is essential in the design of closed-loop controlled FES systems that are used to restore lost motor functions. The transient response of muscles to short-term stimulation can be affected by voluntary activation or involuntary (automatic) reflexes of muscles [18,22]. Indeed, although PRBS or twitch trains in short-term theoretically excite many frequencies at the same time, the power of excitation in each frequency may not be high enough for robust parameter identification. The varying result for linear model parameters in even one study obtained by different transient excitations supports this shortcoming of transient excitations (for example, see [12,22]). On the other hand, although a few sinusoidal excitations (in similar periods of time to short-term excitations) are not enough to identify the model order, they are able to accurately obtain amplitude gains and phase lags which can estimate the model parameters when a first- or second-order model is assumed. Therefore, we proposed a novel methodology to identify this dynamic response as linear models by evaluating the steady-state response of muscles to a number of harmonic FES inputs. For this purpose, we utilized a custom-made mechanical setup that isolated ankle plantar-flexors by means of disrupting, to a large extent, the sensory information that is used by the physiological control system to regulate balance. At the same time, previous studies that compared muscle response models usually applied modulation of FES pulse duration or time interval between single twitches. However, the muscle response to pulse amplitude modulation can be different [18]. In the present study for the first time, we compared first- or second-order models for muscle response to modulation of FES pulse amplitude. This modeling is especially advantageous because many closed-loop FES controllers apply continuous FES pulse trains with modulating pulse amplitude rather than twitches.

4.1. Accuracy of fitting

Both correlation coefficient and RMS difference between the experimental and fitted amplitude gain and phase lag showed no significant difference between first- and second-order models. Therefore, none of the first- and second-order models was found a better fit. We will further discuss the suitability of these models in the following sections.

4.2. Consistency among subjects

The CV% for estimated static gain was around 35%. We did not expect the estimated static (DC) gain (K) to be consistent (highly repeatable) among subjects. The estimated static gain is a function of the individual's muscle strength, electrode placement, and soft tissue conductivity, thus and naturally varies among individuals and experiments. However, the estimated time constant is a function of physiological response of muscle to FES and can show higher consistency among subjects. CV% of the time constants obtained by the second-order model (18.1%) was significantly smaller than that obtained by the first-order model (79.9%). We concluded that the critically-damped second-order model obtained more consistent time constant among subjects and thus, may be a better choice for modeling the dynamic response of ankle plantar-flexor. The inter-subject repeatability of the identified muscle model is particularly beneficial in the design of closed-loop FES controllers. Notably, the static gain of muscle response model considerably varies among subjects and requires individualized FES controller design. Nevertheless, a model in which the time constant is in a similar range among a group of subjects, facilities re-designing and tuning of the FES controller for each subject.

4.3. Model order and parameters

Previous studies usually considered muscle response to EMG as a second-order system, particularly as a critically-damped second-order system [26,27]. Our observations, as follows, confirmed suitability of this model for muscle response to surface FES. Assuming time constant of the critically-damped second-order model as α_2, the denominator of transfer function in Eq. (2), would be $1 + 2\alpha_2 S + \alpha_2^2 S^2$, and the "first-order approximation" of this denominator would be $1 + 2\alpha_2 S$. Therefore, if Eq. (1) approximates

Eq. (2), we expect $\alpha_1 \approx 2\alpha_2$ in the first-order model. Table 1 shows that α_1 is close to $2\alpha_2$, that confirms our assumption. In addition, we further ran another optimization to fit an over-damped second-order model to the response of muscles with a denominator of $1+\beta_1 S+\beta_2 S^2$ instead of $1+2\alpha_2 S+\alpha_2^2 S^2$ in Eq. (2). Our results for parameters among all subjects showed that $\beta_1 \approx 2\alpha_2$ and $\beta_2 \approx \alpha_2^2$ ($\beta_1 = 0.21 \pm 0.07$ s and $\beta_2 = 0.01 \pm 0.01$ s) that further confirmed suitability of a critically-damped second-order model for plantar-flexor muscles response to modulation of FES pulse amplitude. Previous studies used general second-order models as linear component of Hammerstein or multiple locally linear models for muscle response to FES [13–15]. Others found little difference between performances of general and critically-damped second-order models for the muscle response to FES [18]. Our observations suggest that a critically-damped second-order model can accurately represent a linear approximation for plantar-flexors response to FES pulse amplitude modulation, and there is no need to estimate additional model parameters in a general second-order model.

4.4. Comparison to previous models for muscle response to EMG or FES

In the past, researchers considered the transfer function between measured EMG and muscle force as a second-order linear model [28,29], particularly a critically-damped second-order model [26]. This model represents the chemical dynamics in the muscle fiber and the mechanical dynamics due to the sliding filament action [27,30], and showed good fit to the experimentally obtained data [27]. The muscle dynamics in response to applied FES and the measured EMG can be different. This is because muscle fibers activated by FES and natural motor commands may be different due to different activation modes [31]. Also, the natural motor command activates entire muscle, while FES via transcutaneous electrodes activates only muscle fibers underneath the electrodes. Nevertheless, our estimated time constants similar to those obtained for transfer functions between measured EMG and muscle force. The averaged time constants in a critically-damped second-order model obtained in the present study was 0.11 s. Previous studies, cited by Winter [26], reported similar time constants for the transfer functions between measured EMG and muscle force: for soleus (100 and 116 ms). Masani et al. [27] obtained time constant of 152 ms for the soleus, in response to EMG, in the voluntary standing task. Tani and Nagasaki [32] obtained time constant of 86 ms for the soleus, in response to EMG, during sitting, while they used a different experimental methodology. In comparing these times constants among studies, it should be noted that the measured values may vary among subjects, muscles, and experimental techniques [26] and between standing and sitting postures [27].

On the other hand, previous studies that used different experimental methodologies to fit a second-order model to muscle response to FES, reported a natural frequency of $\omega_n = 9.44$, 8.17, and 7.75 rad/s [15] for plantar-flexors. Our obtained natural frequencies for critically-damped second-order models ($\omega_n = 1/\alpha_2$) were close to these results. In summary, our estimated model for plantar-flexors response to FES pulse amplitude modulation confirmed the previously suggested models for plantar-flexors response to FES pulse duration modulation.

4.5. Time delay

In addition to frequency dependent phase lag (α, in Eqs. (1) and (2)), some studies considered an additional frequency-independent time delay (τ, in Eqs. (1) and (2)) of 10–50 ms for muscle response to FES [13,18,30,32,33]. In the present study, we defined τ as an independent parameter in the optimization procedure in order to minimize its effect on the estimated α (see Eqs. (1) and (2)). Our measured τ value was a summation of two time delays: (1) frequency-independent time delay of muscles in responding to FES, and (2) time delay of the stimulator device in producing FES pulses (uniformly distributed between 50 and 100 ms, experimentally measured and characterized prior to the current study). Our estimated time delays (Table 1) were in the expected range of time delay for the measurement setup. Because of the large trial-to-trial variability of the stimulator's time delay, we were not able to separate it from the muscle's time delay and thus we did not further interpret the muscle's frequency-independent time delays. Stimulators hardware with more deterministic time delay would be needed for this purpose.

4.6. Limitations

We showed the applicability of our proposed methodology to model the dynamic response of plantar-flexors to FES for able-bodied subjects and in quiet standing posture. Previous studies [16,17,22,34] that investigated transient response of muscles to FES suggested, in general, similar model structures for different musculoskeletal muscle groups and individuals affected with spinal cord injury, stroke, etc. Therefore, we expect that our proposed methodology could obtain similar model structure for other muscle groups and individuals with neuromuscular conditions. However, the model parameters can be different and should be estimated for other muscle groups and clinical populations. For example, application of our applied range of FES intensity that showed almost linear response of muscle may show significant nonlinearity in pathological condition. Further, the model parameters, especially static gain, would considerably vary among individuals and muscle groups, because of muscle strength, muscle atrophy, and other pathological conditions. Muscle fatigue and spasticity could also cause additional challenges in the experiment. Therefore, the number and duration of stimulation trials, the range of FES amplitude, and their frequencies may need to be altered for subjects with pathological condition.

Nonlinear models such as Hammerstein and locally linear models suggest that the muscle response to FES is close to linear in portions of the input range [12]. Therefore, estimating the muscle response to FES as a linear model can be an acceptable assumption for some ranges of FES pulse amplitude. In our measurement setup, the ankle torque varied almost linearly with FES amplitude between 20 and 60 mA, and thus, this range is practically suggested in the design of closed-loop FES controllers using our setup. Nevertheless, estimation of nonlinear or locally linear models for outside of this amplitude range should be further studied.

Our other studies with the same measurement setup showed that fatigue can be present after 30 s of application of constant stimulation at high intensity (e.g., pulse amplitude: 60 mA and duration: 300 µs). Our observation showed that 30 s of stimulation with pulse amplitude oscillating between 20 and 60 mA did not cause considerable muscle fatigue. The rest time between stimulation trials also contributed to preventing muscle fatigue. Nevertheless, the influence of muscle fatigue on our proposed approach should be further studied. In addition, the threshold and saturation levels are in general individual-specific and must be assessed for each subject, particularly for pathological cases.

5. Conclusions

We introduced a methodology to identify the dynamic response of plantar-flexors to FES in standing posture, as linear models. We proposed to isolate the muscles in the physiological control system, evaluate their steady-state response to harmonic modulation of FES pulse amplitude, and assess the amplitude gain and phase lag in the generated ankle torque. Both first-order and critically-damped second-order models fitted well to the experimental data. Nevertheless, the estimated time constant of the critically-damped second-order model was more consistent among subjects than the first-order model. Therefore, this model is particularly preferred for the design of neuroprostheses since it would require less individual-specific tuning. Besides, comparing the time constants of both models showed that the estimated first-order model was an approximation of the critically-damped second-order model. This latter model also agreed with those suggested for the muscle force relationship with surface EMG. Therefore, we suggest a critically-damped second-order model for plantar-flexors response to FES, either as a linear model or the linear component of more complex models.

Conflict of interest

The authors declare that there is no conflict of interest.

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