# Isometric Torque Control for Neuromuscular Electrical Stimulation With Time-Varying Input Delay

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***Abstract*— Previous results have shown experimental evidence that the muscle response to neuromuscular electrical stimulation (NMES) is delayed; the time lag is often referred to as electromechanical delay. NMES closed-loop control methods have been developed to compensate for a known constant input delay. However, as a muscle fatigues, this delay increases. This paper develops a feedback controller that robustly compensates for the time-varying delay of an uncertain muscle model during isometric contractions. The controller is proven to yield global uniformly ultimately bounded torque tracking error. Experimental results illustrate the effectiveness of the developed controller and the time-varying nature of the delayed response.**

***Index Terms*— Delay estimation, isometric contractions, muscle fatigue, neuromuscular electrical stimulation (NMES), nonlinear control, time-varying input delay.**

## I. Introduction

MOTOR neurons innervate muscle fibers and control their contractions by transmitting electrical potentials along their axons from the brain to the muscle. Motor function can be impaired if muscles are unable to receive the motor signals (e.g., following a stroke or a spinal cord injury). However, an electric current propagating along the muscle fibers between two electrodes can cause the muscle to contract [1]. Consequently, an external stimulus can replace or augment impaired motor function. Neuromuscular electrical stimulation (NMES) has been developed based on this phenomenon and functional electrical stimulation (FES) is the application of NMES to perform functional tasks. NMES is a technique primarily used in postoperative rehabilitation [2], [3] and for muscle strengthening [4]. However, NMES can also be implemented in a closed-loop feedback mechanism where the electrical stimuli are designed to achieve various rehabilitation outcomes involving dynamic or isometric contractions [5]–[14]. Many rehabilitation outcomes mandate dynamic training leading to limb motion. For other outcomes, isometric contractions are more advantageous: it is considered safer than dynamic training since the joint is not moving, develops resistance, can decrease pain during rehabilitation [15], [16] and may lower blood pressure [17], [18].

Experimental evidence exists to demonstrate that there is a time lag, termed electromechanical delay (EMD), between the muscle electrical activation and the onset of muscle force [19]–[23]. NMES-induced delays are a result of the muscle activation process; therefore, they are introduced in the dynamics via a delayed input [24], [25]. Control instability may be caused by the input delay, and therefore, EMD should be considered in the system model and in the control strategy [26]. NMES closed-loop controllers were developed in [27]–[29] to compensate for EMD assuming linear dynamics and a constant known delay. Nonlinear methods were developed in [30]–[32] to compensate for the known constant input delay assuming exact model knowledge in more general dynamic systems. The constant input delay problem for uncertain nonlinear dynamical systems is addressed in [33]–[36] where the delay is assumed to be known, and in [37] where the delay is unknown. Results of known constant EMD compensation are presented in [38] for known muscle dynamics and in [23] and [39] for uncertain muscle dynamics.

While previous research focused on NMES closed-loop stabilization in the presence of constant input delays, experimental results show evidence of fatigue during muscle contractions that limits the control performance. Muscle fatigue is a process whereby the muscle force decreases even though the stimulation signal is maintained [40]–[42] and muscle fatigue is known to occur faster with NMES training than voluntary contractions. There are various suggested causes of NMES-induced fatigue such as a reversal of Henneman's size principle [43] as well as spatially fixed and temporally synchronous fiber recruitment [44]. While fatigue itself presents a control challenge in the sense that increased control effort is required over time to generate equal torque production (motivating an integral term in the subsequently developed controller), fatigue can also lead to instability resulting from its effects on the EMD. Specifically, fatigue causes electrochemical and mechanical alterations, such as impairments of axonal action potential propagation and reduced

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motor-tendon stiffness, that lengthen the EMD [45]–[49]. While researchers have pursued methods to slow the rate of NMES-induced fatigue [50]–[53], the onset of fatigue is inevitable; therefore, the EMD should be considered as time-varying in the control development rather than constant. To compensate for the known time-varying input delay, methods were developed to stabilize input delayed systems in [54] and [55] assuming exact model knowledge and in [56] where semiglobal uniformly ultimately bounded position tracking is achieved for uncertain Euler–Lagrange dynamics.

Previous NMES results have achieved closed-loop control yielding dynamic contractions. Fewer studies have investigated closed-loop control yielding isometric contractions. In [57], a linear torque tracking controller was developed for closed-loop NMES. A group of recent results develops torque tracking controllers for isometric NMES using an electromyogram (EMG)-based fatigue prediction [58]–[60]. However, surface EMG signals are difficult to dissociate from the input stimulation signal during transcutaneous electrical stimulation. Further, EMD was not considered in the aforementioned isometric torque tracking results, although it has influence on the reaction torque and system stability.

In this paper, a torque tracking controller is developed for isometric NMES on the quadriceps femoris muscle group considering a known time-varying input delay in the dynamics; the corresponding model and assumptions are presented in Section II. The control objective is presented in Section III, and a Lyapunov-based stability analysis, developed in Section V, yields a global uniformly ultimately bounded torque tracking error despite the presence of uncertainties, nonlinearities, and time-varying EMD. Experiments were conducted on eight healthy individuals to assess the performance of the developed controller, as detailed in Section VI. Results show that the error between the measured and the desired torque remains stable while the time-varying effects of fatigue are illustrated. A discussion of the results and concluding remarks is provided in Section VII.

## II. Static Model and Properties

The uncertain nonlinear muscle model in [25] is adapted to isometric contractions by fixing the joint angle and adding the reaction torque, which leads to

$$R(t) = f(q) + D(t) + \Omega(t)V(t - \tau(t)) \tag{1}$$

where $R \in \mathbb{R}$ denotes the reaction torque, $f(q) \in \mathbb{R}$ denotes the gravity and elastic components depending only on the constant knee-joint angle $q \in \mathbb{R}$, $D \in \mathbb{R}$ denotes time-varying unknown exogenous disturbances, $\Omega \in \mathbb{R}$ is an unknown nonzero time-varying function relating the input voltage $V \in \mathbb{R}$ to the torque, and $\tau \in \mathbb{R}$ denotes the known time-varying EMD. The reaction torque is the resulting combination of the joint torque produced by electrical stimulation of the quadriceps, the gravitational and elastic effects on the system (constant in isometric conditions), and disturbances. The active knee-joint torque $T \in \mathbb{R}$, as detailed in [36], is related to the muscle-tendon force $F \in \mathbb{R}$, as

$$T = \zeta F$$

where $\zeta \in \mathbb{R}$ is a positive moment arm. The muscle-tendon force results from NMES-induced contractions, and is defined as

$$F = \psi V(t - \tau)$$

where $\psi \in \mathbb{R}$ denotes unknown muscle dynamics (e.g., unknown fatigue and muscle fiber recruitment-force properties). The control effectiveness $\Omega$ is defined as

$$\Omega = \zeta \psi. \tag{2}$$

*Assumption 1:* The disturbance $D(t)$ is bounded and its first time-derivative exists and is bounded [23].

*Assumption 2:* The positive nonzero unknown function $\Omega$ is bounded such that $\underline{\Omega} \leq \Omega(t) \leq \overline{\Omega}$ for all $t$, where $\underline{\Omega}$ and $\overline{\Omega}$ are positive constants. The first time-derivative of $\Omega$ exists and is bounded by a known constant.

*Assumption 3:* The EMD $\tau$ is bounded such that $0 < \tau(t) < \varphi_1$ for all $t$, where $\varphi_1 \in \mathbb{R}^+$ is a known constant. The rate of change of the delay is bounded such that $|\dot{\tau}| < 1-\varepsilon$ where $\varepsilon \in \mathbb{R}^+$ satisfies $0 < \varepsilon < 1$ and its second time derivative is also bounded such that $|\ddot{\tau}| < \varphi_2$ where $\varphi_2 \in \mathbb{R}^+$ is a known constant.

*Remark 1:* As a muscle fatigues, the reaction torque decays, but only to a minimum value. Assumption 2 is mild in the sense that it provides a known conservative lower bound which relates to the minimum torque that can be produced for a given input. For example, this bound could be determined experimentally.

*Remark 2:* Assumption 3 is also mild in the sense that it implies that the delay is bounded and that the change in the delay is slow process. This assumption is demonstrated by the subsequent experimental results.

Throughout this paper, for notational brevity, a time-dependent delayed function $\xi_\tau : [0, \infty) \to \mathbb{R}$ corresponding to a function $\xi$ is defined as

$$\xi_\tau(t) \triangleq \begin{cases} \xi(t - \tau(t)), & t \geq \tau(t) \\ 0, & t < \tau(t). \end{cases}$$

## III. Control Objective

The objective is to design a controller that ensures state $R$ of the input-delayed system in (1) tracks a desired torque trajectory $R_d \in \mathbb{R}$ despite uncertainties, time-varying input delays and additive bounded disturbances. To quantify this objective, the torque tracking error is defined as

$$e \triangleq R_d - R. \tag{3}$$

To facilitate the subsequent stability analysis, an auxiliary tracking error is defined as

$$r \triangleq e - Be_z \tag{4}$$

where the auxiliary signal $e_z \in \mathbb{R}$ is defined as

$$e_z \triangleq \int_{t-\tau}^{t} \dot{V}(\theta)d\theta. \tag{5}$$

In (4), $B \in \mathbb{R}$ is a constant positive best guess estimate of $\Omega$. The mismatch error between $B$ and $\Omega$ is defined as

$$\eta \triangleq B - \Omega \tag{6}$$

which, according to Assumption 2, satisfies the following inequality:

$$|\eta| \leq \overline{\eta} \tag{7}$$

where $\overline{\eta} \in \mathbb{R}$ is a known constant.

## IV. Control Development

Multiplying (4) by $\Omega^{-1}$ and using (1) and (3) yields

$$\Omega^{-1} r = \Omega^{-1}(R_d - f - D) - V_\tau - B\Omega^{-1} e_z. \tag{8}$$

The open-loop error system can be obtained by taking the time-derivative of (8) and using (4)–(6) as

$$\begin{aligned}\Omega^{-1}\dot{r} = &-\frac{1}{2}\frac{d}{dt}(\Omega^{-1})r + N + S - \dot{V}\\ &- \Omega^{-1}\eta(\dot{V} - (1-\dot{\tau})\dot{V}(t-\tau))\end{aligned} \tag{9}$$

where the auxiliary signals $S \in \mathbb{R}$ and $N \in \mathbb{R}$ are defined as

$$S \triangleq \frac{d}{dt}(\Omega^{-1}(R_d - f - D)) \tag{10}$$

$$N \triangleq -\frac{1}{2}\frac{d}{dt}(\Omega^{-1})r - B\frac{d}{dt}(\Omega^{-1})e_z. \tag{11}$$

Based on (3)–(5), the open-loop error system in (9) now contains a delay-free control input. From the subsequent analysis and (9), the control input is designed as the solution[1] to

$$\dot{V} = k_b r = k_b\left(e - BV + BV_\tau\right), \quad V(0) = V_0 \tag{12}$$

where $V_0 \in \mathbb{R}$ is a selectable constant and $k_b \in \mathbb{R}$ is a selectable constant control gain such that

$$k_b \triangleq k_{b_1} + k_{b_2} + k_{b_3} \tag{13}$$

where $k_{b_1}, k_{b_2}, k_{b_3} \in \mathbb{R}^+$. Substituting (12) into (9) yields the following closed-loop error system:

$$\begin{aligned}\Omega^{-1}\dot{r} = &-\frac{1}{2}\frac{d}{dt}(\Omega^{-1})r + N + S - k_b r\\ &- k_b\Omega^{-1}\eta(r - (1-\dot{\tau})r_\tau).\end{aligned} \tag{14}$$

Using Assumptions 1 and 2, the expressions in (10) and (11) can be upper bounded as

$$|S| \leq \varepsilon_2 \tag{15}$$

$$|N| \leq \zeta_1 ||z|| \tag{16}$$

where $\varepsilon_2, \zeta_1 \in \mathbb{R}$ are positive known constants and $z \in \mathbb{R}^2$ is defined as

$$z \triangleq [r e_z]^T. \tag{17}$$

To facilitate the subsequent stability analysis, let $y \in \mathbb{R}^4$ be defined as

$$y \triangleq \left[r e_z \sqrt{P}\sqrt{Q}\right]^T \tag{18}$$

[1]In the subsequent experiments, the control input is calculated by numerically solving the differential equation in (12) using Euler's method with a fixed step size of 1 ms. Implementation of the controller requires the force tracking error, the delay, and knowledge of the previous control input over the delay interval.

where $P, Q \in \mathbb{R}$ are Lyapunov–Krasovskii functionals defined as

$$P \triangleq \omega \int_{t-\tau}^{t}\left(\int_{s}^{t} \dot{V}^2(\theta)d\theta\right)ds \tag{19}$$

$$Q \triangleq \frac{k_b\left(2\underline{\Omega}^{-1}\overline{\eta} + k_b\gamma_2^2\right)}{2(1-\dot{\tau})}\int_{t-\tau}^{t} r^2(\theta)d\theta \tag{20}$$

where $\omega, \gamma_2 \in \mathbb{R}^+$ are selectable constants. Based on the subsequent stability analysis, the constant $\beta_1 \in \mathbb{R}^+$ is defined such that

$$\beta_1 \triangleq \min\{m_1, m_2\} \tag{21}$$

where $m_1, m_2 \in \mathbb{R}^+$ are defined as

$$m_1 \triangleq \inf_{\tau,\dot{\tau}}\left\{k_{b_3} - \frac{k_b\gamma_1^2}{4} - \frac{k_b\left(2\underline{\Omega}^{-1}\overline{\eta}(3-2\dot{\tau}) + k_b\gamma_2^2\right)}{2(1-\dot{\tau})} - k_b^2\omega\tau\right\}$$

$$m_2 \triangleq \inf_{\tau,\dot{\tau}}\left\{\frac{1}{\tau}\left(\omega(1-\dot{\tau}) - \tau\left(\frac{k_b}{\gamma_1^2} + \frac{4}{\gamma_2^2}\right) - \frac{\left(2\underline{\Omega}^{-1}\overline{\eta} + k_b\gamma_2^2\right)\varphi_2}{2k_b(1-\dot{\tau})^2}\right)\right\}$$

and $\gamma_1 \in \mathbb{R}^+$ is a selectable constant.

## V. Stability Analysis

*Theorem 3:* Given the model in (1) with Assumptions 1–3, the control law in (12) ensures global uniformly ultimately bounded torque tracking provided that the following sufficient conditions are satisfied:

$$k_{b_3} > \sup_{\tau,\dot{\tau}}\left\{k_b^2\tau\omega + \frac{k_b\left(2\underline{\Omega}^{-1}\overline{\eta}(3-2\dot{\tau}) + k_b\gamma_2^2\right)}{2(1-\dot{\tau})} + \frac{k_b\gamma_1^2}{4}\right\}$$

$$\omega > \sup_{\tau,\dot{\tau}}\left\{\frac{\tau}{1-\dot{\tau}}\left(\frac{k_b}{\gamma_1^2} + \frac{4}{\gamma_2^2}\right) + \frac{\left(2\underline{\Omega}^{-1}\overline{\eta} + k_b\gamma_2^2\right)\varphi_2}{2k_b(1-\dot{\tau})^3}\right\}$$

$$\beta_1 > \frac{\zeta_1^2}{4k_{b_1}}.$$

*Proof:* Let $V_L : \mathbb{R} \times [0;\infty) \to \mathbb{R}$ be a continuously differentiable positive-definite functional defined as

$$V_L \triangleq \frac{1}{2}\Omega^{-1}r^2 + \frac{1}{2}e_z^2 + P + Q \tag{22}$$

where $P$ and $Q$ are defined in (19) and (20), respectively, such that

$$\lambda||y||^2 \leq V_L \leq \lambda_2||y||^2 \tag{23}$$

where the constants $\lambda_1, \lambda_2 \in \mathbb{R}$ are defined as

$$\lambda_1 \triangleq \frac{1}{2}\min(\overline{\Omega}^{-1}, 1), \quad \lambda_2 \triangleq \max\left(\frac{1}{2}\underline{\Omega}^{-1}, 1\right) \tag{24}$$

and $\overline{\Omega}$ and $\underline{\Omega}$ were defined in Assumption 2. Applying the Leibniz rule to determine the time derivative of (19) and (20), and utilizing (4) and (14), the time derivative of (22) can be expressed as

$$\begin{aligned}\dot{V}_L = &-k_b r^2 + Sr + Nr - k_b\Omega^{-1}\eta r^2 + k_b e_z r\\ &+ k_b\left(\Omega^{-1}\eta r - e_z\right)(1-\dot{\tau})r_\tau + \frac{k_b\left(2\underline{\Omega}^{-1}\overline{\eta} + k_b\gamma_2^2\right)}{2(1-\dot{\tau})}r^2\\ &- \frac{k_b}{2}\left(2\underline{\Omega}^{-1}\overline{\eta} + k_b\gamma_2^2\right)r_\tau^2 - \omega(1-\dot{\tau})\int_{t-\tau}^{t}\dot{V}^2(\theta)d\theta\\ &+ \omega\tau\dot{V}^2 + \frac{k_b\left(2\underline{\Omega}^{-1}\overline{\eta} + k_b\gamma_2^2\right)\ddot{\tau}}{2(1-\dot{\tau})^2}\int_{t-\tau}^{t} r^2(\theta)d\theta.\end{aligned} \tag{25}$$

After using Young's inequality and Assumption 3, the following inequalities can be developed:

$$k_b|e_z||r| \leq k_b\left(\frac{\gamma_1^2}{4}r^2 + \frac{1}{\gamma_1^2}e_z^2\right) \tag{26}$$

$$k_b(1-\dot{\tau})|e_z||r_\tau| \leq \frac{k_b^2\gamma_2^2}{2}r_\tau^2 + \frac{2}{\gamma_2^2}e_z^2 \tag{27}$$

$$k_b\underline{\Omega}^{-1}\overline{\eta}(1-\dot{\tau})|r||r_\tau| \leq k_b\underline{\Omega}^{-1}\overline{\eta}\left(r^2 + r_\tau^2\right). \tag{28}$$

By applying the Cauchy–Schwarz Inequality

$$\begin{aligned} |e_z|^2 &= \left|\int_{t-\tau}^{t}(\dot{V}(\theta)\times 1)d\theta\right|^2 \\ &\leq \int_{t-\tau}^{t}|1|^2 d\theta \int_{t-\tau}^{t}\dot{V}^2(\theta)d\theta \leq \tau\int_{t-\tau}^{t}\dot{V}^2(\theta)d\theta. \end{aligned} \tag{29}$$

Using Assumptions 2 and 3, (7), (12), (15), (16), and (26)–(28), the expression in (25) can be upper bounded as

$$\begin{aligned} \dot{V}_L \leq{}& -k_b r^2 + \varepsilon_2|r| + \zeta_1\|z\||r| + \frac{k_b\gamma_1^2}{4}r^2 \\ &+ \left(\frac{k_b}{\gamma_1^2} + \frac{2}{\gamma_2^2}\right)e_z^2 + \frac{k_b\left(2\underline{\Omega}^{-1}\overline{\eta}(3-2\dot{\tau}) + k_b\gamma_2^2\right)}{2(1-\dot{\tau})}r^2 \\ &- \omega(1-\dot{\tau})\int_{t-\tau}^{t}\dot{V}^2(\theta)d\theta + k_b^2\omega\tau r^2 \\ &+ \frac{\left(2\underline{\Omega}^{-1}\overline{\eta} + k_b\gamma_2^2\right)\varphi_2}{2k_b(1-\dot{\tau})^2}\int_{t-\tau}^{t}\dot{V}^2(\theta)d\theta. \end{aligned} \tag{30}$$

Using (13) and (29), (30) is upper bounded and grouped as

$$\begin{aligned} \dot{V}_L \leq{}& -k_{b_1}r^2 + \zeta_1\|z\||r| - k_{b_2}r^2 + \varepsilon_2|r| \\ &- \left(k_{b_3} - \frac{k_b\gamma_1^2}{4} - \frac{k_b\left(2\underline{\Omega}^{-1}\overline{\eta}(3-2\dot{\tau}) + k_b\gamma_2^2\right)}{2(1-\dot{\tau})} - k_b^2\omega\tau\right)r^2 \\ &- \left(\omega(1-\dot{\tau}) - \frac{\left(2\underline{\Omega}^{-1}\overline{\eta} + k_b\gamma_2^2\right)\varphi_2}{2k_b(1-\dot{\tau})^2}\right)\int_{t-\tau}^{t}\dot{V}^2(\theta)d\theta \\ &+ \tau\left(\frac{k_b}{\gamma_1^2} + \frac{4}{\gamma_2^2}\right)\int_{t-\tau}^{t}\dot{V}^2(\theta)d\theta - \frac{2\tau}{\gamma_2^2}\int_{t-\tau}^{t}\dot{V}^2(\theta)d\theta. \end{aligned} \tag{31}$$

Completing the squares in (31) and using (29) yields

$$\begin{aligned} &\dot{V}_L \\ &\leq -\left(k_{b_3} - \frac{k_b\gamma_1^2}{4} - \frac{k_b\left(2\underline{\Omega}^{-1}\overline{\eta}(3-2\dot{\tau}) + k_b\gamma_2^2\right)}{2(1-\dot{\tau})} - k_b^2\omega\tau\right)r^2 \\ &\quad - \frac{1}{\tau}\left(\omega(1-\dot{\tau}) - \tau\left(\frac{k_b}{\gamma_1^2} + \frac{4}{\gamma_2^2}\right) - \frac{\left(2\underline{\Omega}^{-1}\overline{\eta} + k_b\gamma_2^2\right)\varphi_2}{2k_b(1-\dot{\tau})^2}\right)e_z^2 \\ &\quad - \frac{2\tau}{\gamma_2^2}\int_{t-\tau}^{t}\dot{V}^2(\theta)d\theta + \frac{\zeta_1^2}{4k_{b_1}}\|z\|^2 + \frac{\varepsilon_2^2}{4k_{b_2}}. \end{aligned} \tag{32}$$

Using the definition of $\beta_1$ in (21) and $z$ in (17), the expression in (32) is upper bounded as

$$\dot{V}_L \leq -\left(\beta_1 - \frac{\zeta_1^2}{4k_{b_1}}\right)\|z\|^2 + \frac{\varepsilon_2^2}{4k_{b_2}} - \frac{2\tau}{\gamma_2^2}\int_{t-\tau}^{t}\dot{V}^2(\theta)d\theta. \tag{33}$$

After using (12), (19), (20), and the following inequality:

$$\begin{aligned} \int_{t-\tau}^{t}\int_{s}^{t}\dot{V}^2(\theta)d\theta ds &\leq \int_{t-\tau}^{t}\sup_{s\in[t-\tau,t]}\int_{s}^{t}\dot{V}^2(\theta)d\theta ds \\ &= \tau\sup_{s\in[t-\tau,t]}\int_{s}^{t}\dot{V}^2(\theta)d\theta \\ &= \tau\int_{t-\tau}^{t}\dot{V}^2(\theta)d\theta \end{aligned} \tag{34}$$

the expression in (33) can be bounded as

$$\begin{aligned} \dot{V}_L \leq{}& -\left(\beta_1 - \frac{\zeta_1^2}{4k_{b_1}}\right)\|z\|^2 + \frac{\varepsilon_2^2}{4k_{b_2}} - \frac{1}{\gamma_2^2\omega}P \\ &- \frac{2\tau k_b(1-\dot{\tau})}{\gamma_2^2\left(2\underline{\Omega}^{-1}\overline{\eta} + k_b\gamma_2^2\right)}Q. \end{aligned} \tag{35}$$

From the definition of $y$ in (18), (35) can be upper bound as

$$\dot{V}_L \leq -\beta_2||y||^2 + \frac{\varepsilon_2^2}{4k_{b_2}} \tag{36}$$

where $\beta_2 \in \mathbb{R}^+$ is defined as

$$\beta_2 \triangleq \min\left\{\beta_1 - \frac{\zeta_1^2}{4k_{b_1}}, \frac{1}{\gamma_2^2\omega}, \quad \inf_{\tau,\dot{\tau}}\left\{\frac{2\tau k_b(1-\dot{\tau})}{\gamma_2^2\left(2\underline{\Omega}^{-1}\overline{\eta} + k_b\gamma_2^2\right)}\right\}\right\}.$$

Using (23), the expression in (36) can be written as

$$\dot{V}_L \leq -\frac{\beta_2}{\lambda_2}V_L + \frac{\varepsilon_2^2}{4k_{b_2}}. \tag{37}$$

Finally, the differential equation in (37) can be solved as

$$V_L \leq V_L(0)\exp\left(-\frac{\beta_2}{\lambda_2}t\right) + \frac{\varepsilon_2^2\lambda_2}{4k_{b_2}\beta_2}\left(1 - \exp\left(-\frac{\beta_2}{\lambda_2}t\right)\right). \tag{38}$$

From (38), $V_L$ is globally uniformly ultimately bounded. Using (22), $r$ and $e_z$ are also bounded such that

$$|r| \leq \sqrt{2\overline{\Omega}V_L(0)\exp\left(-\frac{\beta_2}{\lambda_2}t\right) + \frac{\overline{\Omega}\varepsilon_2^2\lambda_2}{2k_{b_2}\beta_2}\left(1 - \exp\left(-\frac{\beta_2}{\lambda_2}t\right)\right)} \tag{39}$$

$$|e_z| \leq \sqrt{2V_L(0)\exp\left(-\frac{\beta_2}{\lambda_2}t\right) + \frac{\varepsilon_2^2\lambda_2}{2k_{b_2}\beta_2}\left(1 - \exp\left(-\frac{\beta_2}{\lambda_2}t\right)\right)}. \tag{40}$$

The expressions (4), (39), and (40) can be used to conclude that $|e|$ is globally uniformly ultimately bounded. Finally, from (1) and (3), the control input $V$ is bounded. ■

## VI. EXPERIMENTS

Experiments were conducted to examine the performance of the controller developed in (12). Surface electrical stimulation was applied to the quadriceps muscle group to produce an isometric torque about the knee joint. The torque produced at the knee joint was measured by a force transducer, and the EMD was approximated as the time lag between the control signal onset and the torque onset where the estimated values were used in the control input. In addition, the experiments examined the time-varying aspect of the EMD.

Eight healthy subjects (age 25.8 ± 3.2 years) participated in the trials after giving written informed consent, as approved

TABLE I
RMS ERRORS, EMD, AND CONTROL EFFECTIVENESS ESTIMATES DURING NMES-INDUCED TORQUE TRACKING TRIALS

| Subject-Trial | RMS Error (N·m) | Delay (ms) Mean | Minimum | Maximum | B (N·m·$\mu$s$^{-1}$) |
|---|---|---|---|---|---|
| 1-a | 1.810 | 101.0 | 80.3 | 116.5 | 0.3732 |
| 1-b | 1.694 | 102.0 | 87.1 | 130.9 | 0.2997 |
| 2-a | 1.448 | 99.5 | 80.5 | 116.6 | 0.7353 |
| 2-b | 1.324 | 93.7 | 79.9 | 103.8 | 0.6251 |
| 3-a | 1.439 | 82.4 | 74.5 | 91.4 | 0.4948 |
| 3-b | 1.039 | 77.7 | 68.9 | 85.1 | 0.3738 |
| 4-a | 1.167 | 90.2 | 58.4 | 123.9 | 0.3191 |
| 4-b | 1.110 | 116.2 | 88.8 | 145.1 | 0.2459 |
| 5-a | 1.830 | 100.3 | 82.6 | 125.1 | 0.8248 |
| 5-b | 1.413 | 93.5 | 78.0 | 107.6 | 0.6362 |
| 6-a | 1.612 | 79.6 | 68.8 | 98.7 | 0.6534 |
| 6-b | 1.127 | 79.3 | 66.8 | 89.9 | 0.3509 |
| 7-a | 1.420 | 74.5 | 66.3 | 83.4 | 0.5738 |
| 7-b | 1.422 | 91.7 | 73.3 | 110.5 | 0.4567 |
| 8-a | 1.722 | 101.0 | 58.8 | 146.0 | 0.4595 |
| 8-b | 1.557 | 99.0 | 72.9 | 124.8 | 0.5104 |
| **Mean** | **1.446** | **92.6** | **74.1** | **112.5** | **0.4958** |
| **SD** | **0.249** | **11.4** | **9.1** | **19.7** | **0.1671** |

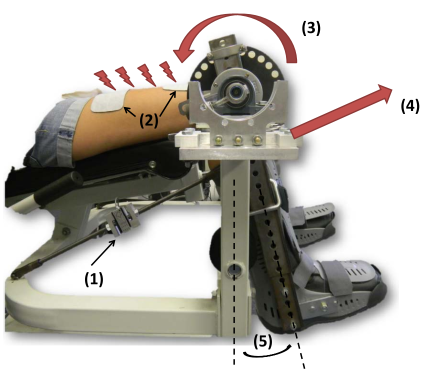


Fig. 1. Experimental apparatus in isometric contractions. (1) Force transducer. (2) Surface electrodes. (3) Torque produced at the knee joint. (4) Force measured by the force transducer. (5) Constant knee-joint angle.

by the Institutional Review Board at the University of Florida. Participants were asked to sit in a modified leg extension machine (LEM). Fig. 1 illustrates the experimental setup used to measure the force while the stimulus was delivered. Using the measured moment arm length, the force exerted on the LEM from the lower shank was converted into a torque and used in the controller. The developed controller was tested on each participant's right leg twice: the torque tracking exercise lasted 2 min and was repeated after 15 min of rest. A current-controlledstimulator (RehaStim, Hasomed GmbH, Germany) was used to deliver a modulated stimulation pattern to each participant's quadriceps femoris muscle group via bipolar surface electrodes while the participant was asked to remain passive. The utilized electrodes were $3'' \times 5''$ PALS® electrodes, provided as compliments by Axelgaard Manufacturing Company Ltd. The electrical stimulation pattern was composed of pulses with a constant pulse frequency of 30 Hz and a constant pulse amplitude of 90 mA, while the pulsewidth was varied according to (12). Since the stimulator used in the experiments has a greater resolution in pulsewidth (20–500 $\mu$s in steps of 1 $\mu$s) than pulse amplitude (0–126 mA in steps of 2 mA) and since the uncertain model in (1) is equivalent regardless of which stimulation parameter is varied, the control input in (12) was implemented as a pulsewidth modulated input, without loss of generality. The desired torque profile was selected to be smooth and periodic with high (fatiguing) and low (resting) plateaus.

Implementation of the controller in (12) requires the estimation of two parameters. The first parameter is the control effectiveness $\Omega$ introduced in (2) that relates the input stimulus to the torque about the knee joint. As described in (6), $\Omega$ is approximated by a constant $B$ from the measured recruitment curve of the muscle, obtained in a pretrial test: $B$ was computed as the linear slope of the recruitment curve which varied with electrode placement and each individual's strength. The control effectiveness estimate was evaluated before each trial (two per participant).

The second parameter is the EMD. An algorithm was designed to compute an estimate for the delay in real-time, based on the definition of EMD (i.e., the time difference between the onset of electrical activity and the onset of torque production) and the study in [61]. The EMD was computed based on the following algorithm.

1) The input pulsewidth and measured torque data were buffered for one period of the desired trajectory.
2) The two vectors were normalized.
3) The cross correlation between the two normalized vectors was calculated.
4) The index that maximized the cross correlation was converted into seconds to obtain the time delay between the two signals.

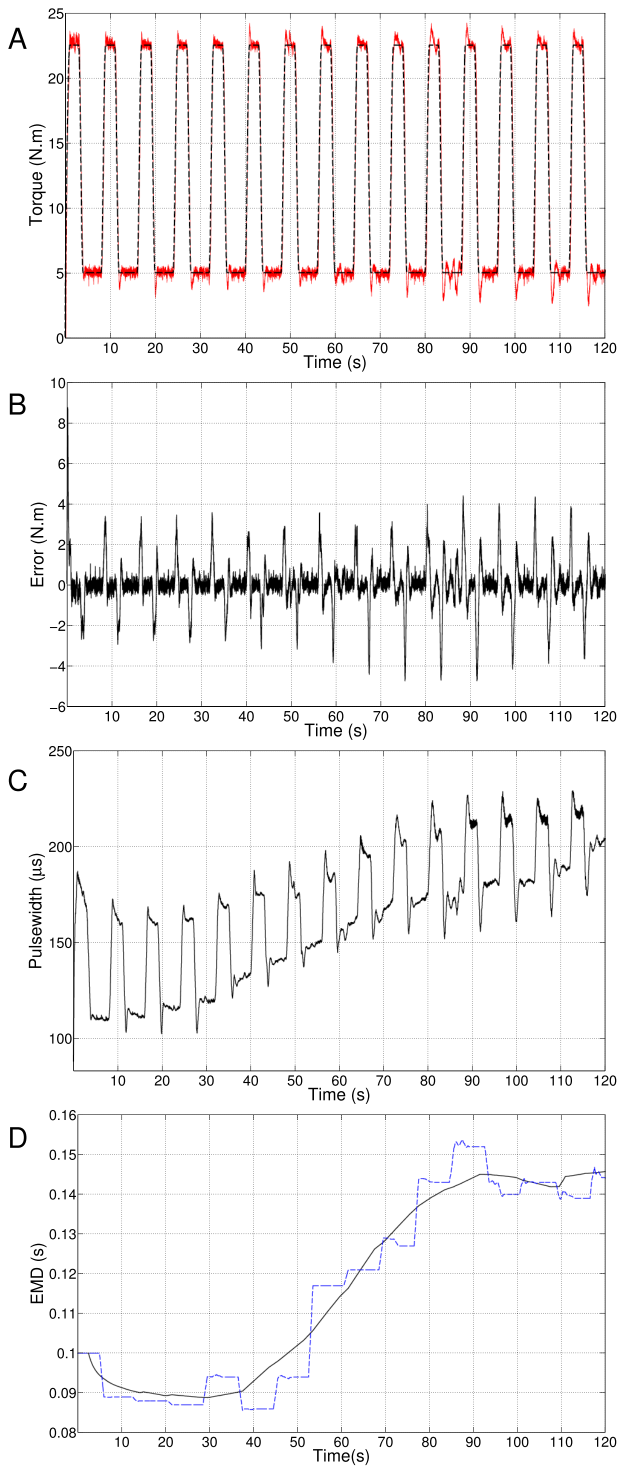


Fig. 2. (a) Desired (dashed line) and measured torque (solid line) during NMES-induced torque tracking trial on Subject 4. (b) Corresponding tracking error. (c) Control input. (d) Averaged EMD with a 30-s moving window (solid line, off-line computation) and estimated time-varying EMD (dashed line, online computation).

The tracking performance for Subject 4 is depicted in Fig. 2, which includes the error between the desired and the measured torque, the control input, and the estimated delay during the trial. The rms errors, EMD measurements, and control effectiveness estimates are provided in Table I. The EMD varied in every trial, as depicted in Fig. 3; the delay variations in Table I are calculated as the variations between the lowest and the highest value for one trial.

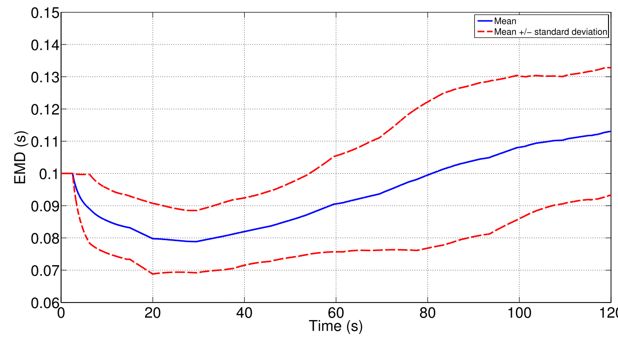


Fig. 3. The solid line represents the mean EMD values for all trials, and the dashed lines correspond to the mean value + or − the standard deviation. The EMD calculations were all initialized at 0.1 s, computed online with a cross-correlation method, and averaged offline with a 30-s moving window.

## VII. Discussion

The tracking error remained stable[2] with an overall rms error of 1.45 N $\cdot$ m in the presence of uncertain parameters and increasing input delays. Although EMD was assumed to be known and continuous, the experiments proved that torque tracking was possible despite the fact that the delay was estimated rather than exactly known. The minimum EMD over all trials was 74.1 ms (SD 9.1 ms). Vos *et al.* [61] found a mean delay of 86 ms for nonfatiguing voluntary contractions of the vastus lateralis muscle, supporting the delay estimated during the experiments in fatigue conditions. However, different results can be found in the literature, for instance, 8.5 ms in [46], 17.2 ms in [21], or 27.5 ms in [48]; these discrepancies may be due to different measurement methods. The delay estimation performed during the torque tracking indicates that the delay increases with fatigue (52% increase on average); Häkkinen and Komi [62] and Zhou *et al.* [45] found similar results with EMD increasing 29% and 45%, respectively, after a fatiguing isometric knee extension. Considering such a variability in the values, the EMD should be considered in the control method. Moreover, an overall decrease in the control effectiveness approximation and an increase in the general shape of the control input [Fig. 2(c)] illustrated the effect of fatigue on the muscle characteristics.

This paper provides a Lyapunov-based stability proof for torque tracking in the presence of a known time-varying input delay and exogenous disturbance where experiments were conducted to evaluate the controller's performance. This control method accounts for increasing fatigue in the muscles and may enable longer FES exercises. The controller designed in this study does not require exact model knowledge of the muscle parameters, facilitating clinical practice. However, the EMD needs to be known and the control effectiveness needs to be estimated; therefore future efforts are focused on compensating for unknown delays in the control strategy. Future work will also seek to develop a measurable time-varying estimate of the control effectiveness to yield improved tracking performance.

[2]The tracking error remained bounded within a small neighborhood of the origin. Although the peak-to-peak error in Fig. 2(b) increased after 60 s, the stability analysis shows that the gain conditions are more difficult to satisfy for longer delays. Therefore, increased EMD may explain the increased tracking error after 60 s.

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