Characterization of the Time-Varying Nature of Electromechanical Delay During FES-Cycling

Brendon C. Allen, *Graduate Student Member, IEEE*,
Kimberly J. Stubbs, *Graduate Student Member, IEEE*,
and Warren E. Dixon, *Fellow, IEEE*

***Abstract*— Functional electrical stimulation (FES) induced cycling is a common rehabilitative technique for people with neuromuscular disorders. A challenge for closed-loop FES control is that there exists a potentially destabilizing time-varying input delay, termed electromechanical delay (EMD), between the application of the electric field and the corresponding muscle contraction. In this article, the FES-induced torque production and EMD are quantified on an FES-cycle for the quadriceps femoris and gluteal muscle groups. Experiments were performed on five able-bodied individuals and five individuals with neurological conditions. Closed-loop FES-cycling was applied to induce fatigue and torque and EMD measurements were made during isometric conditions before and after each minute of cycling to quantify the effect of fatigue on EMD and torque production. A multiple linear regression and other descriptive statistics were performed to establish a range of expected EMD values and bounds on the rate of change of the EMD across a diverse population. The results from these experiments can be used to assist in the development of closed-loop controllers for FES-cycling that are robust to time-varying EMD and changes in torque production.**

***Index Terms*— Functional electrical stimulation (FES), electromechanical delay (EMD), time-varying delay, FES-cycling, human-robot interaction.**

I. Introduction

FUNCTIONAL electrical stimulation (FES) involves the application of an electric field to induce muscle contractions yielding functional tasks (e.g., walking [1], [2] or cycling [3]–[7]). FES-cycling is a common rehabilitative exercise for those with neurological disorders such as stroke, Parkinson's Disease, Cerebral Palsy, Multiple Sclerosis, etc., [3]–[8], because FES-cycling has been shown to have numerous health benefits, such as improved cardiovascular parameters, increased bone mineral density and muscle mass, among other benefits [9]–[11]. Closed-loop FES control is an active and challenging research domain focused on improving rehabilitation devices and strategies.

Manuscript received January 23, 2020; revised May 28, 2020 and July 20, 2020; accepted August 12, 2020. Date of publication August 17, 2020; date of current version October 8, 2020. This work was supported in part by the NSF under Award 1762829 and in part by the NDSEG Fellowship Program. *(Corresponding author: Brendon C. Allen.)*

The authors are with the Department of Mechanical and Aerospace Engineering, University of Florida, Gainesville, FL 32611-6250 USA (e-mail: brendoncallen@ufl.edu; kimberlyjstubbs@ufl.edu; wdixon@ufl.edu).

Digital Object Identifier 10.1109/TNSRE.2020.3017444

A challenge of closed-loop FES control is that there exists a complex electro-physiological mechanism involved in force production in response to electrical stimulation. A result of this complex energy conversion process is that there exists an input delay between the application of an electric field and the onset of force production, (i.e., an electromechanical delay (EMD)) [12]–[14]. Often in literature the EMD corresponds to the time latency between the onset of EMG activity and muscle force [15], however, in this article we refer to the EMD in a broader sense as the time latency between the application of stimulation and the corresponding torque, such as the EMD is defined in [5]–[7], [13], [16]–[19]. A focus of prior works has been to examine the underlying physiological factors for the latency between electrical input and force output [20]–[22], but in this article the EMD is considered at the macro level, motivated by the desire to compensate for the phenomenological effects within a closed-loop control structure, and the underlying physiological factors are beyond the scope.

EMD can potentially destabilize a control system such as FES-cycling (e.g., the cadence tracking error is not contained in a bounded set). To prevent delay-induced instability, EMD needs to be included in the dynamic model that is used in the stability analysis of the closed-loop system. Initial control efforts to compensate for the EMD modeled the EMD as a constant [18], [19], [23]. Delay-compensation methods have since been developed that allow for the EMD to be unknown and time-varying, such as in [5], [6], [16], [17]; however, in these studies the EMD is estimated by a constant. A constant estimate is not ideal because EMD has been shown to change due to FES-induced fatigue and a more accurate estimate will improve performance [13]. A preliminary result by the authors' has been developed to compensate for an unknown time-varying delay by using a time-varying estimate of the delay [7]. Each of the prior results require for certain aspects of the EMD to be known. For example, often the EMD is assumed to be bounded by a known lower and upper bound. However, all previous studies to understand the time-varying effects of FES-induced fatigue on torque production and EMD have focused on simple single joint (e.g., knee extension [13], [19], [24], [25]) tasks, and the effects of FES-induced fatigue during more complex tasks that involve multiple muscle groups (e.g., cycling) remains unclear. In fact, it is unclear if closed-loop control during motorized FES-cycling

produces enough fatigue to cause the EMD to vary and bounds on the EMD are unknown. Additionally, the more complex interaction and timing of multiple muscle groups involved in FES-cycling requires both the contraction delay (CD) and residual delay (RD) to be considered [5]–[7], where the CD is the time latency between the start of stimulation and the onset of torque and the RD is the time latency between the end of stimulation and the cessation of torque. An increased understanding of the CD and RD will allow closed-loop controllers to determine when to apply/cease stimulation to reduce muscle contractions in antagonistic muscles [5]–[7].

The objective of this article is to test the hypothesis that FES-induced cycling will induce sufficient fatigue such that the EMD and torque about the cycle crank axis will vary with cycling time and to then establish bounds on the torque and EMD and on the rate of change of the torque and EMD. To provide additional information for the control designer, two types of EMD were considered, the CD and RD, and both the CD and RD were measured in three different ways (see Fig. 2). To test the hypothesis, experiments were performed on five able-bodied individuals and five individuals with neurological conditions (NCs). The experiments consisted of 10 minutes of FES-cycling. Before and after each minute of cycling the motor fixed the crank at desired angles to create isometric conditions. Transcutaneous electrical stimulation was delivered to a combination of the quadriceps femoris and gluteal muscle groups in these isometric conditions. The resultant torque data was then examined to determine information about the torque and EMD. A multiple linear regression was performed on the data and the result provides evidence that FES-cycling does result in fatigue and that the EMD is time-varying. Figures were constructed to show how the torque and EMD vary with cycling time. The results in this article can be used to improve the future development of closed-loop controllers for FES-cycling that are robust to time-varying input delays.

II. Methods

Transcutaneous electrical stimulation was applied to the quadriceps femoris and the gluteal muscle groups and the resulting crank arm torque was recorded during isometric conditions to examine the torque production and EMD. During dynamic conditions the recorded torque measurements are a complex function of the leg and muscle dynamics, disturbances such as volitional movement, motor contribution, FES-induced muscle contribution, and the muscle effectiveness across various angles and velocities. Since the FES-induced muscle contribution cannot be extracted from dynamic torque measurements the EMD cannot be measured during dynamic conditions; thus, isometric conditions are utilized in this article. The current amplitude (90 mA for the quadriceps and 70 mA for the gluteals) and stimulation frequency (60 Hz) were fixed while the pulse width was used as the control input.[1] When recording the torque, the motor held the crank at a pre-specified angle to create isometric conditions and then the pulse width was varied in an open-loop manner (i.e., the stimulation pattern was predetermined) to induce muscle contractions. The pulse width pattern was designed to enable repeated EMD and torque measurements throughout the experiment. To fatigue the muscle, FES-induced cycling was implemented in one-minute intervals between torque measurements.

TABLE I
Participant Demographics

Participant	Age	Sex	Condition	Time Since Diagnosis
S1	27	M	None	- -
S2	28	M	None	- -
S3	22	F	None	- -
S4	21	M	None	- -
S5	23	M	None	- -
N1	26	M	Spina Bifida (L5-S1)	26yr
N2	57	F	Multiple Sclerosis	10yr
N3	42	F	Cerebral Palsy	42yr
N4	34	F	Multiple Sclerosis	5yr
N5	64	F	Multiple Sclerosis	23yr

A. Subjects

Five able-bodied individuals and five individuals with NCs, whose demographics are listed in Table I, participated in the study. Since an objective of this study is to characterize and establish bounds on the EMD and the torque about the cycle crank axis to inform the development of closed-loop controllers, participants with and without NCs and with varied demographics were recruited. However, to investigate the EMD for specific NCs including differences in levels of severity, clinical trials would need to be pursued to yield a larger data set. Able-bodied participants are referred to by the letter "S" followed by their participant number, while participants with NCs are referred to by the letter "N" followed by their participant number. Prior to participation, written informed consent was obtained from each participant, as approved by the University of Florida Institutional Review Board (IRB201901676).

B. Apparatus

The experimental testbed was created by modifying an existing recumbent tricycle (TerraTrike Rover) to include actuators and sensors. Orthotic boots (Össur Rebound Air Tall) were used to couple the rider to the cycle, to securely constrain the ankles, and to maintain sagittal alignment of the legs. A trainer and rider rings were used to offset the cycle from the ground. The original bike crank was replaced with a SRM Science Road Powermeter crankset to measure the torque. A US Digital H1 encoder was mounted to the cycle and attached to the crank via spur gears to measure the position and cadence. A 250 W motor (Unite Motor Co. Ltd. MY1016Z2) was coupled to the drive train and actuated using a current-controlled Advanced Motion Controls[2] (AMC) AB25A100 motor driver and an AMC PS300W24 power

[1]These current amplitudes and stimulation frequency were selected based on prior literature [4].

[2]ADVANCED Motion Controls supported the development of this testbed by providing discounts on their branded items.

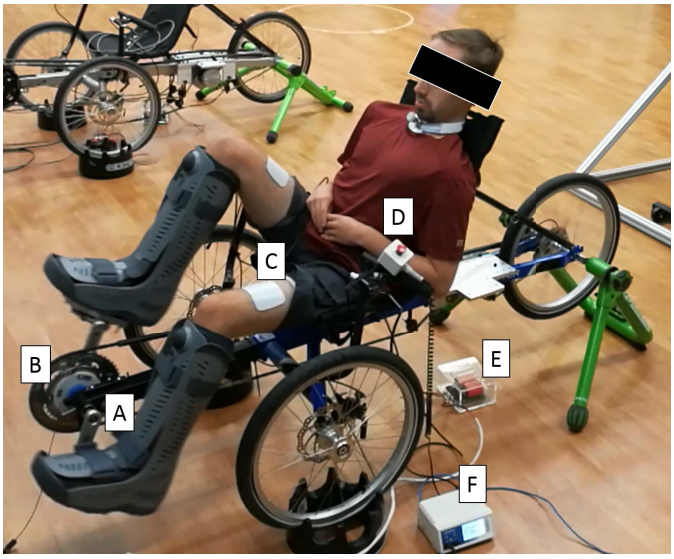

Fig. 1. Motorized FES cycle: (A) Encoder (B) Power Meter (C) Electrodes (D) E-Stop (E) Filter Card (F) Stimulator [8].

supply, and an AMC FC15030 filter card was added in-line with the motor. A current-controlled, 8-channel RehaStim HASOMED stimulator (operating in science mode) was used to deliver symmetric, biphasic, and rectangular pulses via self-adhesive electrodes (Axelgaard ValuTrode CF7515).[3] For safety, an emergency stop switch was mounted to the cycle's handle to allow the rider to halt the experiment if required. The powermeter, encoder, motor, and stimulator were interfaced with a desktop computer running MATLAB/Simulink/Quarc through a Quanser Q-PIDe data acquisition board at 500 Hz. The motorized FES cycle with a rider is depicted in Fig. 1.

C. Experimental Protocol

Prior to the experiment, electrodes were placed medial-distal and lateral-proximal over the quadriceps femoris muscle and over the proximal and distal components of the gluteal muscle group in accordance with the Axelgaard electrode placement manual.[4] The participant was then seated in the recumbent tricycle with their legs constrained using orthotic boots. Next, a participant specific angle was determined for efficient stimulation of both their left (right) quadriceps and left (right) gluteal muscle groups called the left (right) angle, denoted by $q_L \in \mathbb{R}$ ($q_R \in \mathbb{R}$). To allow for a comparison between participants, q_L and q_R were selected using the torque transfer ratios from [4], denoted by $T_m : \mathcal{Q} \to \mathbb{R}$, where $m \in \mathcal{M} \triangleq \{RQ, RG, LQ, LG\}$ indicates the right (R) and left (L) quadriceps femoris (Q) and gluteal (G) muscle groups and the set of all possible crank angles is denoted by $\mathcal{Q} \subseteq \mathbb{R}$. The left and right angle were selected as

$$q_* \triangleq \left\{ q \in \mathcal{Q} \,|\, T_*(q) = \max(T_*) \,\&\, T_{*Q}(q), T_{*G}(q) > 0 \right\},$$

where the $*$ can be replaced by R or L to create distinct expressions, q denotes the crank angle, and where

$$T_*(q) \triangleq \sqrt{\left(\frac{T_{*Q}}{\max(T_{*Q})}\right)^2 + \left(\frac{T_{*G}}{\max(T_{*G})}\right)^2}.$$

Lastly, a proportional–integral–derivative (PID) controller was used to fix the crank at the left (right) angle to create isometric conditions and comfort limits on the pulse width, called the comfort threshold, were determined for the left (right) leg's muscle groups. The participant was instructed to be a passive rider and to provide no volitional effort and no practice was allowed.

The purpose of this study is to understand the effect of fatigue on the EMD and torque production of two muscle groups in response to FES-induced cycling. In previous studies, the gluteal muscle group is often stimulated only when the quadriceps femoris group is also being stimulated [4]–[6], [8]. Therefore, in this study the EMD and torque production are examined for two muscle combinations: quadriceps only, and quadriceps and gluteal together.

The experimental protocol has two components: the measurement sequence and the cycling sequence. For the measurement sequence, the motor randomly fixed the crank at the right or left angle, followed by fixing the crank at the other angle. When at the right (left) angle, 0.25 s of stimulation, at each muscle's comfort threshold, was applied in a random sequence to the RQ, RQRG (LQ, or LQLG) muscle groups with a 2 s rest period provided between each bout of stimulation. The cycling sequence was 80 seconds and the first 20 seconds consisted of the motor tracking a smooth cadence ramp from 0 to 50 RPM, at which point the closed-loop FES controller from [4] was implemented for a one minute duration of FES-cycling. For added comfort, the maximum allowed stimulation for each muscle, during the cycling sequence, was set between 80% to 90% of each muscle's comfort threshold based on user comfort. The experimental protocol consisted of an initial measurement sequence and thereafter a combination of a cycling sequence followed (after a brief cool down of 5 s) by a measurement sequence repeated ten times for a total of ten minutes of cycling.

D. Precautions

Since an aim of the article was to characterize the effect of fatigue on the EMD and torque production, the experiments were only performed if the participant reported that their muscles were adequately rested (i.e., no sore muscles from previous exercises or activities). Additionally, an aim was to understand the effect of fatigue in various muscle groups in both legs. However, only one leg and one muscle group combination can be tested at a time. Therefore, during the measurement sequence, the order in which the leg and muscle groups were stimulated was randomized. Randomization and consistent timing were each managed automatically through the feedback controller software.

Due to the non-selective nature of FES [26], [27], fatigue should be similar across intensity levels. Therefore, the comfort threshold for each muscle was used to set the pulse width

[3]Surface electrodes for this study were provided compliments of Axelgaard Manufacturing Co., Ltd.

[4]If desired, images of electrode placement can be found at https://www.axelgaard.com/Education/Knee-Extension and https://www.axelgaard.com/Education/Hip-Extension

for each participant. During the cycling portion, 80% to 90% of the comfort threshold was used as an upper limit of the stimulation input in each muscle. These thresholds ensure comfort for the participant while simultaneously producing strong contractions from each muscle group. To provide additional safety, an emergency stop button was provided to halt the experiment if required.

E. Measurements

FES inputs (pulse width) and the resulting torque output were recorded with a sampling frequency of 500 Hz. To reduce noise in the torque data, a 2nd order Butterworth IIR low-pass filter with a half power frequency of 8 Hz was implemented using the MATLAB functions designfilt and filtfilt to forward and reverse filter the torque data so that the filter would not introduce a delay.

The pulse width and torque data were segmented such that each segment contained 0.25 s of stimulation and its associated torque response. Each segment included 1 s of data from the moment the 0.25 s of stimulation began. The torque response in each segment contained 3 distinct regions: a pre-contraction region called the initial torque baseline, a region that represented the muscle contraction, and once the contraction ceased, a post-contraction region called the post torque baseline. The torque data of each segment was shifted so that the average torque of the initial torque baseline was 0 to remove the inertia effects of the leg pushing against the torque sensor. A plot of a single segment after being shifted is shown in Fig. 2.

1) Torque: The peak and average torques were measured to determine the effect of fatigue and the effect of crank angle on the FES-induced torque. The peak torque (T_{max}) and the average torque (T_{avg}) are defined as the maximum and average value of the resultant torque after 0.25 s of stimulation, respectively.

2) Delay: CD was measured in three ways: the initial CD (CD0), the CD to reach 25% of the peak torque (CD25), and the CD to reach 75% of the peak torque (CD75). CD0 is the time difference between when the first electrical pulse was delivered to the muscle and the time the torque increased to 0.05 Nm above the initial torque baseline. CD25 (CD75) is the time difference between when the first electrical pulse was delivered to the muscle and the time the torque increased to 25% (75%) of the peak torque value. RD was measured in three ways: the initial residual delay (RD0), the RD to decay to 25% of the peak torque (RD25), and the RD to decay to 75% of the peak torque (RD75). RD0 is the time the difference in time between when the last electrical pulse was delivered to the muscle and the time that the torque fell to 0.05 Nm below the peak torque. RD25 (RD75) is the time difference between when the last electrical pulse was delivered to the muscle and the time that the torque fell to 25% (75%) of the difference between the peak torque value and the post torque baseline. Fig. 2 illustrates the different delay measurements.

F. Statistical Analysis

To quantify the effect of fatigue on FES-cycling a series of multiple linear regressions were performed using the fitlm function in MATLAB. For each regression the dependent variable was selected as one of the measurements (T_{max}, T_{avg}, CD0, CD25, CD75, RD0, RD25, and RD75). Each of the regression analyses used the following predictors (independent variables): number of minutes spent cycling (CycleTime; quantitative predictor ranging from 0 to 10), leg dominance[5] (Side; Non-dominant or Dominant), if the gluteal muscle group was stimulated (Muscle; No Glute or Glute), and the individual being tested (Subject; S1, …, S5, N1,.., N5). To improve the model, the following quadratic and interaction terms were included in all the regressions: CycleTime^2, Side × Muscle, Side × Subject, Muscle × Subject, and CycleTime × Subject. The interactions CycleTime × Side and CycleTime × Muscle were initially included, however they were subsequently removed because they did not have a significant effect (P-value > 0.05) for any of the regressions. The model for each regression included each independent variable, the quadratic term, and the aforementioned interactions. To provide additional information about how the delay varies with time the quantitative predictors CD0, CD25, RD0, and RD75 were included in the regression on CD25, CD75, RD75, and RD25, respectively. To assess goodness of each model the adjusted R^2 was utilized.

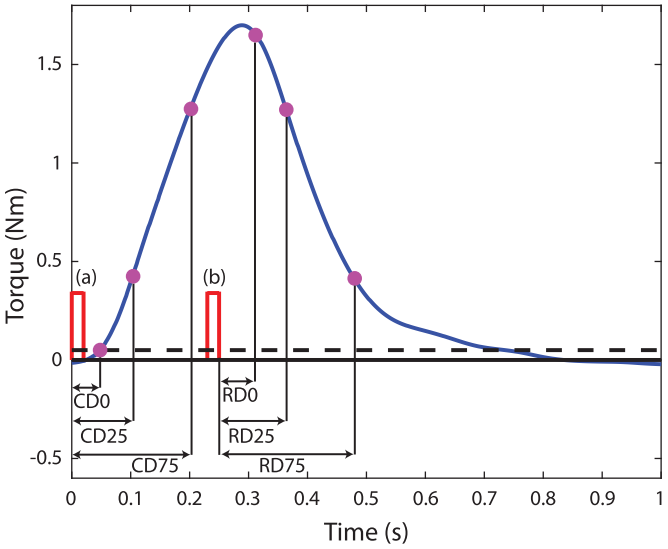

Fig. 2. Schematic illustration to depict the six EMD measurements, where (a) represents the initial pulse of a pulse train and (b) denotes the final pulse. The first and last pulse of the pulse train are shown to represent timing information for 0.25 s of stimulation where the height and width is arbitrarily drawn. The EMD measurements are the initial contraction delay (CD0), the contraction delay to reach 25% of the peak torque (CD25), the contraction delay to reach 75% of the peak torque (CD75), the initial residual delay to decay to 0.05 Nm below the peak torque (RD0), the residual delay to decay to 25% of the peak torque (RD25), and the residual delay to decay to 75% of the peak torque (RD75). The dashed black line indicates a torque threshold of 0.05 Nm.

1) Interpretation: The statistical significance of the CycleTime, CycleTime×Subject, and CycleTime^2 predictor coefficients was used to infer the effect of FES-cycling induced

[5]Each participant was asked to self-identify their dominant leg. If they were uncertain they were asked, "which leg would you use to kick a ball?" to identify their dominant leg [28].

fatigue on each measurement. The coefficients for quantitative predictors represent slopes. For example, the quantitative predictor, CycleTime, being a significant predictor of CD0, and CycleTime having a coefficient of 2, indicates that on average the CD0 increases by 2 ms per minute of cycling and the effect is significantly different from zero. Likewise, if the quadratic term, CycleTime^2, had a significant coefficient of 3 for the CD0, then on average the CD0 would increase by 3 ms per squared minute of cycling. A significant effect of the CycleTime×Subject interaction indicates that the effect of CycleTime on the measured parameter depends on the subject. As an example, if CycleTime×Subject has a significant S2 interaction coefficient of 5 for the CD0, then this indicates that the CD0 increased by 5 ms more per cycling minute for S2 than for S1. This means that the slope that CycleTime represents is steeper for S2 than for S1 (i.e., the delay increased faster for S2).

To interpret the additional quantitative predictors, consider the CD25 regression as an example. Including CD0 in the CD25 regression essentially segments the measurements. The CycleTime parameter from the CD25 regression indicates the rate CD25 is changing per minute of cycling relative to the CD0 measurement. For example, if CycleTime is 2 from the CD0 regression and CD0 and CycleTime from the CD25 regression are 1 and 3, respectively, then on average CD25 would increase by 3 ms per minute of cycling relative to CD0 and would increase by 5 ms per minute of cycling $(3 + 2(1) = 5)$ relative to the instant the stimulation began.

III. Results

The effect of the number of minutes spent cycling on the T_{max} and T_{avg} is depicted in Fig. 3 and the effect of the number of minutes spent cycling on the six delay measurements is depicted in Fig. 4. To better understand the range of the two torque and six delay measurements over all the experiments, a table of descriptive statistics for each measurement is provided in Table II. Measurements for 10 subjects in Table II resulted in N = 440 samples. Regressions were performed on T_{max}, T_{avg}, CD0, CD25, CD75, RD0, RD25, and RD75 and the results for CD0, CD25, and CD75 are provided in Table III and the results for RD0, RD25, and RD75 are provided in Table IV. For visual clarity, statistical significance of the fitted coefficients is indicated in each table by *, **, and *** for P-Values less than or equal to 0.05, 0.01, and 0.001, respectively. Fitted coefficients that are not significant (P-Values > 0.05) are indicated by ns. For each regression the adjusted R^2 was between 66% and 95%, which indicates a good fit was achieved [13]. To validate each regression model, normal probability plots were created and normality of the residual errors was visually confirmed for each measurement. To quantify the rate at which the two torque and six delay measurements vary with cycling time (fatigue), a table of descriptive statistics for the rates of change of each variable is provided in Table V.

A. Torque

CycleTime, CycleTime×Subject, and CycleTime^2 were all statistically significant predictors (P-value < 0.05) of

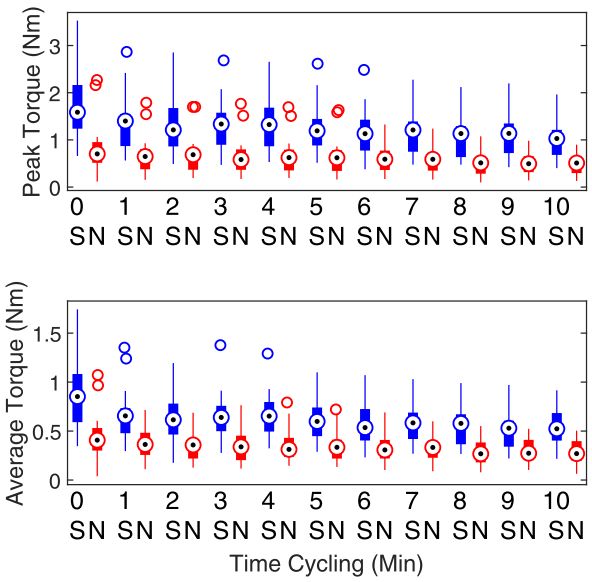

Fig. 3. Box plots of the torque measurements for participants with (N) and without (S) neurological conditions. The median is depicted by a black dot within a white circle, the edges of the box denote the 25th (Q1) and 75th (Q3) percentiles, the whiskers denote the most extreme data points that are not considered to be outliers, and the outliers are indicated by circles. A data point is considered an outlier if it is below Q1-1.5(Q3-Q1) or above Q3+1.5(Q3-Q1). As the cycling time increased the peak and average torques tended to decrease, indicating that cycling resulted in fatigue.

T_{max} and T_{avg}, indicating that fatigue (induced by cycling) occurs and has a significant effect on the FES-induced torque production confirming a hypothesis of the article. By using the coefficients from the regression analyses for CycleTime and CycleTime×Subject, it was determined that both T_{max} and T_{avg} tended to decrease for each participant per cycling minute. Since CycleTime^2 had a positive coefficient for both T_{max} and T_{avg}, the rate at which T_{max} and T_{avg} decreases per cycling minute becomes less steep as the cycling time progresses.

B. Delay

From Tables III and IV, it can be seen that CycleTime, CycleTime×Subject, and CycleTime^2 are significant predictors and hence fatigue (induced by cycling) has a significant effect on the FES-induced EMD confirming the other hypothesis of the article.

IV. Discussion

Although the results of this study, such as those in Table II, can be compared to the results of prior studies, all prior studies focused on single joint tasks where the effects of more complicated tasks that require multiple muscle groups at once were not considered. Previous studies also only focused on the initial CD and not the time to produce different levels of torque or force production (CD25 or CD75) and few considered the RD [13]. Further, an important observation of this study is that similar trends occurred across the different populations in this study, which is likely due to the fact that the energy conversion process resulting from the application of an electric field to

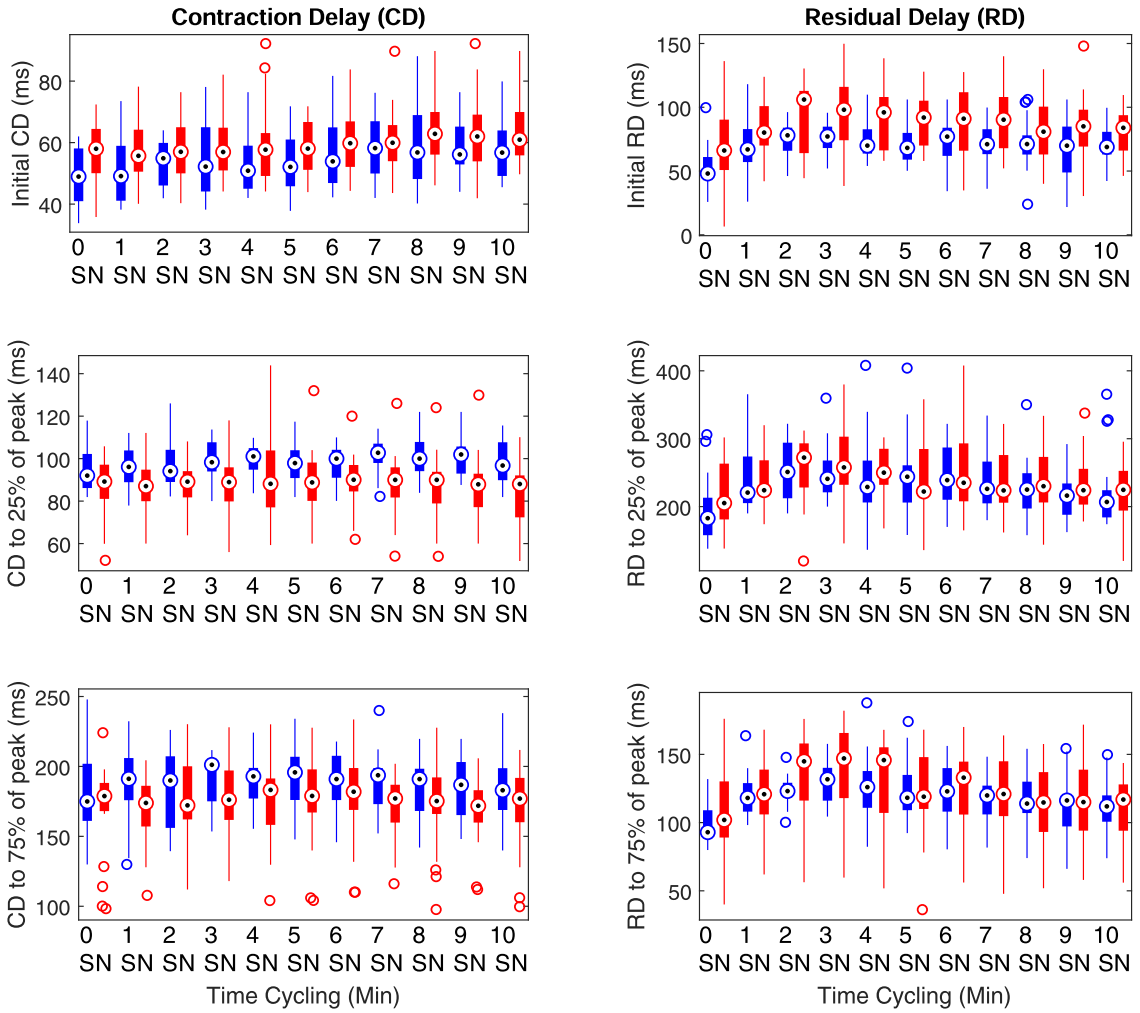

Fig. 4. Box plots of the delay measurements for participants with (N) and without (S) neurological conditions. The median is depicted by a black dot within a white circle, the edges of the box denote the 25th (Q1) and 75th (Q3) percentiles, the whiskers denote the most extreme data points that are not considered to be outliers, and the outliers are indicated by circles. A data point is considered an outlier if it is below Q1-1.5(Q3-Q1) or above Q3+1.5(Q3-Q1). The CD subplots show a general increasing or flat trend. The RD subplots depict that the RD initially increased and later began to decrease.

TABLE II
DESCRIPTIVE STATISTICS

Subjects		S1-S5			N1-N5			S1-S5, N1-N5				
Variable	Units	Q1	Median	Q3	Q1	Median	Q3	Min	Q1	Median	Q3	Max
T_{max}	Nm	0.822	1.203	1.493	0.354	0.584	0.782	0.116	0.538	0.796	1.265	3.529
T_{avg}	Nm	0.433	0.597	0.741	0.221	0.318	0.425	0.038	0.296	0.427	0.632	1.742
CD0	ms	46.1	54.0	62.0	52.1	59.6	66.0	33.8	49.7	56.3	64.2	92.2
CD25	ms	91.0	99.0	104.9	81.0	88.3	96.1	51.8	85.8	93.9	102.1	143.9
CD75	ms	170.1	191.9	202.4	162.7	176.2	191.8	97.7	164.3	182.0	198.1	247.9
RD0	ms	59.8	70.2	82.0	64.1	88.3	106.0	6.5	62.1	75.8	94.2	149.8
RD25	ms	199.9	223.8	257.3	207.8	233.2	279.1	119.9	204.3	226.2	272.0	408.0
RD75	ms	106.1	118.0	130.1	102.9	123.8	145.7	36.2	104.1	119.8	138.2	187.8

the generation of torque is largely invariant to the causation of different neurological conditions.

A. Torque

Prior studies have investigated the change in torque or force production as a result of fatigue [13], [29].[6] In Rampichini *et al.* [29], after two minutes of stimulation to the gastrocnemius medialis the peak force decreased from 687 N to 639 N and in Downey *et al.* [13], FES over a 5 minute duration in the quadriceps femoris resulted in the peak torque decreasing from 25.05 Nm to 5.35 Nm. However, it is unknown how the torque production will vary as a result of FES-cycling. In Fig. 3, the peak and average torques decreased as the cycling time increased. As

[6]In [13], [29] results were for clinically healthy participants.

TABLE III
REGRESSIONS ON CD MEASUREMENTS (*ms*)

	CD0				CD25				CD75			
Term	Coef	SE Coef	P-Value	Sig.	Coef	SE Coef	P-Value	Sig.	Coef	SE Coef	P-Value	Sig.
Constant	66.60	2.12	0.000	***	28.43	3.38	0.000	***	50.45	8.97	0.000	***
CycleTime	1.35	0.42	0.001	***	0.13	0.361	0.711	ns	2.04	0.898	0.024	*
CycleTime^2	-0.017	0.031	0.594	ns	-0.089	0.027	0.001	***	-0.169	0.067	0.012	*
CD0/CD25*					1.03	0.04	0.000	***	1.19	0.08	0.000	***
Side												
Dominant	-1.80	1.83	0.326	ns	0.60	1.57	0.704	ns	12.84	3.91	0.001	***
Muscle												
Glute	-13.40	1.83	0.000	***	21.68	1.67	0.000	***	10.81	3.96	0.007	**
Subject												
S2	-6.65	2.90	0.022	*	5.05	2.49	0.043	*	20.20	6.18	0.001	***
S3	-13.47	2.90	0.000	***	4.58	2.54	0.072	ns	6.35	6.22	0.308	ns
S4	-22.46	2.90	0.000	***	7.08	2.65	0.008	**	-6.57	6.31	0.298	ns
S5	-11.78	2.90	0.000	***	5.63	2.53	0.026	*	1.92	6.20	0.757	ns
N1	-17.24	2.90	0.000	***	3.71	2.58	0.151	ns	30.18	6.28	0.000	***
N2	2.34	2.90	0.420	ns	-4.69	2.48	0.059	ns	31.43	6.18	0.000	***
N3	-5.49	2.90	0.059	ns	-31.04	2.49	0.000	***	-13.40	6.84	0.051	ns
N4	-15.31	2.90	0.000	***	6.28	2.56	0.015	*	14.60	6.22	0.019	*
N5	3.95	2.90	0.174	ns	5.02	2.48	0.044	*	-3.88	6.22	0.533	ns
Side×Muscle												
Dominant×Glute	1.80	1.11	0.105	ns	-4.34	0.95	0.000	***	-5.69	2.36	0.016	*
CycleTime×Subject												
S2	-1.00	0.39	0.010	**	0.13	0.34	0.709	ns	-0.03	0.84	0.970	ns
S3	0.57	0.39	0.143	ns	0.68	0.33	0.044	*	-1.97	0.84	0.019	*
S4	-0.68	0.39	0.083	ns	0.27	0.33	0.418	ns	-1.37	0.83	0.101	ns
S5	-0.05	0.39	0.889	ns	0.15	0.33	0.663	ns	-0.96	0.83	0.251	ns
N1	-0.35	0.39	0.370	ns	-0.73	0.33	0.030	*	-1.09	0.84	0.195	ns
N2	-0.96	0.39	0.015	*	-0.08	0.34	0.810	ns	0.56	0.84	0.501	ns
N3	-0.50	0.39	0.199	ns	-0.01	0.33	0.983	ns	0.01	0.83	0.986	ns
N4	-0.43	0.39	0.276	ns	0.26	0.33	0.433	ns	-0.76	0.83	0.359	ns
N5	-0.12	0.39	0.760	ns	0.09	0.33	0.787	ns	-0.06	0.83	0.941	ns
Side×Subject												
Dominant×S2	-7.84	2.47	0.002	**	9.80	2.14	0.000	***	0.69	5.27	0.895	ns
Dominant×S3	-3.28	2.47	0.185	ns	9.53	2.11	0.000	***	4.30	5.29	0.417	ns
Dominant×S4	0.84	2.47	0.734	ns	13.39	2.11	0.000	***	16.74	5.39	0.002	**
Dominant×S5	-5.82	2.47	0.019	*	13.20	2.12	0.000	***	-0.25	5.30	0.963	ns
Dominant×N1	-5.11	2.47	0.039	*	19.06	2.12	0.000	***	-19.50	5.38	0.000	***
Dominant×N2	-3.61	2.47	0.145	ns	7.06	2.12	0.001	***	-38.38	5.27	0.000	***
Dominant×N3	3.08	2.47	0.214	ns	-0.55	2.11	0.796	ns	11.28	5.27	0.033	*
Dominant×N4	-3.14	2.47	0.205	ns	7.42	2.11	0.000	***	10.45	5.28	0.048	*
Dominant×N5	-4.63	2.47	0.062	ns	-12.84	2.12	0.000	***	4.145	5.45	0.415	ns
Muscle×Subject												
Glute×S2	12.38	2.47	0.000	***	-11.86	2.18	0.000	***	-3.36	5.27	0.524	ns
Glute×S3	0.62	2.47	0.803	ns	-9.94	2.11	0.000	***	15.80	5.32	0.003	**
Glute×S4	8.50	2.47	0.001	***	-12.13	2.14	0.000	***	-0.02	5.27	0.997	ns
Glute×S5	8.60	2.47	0.001	***	-10.65	2.14	0.000	***	4.14	5.27	0.432	ns
Glute×N1	11.81	2.47	0.000	***	-19.59	2.12	0.000	***	-5.70	5.30	0.283	ns
Glute×N2	4.85	2.47	0.051	ns	-13.60	2.12	0.000	***	0.76	5.31	0.886	ns
Glute×N3	4.61	2.47	0.063	ns	-8.20	2.12	0.000	***	-14.16	5.27	0.008	**
Glute×N4	12.74	2.47	0.000	***	-19.98	2.18	0.000	***	-7.94	5.30	0.135	ns
Glute×N5	9.95	2.47	0.000	***	-19.17	2.15	0.000	***	-6.29	5.31	0.238	ns
R^2_{adj}	71.2%				87.7%				79.8%			

*The quantitative predictor CD0 was included in the regression on CD25 and the quantitative predictor CD25 was included in the regression on CD75.

a muscle fatigues, the force that it generates decreases, thus Fig. 3 confirms the hypothesis that FES-cycling does induce fatigue. The median peak torque was found to be 1.59 Nm (0.70 Nm) before cycling and 1.03 Nm (0.51 Nm) after 10 minutes of cycling for the participants without (with) NCs. A one-tail, unpaired t-test was performed using the combined data across all cycling times to conclude that T_{max} and T_{avg} are significantly smaller (P-value $<$ 0.001) for participants

TABLE IV
REGRESSIONS ON RD MEASUREMENTS (*ms*)

	RD0				RD75				RD25			
Term	Coef	SE Coef	P-Value	Sig.	Coef	SE Coef	P-Value	Sig.	Coef	SE Coef	P-Value	Sig.
Constant	59.13	4.97	0.000	***	47.18	3.83	0.000	***	112.51	14.18	0.000	***
CycleTime	6.58	0.98	0.000	***	1.50	0.68	0.029	*	9.95	2.17	0.000	***
CycleTime^2	-0.573	0.07	0.000	***	-0.298	0.052	0.000	***	-0.536	0.17	0.002	**
RD0/RD75*					0.84	0.03	0.000	***	1.06	0.10	0.000	***
Side												
Dominant	7.70	4.30	0.074	ns	0.457	2.85	0.873	ns	-48.57	9.08	0.000	***
Muscle												
Glute	-0.27	4.30	0.951	ns	8.66	2.84	0.002	**	1.52	9.09	0.867	ns
Subject												
S2	-14.27	6.80	0.036	*	-1.90	4.52	0.675	ns	-25.10	14.34	0.081	ns
S3	8.61	6.80	0.206	ns	11.55	4.50	0.011	*	-12.61	14.43	0.383	ns
S4	-20.07	6.80	0.003	**	23.51	4.54	0.000	***	-10.91	14.32	0.447	ns
S5	-12.09	6.80	0.076	ns	3.26	4.51	0.470	ns	-19.36	14.33	0.177	ns
N1	4.61	6.80	0.498	ns	3.11	4.49	0.489	ns	-22.68	14.33	0.114	ns
N2	46.40	6.80	0.000	***	4.13	4.75	0.385	ns	12.02	14.93	0.421	ns
N3	18.20	6.80	0.008	**	1.68	4.53	0.711	ns	55.05	14.41	0.000	***
N4	35.95	6.80	0.000	***	9.92	4.65	0.033	*	-37.70	14.85	0.012	*
N5	-11.47	6.80	0.092	ns	1.55	4.51	0.730	ns	-22.25	14.33	0.121	ns
Side×Muscle												
Dominant×Glute	3.77	2.59	0.147	ns	-0.83	1.72	0.629	ns	4.71	5.47	0.390	ns
CycleTime×Subject												
S2	0.20	0.92	0.830	ns	0.69	0.61	0.256	ns	-5.49	1.93	0.005	**
S3	-0.38	0.92	0.678	ns	0.81	0.61	0.184	ns	-6.57	1.93	0.001	***
S4	0.65	0.92	0.477	ns	1.92	0.61	0.002	**	-3.32	1.94	0.088	ns
S5	-1.79	0.92	0.052	ns	0.77	0.61	0.206	ns	-9.62	1.93	0.000	***
N1	-1.51	0.92	0.099	ns	-0.07	0.61	0.912	ns	-4.85	1.93	0.012	*
N2	-1.87	0.92	0.041	*	0.28	0.61	0.642	ns	-1.75	1.93	0.366	ns
N3	0.72	0.92	0.433	ns	0.21	0.61	0.731	ns	-5.84	1.93	0.003	**
N4	-1.53	0.92	0.095	ns	0.47	0.61	0.439	ns	-5.99	1.93	0.002	**
N5	0.58	0.92	0.530	ns	1.33	0.61	0.040	*	-3.72	1.94	0.055	ns
Side×Subject												
Dominant×S2	5.20	5.80	0.370	ns	6.51	3.83	0.090	ns	34.59	12.25	0.005	**
Dominant×S3	-12.83	5.80	0.027	*	-6.72	3.85	0.082	ns	46.75	12.33	0.000	***
Dominant×S4	23.56	5.80	0.000	***	-4.50	3.91	0.250	ns	15.46	12.30	0.209	ns
Dominant×S5	11.12	5.80	0.056	ns	13.19	3.85	0.001	***	112.59	12.40	0.000	***
Dominant×N1	-17.71	5.80	0.002	**	10.59	3.88	0.007	**	43.01	12.21	0.000	***
Dominant×N2	-5.91	5.80	0.309	ns	8.77	3.84	0.023	*	0.33	12.21	0.979	ns
Dominant×N3	-40.81	5.80	0.000	***	-38.01	4.06	0.000	***	-0.94	14.14	0.947	ns
Dominant×N4	-3.01	5.80	0.603	ns	8.88	3.83	0.021	*	67.93	12.22	0.000	***
Dominant×N5	42.10	5.80	0.000	***	-7.76	4.08	0.058	ns	71.94	12.50	0.000	***
Muscle×Subject												
Glute×S2	-0.58	5.80	0.920	ns	-8.64	3.83	0.025	*	2.02	12.24	0.869	ns
Glute×S3	-10.19	5.80	0.080	ns	-1.95	3.85	0.612	ns	12.30	12.25	0.316	ns
Glute×S4	8.30	5.80	0.153	ns	-18.09	3.84	0.000	***	23.67	12.25	0.054	ns
Glute×S5	3.52	5.80	0.544	ns	-4.16	3.83	0.278	ns	4.85	12.20	0.692	ns
Glute×N1	-2.77	5.80	0.633	ns	-9.06	3.83	0.019	*	-0.59	12.25	0.962	ns
Glute×N2	-14.35	5.80	0.014	*	-9.53	3.86	0.014	*	-2.19	12.39	0.860	ns
Glute×N3	-2.75	5.80	0.636	ns	-8.15	3.83	0.034	*	-11.64	12.25	0.342	ns
Glute×N4	-1.83	5.80	0.752	ns	-9.06	3.83	0.019	*	-3.26	12.25	0.790	ns
Glute×N5	-9.20	5.80	0.113	ns	-7.55	3.84	0.050	*	3.50	12.30	0.776	ns
R^2_{adj}	69.6%				88.8%				66.2%			

*The quantitative predictor RD0 was included in the regression on RD75 and the quantitative predictor RD75 was included in the regression on RD25.

with NCs than those with none. Future attempts to minimize FES-cycling induced fatigue can be compared against the rates and findings of this article.

B. Delay

Recently studies have investigated EMD changes due to FES induced fatigue [13], [29]. Rampichini *et al.* [29] reports

TABLE V
DESCRIPTIVE STATISTICS OF THE RATES OF CHANGE FOR EACH VARIABLE

Subjects			S1-S5		N1-N5		S1-S5, N1-N5				
Variable	Rate Type*	Units†	Mean	SD	Mean	SD	Mean	SD	Slowest	Fastest	Quadratic
T_{max}	N/A	Nm/min	-0.081	0.023	-0.051	0.029	-0.066	0.029	-0.032	-0.104	0.0025 Nm/min^2
T_{avg}	N/A	Nm/min	-0.044	0.011	-0.030	0.010	-0.037	0.013	-0.022	-0.057	0.0019 Nm/min^2
CD0	Overall	ms/min	1.118	0.617	0.878	0.308	0.998	0.476	0.35	1.92	-0.017 ms/min^2
CD25	Relative to CD0	ms/min	0.376	0.261	-0.011	0.335	0.183	0.349	0.05	0.81	-0.089 ms/min^2
CD25	Overall	ms/min	1.528	0.808	0.894	0.452	1.211	0.702	0.43	2.79	-0.107 ms/min^2
CD75	Relative to CD25	ms/min	1.174	0.856	1.772	0.656	1.473	0.785	0.07	2.6	-0.169 ms/min^2
CD75	Overall	ms/min	2.992	0.707	2.835	0.871	2.913	0.752	1.46	3.85	-0.296 ms/min^2
RD0	Overall	ms/min	6.316	0.930	5.858	1.261	6.087	1.073	4.71	7.30	-0.573 ms/min^2
RD75	Relative to RD0	ms/min	2.338	0.689	1.944	0.532	2.141	0.616	1.43	3.42	-0.298 ms/min^2
RD75	Overall	ms/min	7.643	1.193	6.865	1.410	7.254	1.298	5.69	9.49	-0.779 ms/min^2
RD25	Relative to RD75	ms/min	4.950	3.601	5.52	1.752	5.235	2.687	0.33	9.95	-0.536 ms/min^2
RD25	Overall	ms/min	13.052	4.234	12.797	2.126	12.924	3.162	7.00	17.40	-1.362 ms/min^2

*For CD (RD) measurements the overall rate is relative to the instant stimulation began (ended).
†The unit min represents the number of minutes cycling.

that after two minutes of stimulation to the gastrocnemius medialis the EMD increased from 26.85 ms to 31.74 ms. Downey *et al.* [13] reports that FES over a 5 minute duration in the quadriceps femoris from a high-fatiguing protocol (10 s of stimulation every 15 s) resulted in CD0 increasing from 52.06 ms to 128.34 ms. From Fig. 7 in [13], it can be seen that the low-fatiguing protocol (5 s of stimulation every 15 s) caused CD0 to increase from 52 ms to 62 ms. In the present study, the quadriceps femoris and the gluteal muscle groups were stimulated over 10 minutes of FES-induced cycling resulting in an increased median of CD0 from 54.0 ms to 59.8 ms after 10 minutes of cycling. From [13], a protocol that only changes the duration of stimulation can result in a significant difference in the change in CD0 from before to after the protocol. Therefore, the variation between the change in CD0 across different studies is likely due to a variation in stimulation intensity or duration for each study.

As indicated in Fig. 4, CD0 tends to increase with cycling time, indicating CD0 increased with fatigue; thus as T_{max} decreased (Fig. 3), CD0 increased. This is consistent with the finding in [22], where it was found that at lower isometric forces the delay was larger. From Fig. 4, it can be seen that the other EMD measurements do not generally increase with cycling time. After normalizing each EMD measurement by its respective peak torque measurement, the normalized EMD tended to increase with time indicating that T_{max} has a strong influence on the EMD. Therefore, it is possible that EMD varies with fatigue because fatigue causes T_{max} to decrease.

By inspection of Fig. 4 and Tables II and V it can be seen that the EMD was different for the participants with NCs and those with none. A two-tail, unpaired t-test indicated that CD0, CD25, CD75, and RD0 were significantly different (P-value < 0.001) for able-bodied participants and those with NCs (Table II). From Table II, participants with NCs had on average a 48.5% smaller median T_{max} than the able-bodied participants, which likely contributes to the difference between the groups. The results in Tables II and V provide results on the EMD for both groups of participants as well as a combination, which can be used to bound the torque and EMD and their rates of change. A diverse population was recruited because it was desired for the bounds to represent a varied population.

Another finding of this study is that the CD and RD are not the same. For example, using the CD0 and RD0 data for all participants, a two-tail, unpaired t-test was used to conclude that CD0 and RD0 are different (P-value < 0.001). The difference between the CD and RD measurements is also apparent in Fig. 4. In Table V, it is noticeable that the RD increases with cycling time at a faster rate than the CD. Additionally, the regression results in Tables III and IV confirm that the muscle group, the side, and the interaction Side × Muscle were statistically significant (P-value < 0.05) predictors of the EMD, indicating that the EMD varies among different muscle groups.

C. Closed-Loop Control

The results in this article can be used to improve the future development of closed-loop controllers for FES-cycling by providing insight into how the torque and EMD should be modeled and by establishing a range of expected values for the torque and the EMD. For example, the results indicate that future control designs should include different delays in the dynamic model for the CD, RD, and each muscle combination. To account for inter-subject variability, previous closed-loop controllers for FES systems typically utilize robust control design methods, which require the delay to be lower and upper bounded by known constants [5]–[7]. The EMD bounds are important because they are used to determine when to apply/cease stimulation in an effort to properly time when FES should be applied so that muscle contractions occur at times that yield effective torque production, which can potentially reduce the rate of fatigue. The results in this article provide the control designer with a range of expected bounds for each EMD measurement. The bounds on the rate of change of each EMD measurement can likewise inform adaptive update

laws that estimate the delay to yield a more accurate estimate of the EMD throughout an experiment, which will improve performance.

V. Conclusion

The present study used plots and statistical results to confirm the hypothesis that FES-induced cycling will result in the torque and EMD varying with cycling time. The EMD was divided into six different measurements to better understand how the EMD varied with time. To aid future control development, bounds were established on the torque and EMD and on the rate of change of both. Additionally, the study indicated that the CD and RD are different and that the EMD varies between muscle combinations. The results in this study can be used to inform the development of closed-loop controllers that account for the existence of a time-varying EMD. These future efforts may lead to improved assistive devices and rehabilitative treatments. Additional studies could further investigate the effects of fatigue on the EMD at various crank angles.

Acknowledgment

Any opinions, findings and conclusions, or recommendations expressed in this material are those of the author(s) and do not necessarily reflect the views of the sponsoring agency.

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