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Torque and Neuromuscular Responses are not Joint Angle Dependent During a Sustained, Isometric Task Anchored to a High Perceptual Intensity

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Abstract Few studies have assessed changes in the time course of the torque and neuromuscular responses during a sustained, isometric task anchored to a constant rating of perceived exertion. The purpose of the present study was to examine the effects of joint angle on the torque and neuromuscular responses during sustained, isometric forearm flexion tasks anchored to RPE = 7 (OMNI-RES scale). Ten college-aged (mean ± SD: age = 21.3 ± 1.8 yrs.) men agreed to participate in this cross-sectional study and performed two, 3s maximal voluntary isometric contractions (MVIC) at elbow joint angles of 75° and 125° before sustained, isometric, forearm flexions anchored to RPE = 7 to task failure at the respective joint angles. The amplitude (AMP) and frequency (MPF) of the electromyographic (EMG) and mechanomyographic (MMG) signals from the biceps brachii were recorded. Repeated measures ANOVAs and Bonferroni corrected dependent t-tests were used to examine differences across time and between joint angles for torque and neuromuscular parameters. There were decreases ($p < 0.05$) in torque and EMG AMP across time that were not joint angle dependent, but, there were no changes ($p > 0.05$) for the other neuromuscular parameters. The results indicated three distinct phases for the torque versus time relationships for both joint angles, including 1) An initial rapid decrease in torque; 2) followed by a plateau; and 3) a final decline in torque to task failure. From these responses, we hypothesized that afferent feedback from group III/IV motor neurons and corollary discharge caused decreases in torque to maintain the prescribed RPE.

Keywords: *perception, exertion, fatigue, torque, electromyography, mechanomyography*

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1. Introduction

During exercise, ratings of perceived exertion (RPE) have been used to quantify the conscious sensations of how hard, heavy, and strenuous a given task is [1]. Robertson and Noble [2] hypothesized that RPE is a numerically coded, gestalt-like, global response reflective of the complex integration of sensory and perceptual factors as influenced by physiological signal mediators, psychological factors, the performance milieu, and exertional symptoms. Examples of these mediators, factors, symptoms, and the performance milieu include variables such as oxygen delivery, oxygen uptake, metabolite accumulation, availability of energy substrates, motivation, mood, exercise experience, competitive strategy, time and/or distance required, competitive history, heavy breathing, sweat, joint and/or muscle pain. More recently, coaches, clinicians, and researchers have utilized RPE to prescribe exercise intensity [3], autoregulate resistance training intensity and volume [4], and assess the physiological and psychological mechanisms of fatigue [5,6].

Over the years, the interpretation of fatigue has been confounded by the various methods and techniques used to study this phenomenon as well as the many characteristics assigned to its definition [7,8]. Therefore, Kluger et al. [8] proposed a unified taxonomy of fatigue that included interdependent aspects of performance fatigability and perceived fatigability. Performance fatigability was defined as "…the magnitude or rate of change in a performance criterion relative to a reference value over a given time of task performance or measure of mechanical output" [8]. Enoka and Duchateau [7] have stated that performance fatigability is modulated by factors associated with contractile function such as calcium kinetics, force capacity, blood flow, and metabolism. It has also been proposed that muscle activation including voluntary activation, activation patterns, motor neurons, afferent feedback, and neuromuscular propagation influence performance fatigability. Perceived fatigability refers to "…subjective sensations of weariness, increases in sense of effort, mismatch between effort expended and actual performance, or exhaustion" [8]. Perceived fatigability is influenced by factors associated with the maintenance of homeostasis such as blood glucose, core temperature, hydration, neurotransmitters, metabolites, oxygenation, and wakefulness, as well as the individual's psychological state including arousal, executive function, expectations, mood, motivation, pain, and performance feedback [7].

Recent studies have reported [6,9] that anchoring a fatiguing task to RPE while assessing various aspects of performance [7,8,10] allows for the examination of the interactions among factors associated with perceived fatigability and performance fatigability. It has been suggested [11], that the magnitude of performance fatigability is determined by the mode and intensity of exercise which dictates the amount of muscle mass activated and the subsequent demands on the various systems of the body during the task. According to Thomas et al. [11], lesser engaged muscle mass should produce less systemic perturbations and result in greater performance fatigability before "…the task is perceived as intolerable" (p. 242). This hypothesis has recently been examined by anchoring unilateral and bilateral leg extension tasks by RPE and examining the performance-related changes in force and neuromuscular parameters [12]. Few studies, however, have used the RPE Clamp Model [13] to assess fatigue-induced changes in torque and neuromuscular responses for forearm flexion which includes even less activated muscle than that associated with the leg extensors. For example, Smith et al. [9] anchored torque to a constant RPE of 7 using the OMNI-RES (0 – 10) RPE Scale [14] during a sustained, isometric forearm flexion task to examine the patterns of fatigue-induced changes in torque as well as the electromyographic (EMG) and mechanomyographic (MMG) signals.

The simultaneous examination of the time and frequency domain parameters of the EMG and MMG signals have been used to describe the fatigue-induced patterns of responses during various exercise modalities [5,13,15]. Specifically, the amplitude (AMP) of the EMG signal reflects muscle activation, while the mean power frequency (MPF) is associated with muscle fiber action potential conduction velocity [16]. Under some conditions, the AMP of the MMG signal reflects motor unit recruitment [17] and the MMG MPF qualitatively represents changes in the global firing rate of the activated, unfused motor units [18]. Thus, these neuromuscular measures potentially allow for inferences to be made regarding fatigue-induced changes in motor unit activation strategies.

A number of studies have examined the influence of joint angle on force (or torque) output during isometric tasks as well as the fatigue-induced changes in EMG and MMG parameters during sustained, isometric tasks [19,20]. It is well established that when muscle length and joint angle are altered, there is a significant impact on torque production [21]. For example, it has been suggested that the joint angle associated with the greatest isometric torque ranges from approximately 90 - 120°, while torque decreases at a longer or shorter muscle length beyond that range of motion [21]. Furthermore, Weir et al. [20] reported joint-angle specific EMG and MMG responses from the tibialis anterior muscle during sustained, isometric plantarflexion and dorsiflexion tasks and suggested that motor unit recruitment was greater for the larger joint angle versus the smaller joint angle. Doheny et al. [19], however, reported no differences in muscle activation (EMG AMP) during brief, maximal, isometric forearm flexion tasks at eight different elbow joint angles (range: 10 - 120°). The contrasting findings suggest that the relationship between torque and motor unit activation strategies may be joint angle and/or task dependent. Furthermore, the relationship between joint angle and fatigue-induced changes for torque and neuromuscular responses during sustained, isometric upper body tasks at a fixed RPE remains unclear. Thus, the purpose of the present study was to examine the effects of joint angle on the torque and neuromuscular responses during sustained, isometric forearm flexion tasks anchored to RPE = 7 (OMNI-RES scale). Based on the findings of previous studies [9,22,23] that utilized the RPE-Clamp Model during sustained, isometric tasks, we hypothesized that: 1) torque would decrease across the sustained task; 2) there would be fatigue-induced decreases in EMG AMP and increases in MMG AMP, but no changes in the MPF of the EMG and MMG signals. Furthermore, based on the findings of previous studies [19,24,25] that examined the effects of joint angle on maximal isometric torque production and neuromuscular responses of the forearm flexors, we hypothesized that: 3) the torque values from the maximal voluntary isometric contractions (MVIC) would be similar at the elbow joint angle (EJ) of 75° and 125°; and, 4) joint angle would not affect the neuromuscular parameters during the MVICs or the sustained, isometric tasks.

2. Materials and Methods

An a priori G*Power3 power analysis determined that a minimum of 4 subjects were required to demonstrate mean differences between 2 dependent groups using repeated-measures ANOVAs, an effect size of η_p^2 = 0.738 [26], a power of a power of 0.95, and an alpha of 0.05. Thus, to ensure adequate power, ten men (mean ± SD: age = 21.3 ± 1.8 yrs.; height = 179.9 ± 6.5 cm; body mass = 85.9 ± 18.1 kg) volunteered to participate in this study. Via the Health History Questionnaire, the subjects were identified as recreationally active (defined as participating in resistance and/or aerobic exercise at least 3 $d \cdot wk^{-1}$ for at least 30 minutes for ≥ 6 months prior to screening), and all of the subjects were free of upper body pathologies that would affect their performance. The subjects in the present study were part of a large multiple independent and dependent variable investigation, but none of the data in the present study have been previously published (Smith et al., 2021). The study was approved by the University Institutional Review Board for Human Subjects (IRB approval #: 20201220785FB), and all subjects completed a Health History Questionnaire and signed a written Informed

Consent prior to testing. Individuals were eligible to participate if they were between the ages of 19 and 29, were recreationally active, in good health as assessed by the Health History Questionnaire and were willing to comply with the study protocol. Individuals were not eligible to participate if there were indications of health-related issues based on the Health History Questionnaire. Such indications included symptoms of chest pain, breathing difficulties, irregular heartbeat, kidney or liver problems, high blood pressure or cholesterol, and/or abnormal electrocardiogram (ECG) as well as muscle or skeletal disorders including previous or current shoulder, arm, and/or forearm injuries were also considered when determining eligibility.

2.1. Time Course of Procedures

Each subject visited the laboratory on three separate occasions (orientation session, testing visit 1 and testing visit 2) each was separated by 24 – 96 hours. The initial visit was an orientation session where demographic information was recorded, and the subjects were familiarized with the standardized warm-up consisting of six, 3 s submaximal, (~ 50-75% of their maximal effort), isometric forearm flexion contractions as well as the experimental protocol. Testing visits 1 and 2 included: 1) The standardized warm-up; 2) the anchoring procedures; 3) two, 3 s forearm flexion maximal voluntary isometric contractions (MVICs) to set a perceptual anchor to RPE = 10; and 4) a sustained, isometric forearm flexion task to failure anchored to RPE = 7. In addition, during each testing visit, EMG and MMG signals from the biceps brachii were simultaneously recorded.

2.2. Orientation Session

During the orientation session, the subject's dominant arm (based on throwing preference), age, height, and body mass were recorded. In addition, the subject was oriented to their testing position on the isokinetic dynamometer (Cybex 6000, Cybex International Inc. Medway, MA). While positioned, the subject was familiarized with the 0 – 10 OMNI-RES scale [14] and read the standardized OMNI-RES instructions that were used during the testing visits [14]. The OMNI-RES (0 – 10) RPE scale has been shown to be valid and reliable for the quantification of perception of exertion during resistance exercise [14]. The subject then completed the standardized warm-up, two, 3 s isometric forearm flexion MVICs to set a perceptual anchor corresponding to RPE = 10, and a brief (approximately 1 min), sustained, isometric task anchored to RPE = 7 to become familiar with the testing/anchoring procedures.

2.3. OMNI-RES Scale Standardized Anchoring Instructions

The anchoring instructions used in the present study have been modified for use during isometric forearm flexion tasks [9]. Therefore, to promote the proper use of the OMNI-RES scale, the following standardized anchoring instructions were read to each subject during the orientation session and prior to each sustained, isometric task anchored to RPE = 7, "You will be asked to set an anchor point for both the lowest and highest values on the perceived exertion scale. In order to set the lowest anchor, you will be asked to lay quietly without contracting your forearm flexor muscles to familiarize yourself with a zero. Following this, you will be asked to perform a maximal voluntary isometric contraction to familiarize yourself with a 10. When instructed to match a perceptual value corresponding to the OMNI-RES scale, perceived exertion should be relative to these defined anchors."

2.4. Testing Visits

Before and after each testing visit, subjects were instructed to avoid upper body exercise at least 24 hours prior to testing. During each testing visit, the subject was positioned in accordance with the Cybex 6000 user's manual on an upper body exercise table (UBXT) with the lateral epicondyle of the humerus of the dominant arm aligned with the lever arm of the dynamometer. Once positioned, the subject performed the standardized warm-up. After the warm-up, the subject was read the OMNI-RES instructions relating to the anchoring procedures. The subject then performed two, 3 s forearm flexion MVICs on a calibrated dynamometer at EJ_{75} and EJ_{125} in randomized order. Strong verbal encouragement was provided during each MVIC trial, and the MVIC was performed to familiarize the subject with RPE = 10 on the OMNI-RES scale. A rest period of approximately 1 minute was provided between the standardized warm-up and the MVIC trials. For forearm flexion, EJ_{75} and EJ_{125} were selected to reflect a range of muscle lengths and isometric torque production [21]. Following the MVIC trials, the sustained, submaximal, isometric forearm flexion task anchored to RPE = 7 on the OMNI-RES scale was performed at a randomly selected EJ angle (EJ_{75} or EJ_{125}). A rest period of approximately 2 minutes was provided between the MVIC trials and the sustained, isometric tasks. During the sustained isometric task, the subject was blinded to torque and elapsed time to avoid pacing strategies [27]. The RPE trial was sustained until task failure, which was defined as a torque that would require RPE > 7, or the torque was reduced to zero. Thus, during the RPE trial, the subject was free to decrease torque to maintain a constant RPE of 7. Upon task failure, the forearm flexion task was terminated and time to task failure (TTF) was recorded. The percent decline in torque was calculated by the following equation:

$$Percent\ Decline = \frac{Initial\ torque\ value - Torque\ at\ Task\ Failure}{Initial\ Torque\ Value} \times 100$$

In addition, during the sustained isometric task, the subject was reminded to be attentive to sensations such as strain, intensity, pain, and effort during the contraction to maintain appropriate levels of exertion [2]. Furthermore, the subject was continuously reminded that there were no incorrect contractions or perceptions and were reminded to relate levels of exertion to the previously set anchors.

2.5. Electromyographic, Mechanomyographic, and Torque Signal Acquisition

During the testing visits, bipolar (30-mm center-to-center) EMG electrodes (pregelled Ag/AgCl, AccuSensor; Lynn Medical, Wixom, MI) were attached to the biceps brachii (BB) of the dominant arm based on the recommendations of the Surface Electromyography for the Non-Invasive Assessment of Muscles [28]. A reference electrode was placed on the styloid process of the radius of the forearm. Prior to electrode placement, the skin was shaved, carefully abraded, and cleaned with alcohol. The electrodes were placed between the medial acromion and the fossa cubit, at one-third the distance from the fossa cubit over the BB. Using double-sided adhesive tape, a miniature accelerometer (ICP® Accelerometer, bandwidth 0-1000 Hz, dimensions 0.48 × 1.22 × 0.71 cm, mass 0.85 g, sensitivity 103.4 $mV \cdot g^{-1}$; PCB Piezotronics, Depew, NY) was placed between the bipolar EMG electrodes to detect the MMG signals for the BB muscles.

The raw EMG and MMG signals were digitized at 2000 samples/second with a 12-bit analog-to-digital converter (Model MP150; Biopac Systems, Inc.) and stored on a personal computer (Acer Aspire TC-895-UA91 Acer Inc., San Jose, CA, USA) for analyses. The EMG signals were amplified (gain: × 1000) using differential amplifiers (EMG2-R Bionomadix, Biopac Systems, Inc. Goleta, CA, USA; bandwidth—10-500 Hz). The EMG and MMG signals were digitally bandpass filtered (fourth-order Butterworth) at 10-500 Hz and 5-100 Hz, respectively. Signal processing was performed using custom programs written with LabVIEW programming software (version 20.0f1, National Instruments, Austin, TX, USA). The TTF (0 – 100%) was divided into 5% increments and a 1 s epoch from the center of each 5% increment (i.e., 500ms before and 500ms after) was used to calculate the AMP (root mean square) for EMG (µVrms) and MMG ($m \cdot s^{-2}$) signals, as well as the mean power frequency (MPF in Hz) for both signals. The MPF was selected to represent the power density spectrum and was calculated as described by Kwatny et al. [29]. The torque signals were sampled from the digital torque of the Cybex 6000 dynamometer and stored on a personal computer (Acer Aspire TC-895-UA91 Acer Inc., San Jose, CA, USA) for analysis. The pretest forearm flexion MVIC with the greatest torque production was used to normalize the torque, EMG, and MMG parameters for each 5% of the TTF corresponding to the respective EJ angle tested during the sustained task (Figure 1 – Figure 3). In addition, the pretest forearm flexion MVIC with the greatest torque production was used to normalize the initial torque value and neuromuscular parameters (EMG AMP, EMG MPF, MMG AMP, and MMG MPF) of the first 3 s of the sustained, isometric task anchored to RPE = 7 (Figure 1 – Figure 5). The initial torque value was defined as the average torque value during the first 3 s of the sustained, isometric forearm flexion task anchored to RPE = 7.

2.6. Statistical Analysis

The mean differences between joint angles for the pretest MVIC, TTF, initial torque, and percent decline (PD) in torque values were determined with four, separate dependent t-tests. Furthermore, the mean differences for the normalized torque and neuromuscular parameters were determined with five, separate 2 (Joint Angle: 75° and 125°) × 21 (Time: Initial Value and % TTF) repeated measures ANOVAs. Tests for sphericity (Mauchly's Test of Sphericity) were conducted for all dependent variables and if sphericity was violated, the Greenhouse-Geisser correction was utilized. Significant interactions were decomposed with appropriate follow-up ANOVAs and Bonferroni corrected dependent t-tests were used to identify the time course of when the normalized torque and neuromuscular values changed from the value corresponding to 5% TTF. In addition, Bonferroni corrected dependent t-tests were used to identify any differences for the normalized torque and neuromuscular values between each elbow joint angle (EJ_{75} vs. EJ_{125}) at each time point. Effect sizes were reported as partial eta-squared (η_p^2) and Cohen's *d* for the ANOVAs and pairwise comparisons, respectively. An alpha value of *p*-value ≤ .05 was considered statistically significant for the ANOVAs and Bonferroni corrected alpha values of $p \leq 0.0025$ were considered statistically significant for the dependent t-tests across time and between elbow joint angles, respectively. All the data was reported as mean ± SD and all calculations and statistical analyses were carried out in IBM SPSS v. 28 (Armonk, NY, USA).

3. Results

3.1. Pretest MVIC, Initial Torque, Percent Decline, and TTF

The pretest MVIC (N·m), initial torque (N·m and % of pretest MVIC), torque at task failure, and percent decline from the initial torque to task failure values for EJ_{75} and EJ_{125} are presented in Table 1. The TTF values for EJ_{75} and EJ_{125} during the sustained, isometric tasks anchored to RPE = 7 were 521.8 ± 327.2 s (range = 164.0 – 1273.0 s) and 572.7 ± 333.2 s (range = 239.0 – 1234 s), respectively. In addition, there were no significant differences between EJ_{75} and EJ_{125} for pretest MVIC ($p = 0.081$, $d = 0.825$), initial torque ($p = 0.201$, $d = 0.594$), percent decline ($p = 0.904$, $d = 0.055$), or TTF ($p = 0.750$, $d = 0.145$).

3.2. Torque Responses

The mean (± SD) torque responses for EJ_{75} and EJ_{125} are presented in Figure 1. For the torque responses during the sustained, isometric tasks, there was a significant ($p = 0.023$, $\eta_p^2 = 0.167$) Joint Angle × Time interaction. Bonferroni corrected dependent t-tests indicated that the initial torque value was significantly greater than the value at 5% TTF for EJ_{75} ($p = 0.0012$, $d = 1.464$) and EJ_{125} ($p = 0.0013$, $d = 1.461$), respectively. Bonferroni corrected dependent t-tests indicated that the mean torque values from 20 – 100% TTF ($p < 0.001$, *d* range: 1.510 – 3.264) and 15 – 100% TTF ($p < 0.001$, *d* range: 1.523 – 2.820) were less than the value at 5% TTF for EJ_{75} and EJ_{125}, respectively. In addition, Bonferroni corrected dependent t-tests indicated that there were no significant ($p > 0.0025$) differences between EJ_{75} and EJ_{125} at the initial torque value or the mean torque values from 5 to 100% TTF.

Table 1. Pretest MVIC (N·m), initial torque (N·m), normalized initial torque (% of pretest MVIC), and percent decline for the torque output from the initial torque at an elbow joint (EJ) angle of 75° and 125° at RPE = 7 during the sustained isometric tasks to failure

Joint Angle	Subjects	Pretest MVIC	Initial Torque	% MVIC	Torque at Task Failure	Percent Decline (%)
EJ_{75}	1	46	17.5	38.0	0.0	100.0
	2	25	17.0	68.0	5.0	70.6
	3	47	10.8	23.0	0.0	100.0
	4	61	22.8	37.3	0.0	100.0
	5	69	30.0	43.5	0.0	100.0
	6	44	21.5	48.9	0.0	100.0
	7	48	21.8	45.5	1.0	95.4
	8	55	30.8	56.1	3.0	90.3
	9	52	21.3	40.9	0.0	100.0
	10	40	22.3	55.8	0.0	100.0
Mean ± SD		48.7 ± 11.9	21.6 ± 5.9	45.7 ± 12.4	0.9 ± 1.7	95.6 ± 9.4
EJ_{125}	1	56	18.5	33.0	7.0	100.0
	2	46	26.9	58.5	0.0	100.0
	3	51	43.3	84.9	0.0	100.0
	4	64	29.0	45.3	0.0	100.0
	5	70	47.3	67.5	0.0	100.0
	6	68	31.0	45.6	0.0	100.0
	7	62	33.7	54.3	0.0	100.0
	8	48	18.8	39.1	0.0	100.0
	9	59	37.7	63.8	0.0	100.0
	10	49	23.3	47.6	0.0	100.0
Mean ± SD		57.3 ± 8.6	30.9 ± 9.7	54.0 ± 15.3	0.7 ± 2.2	96.2 ± 12.0

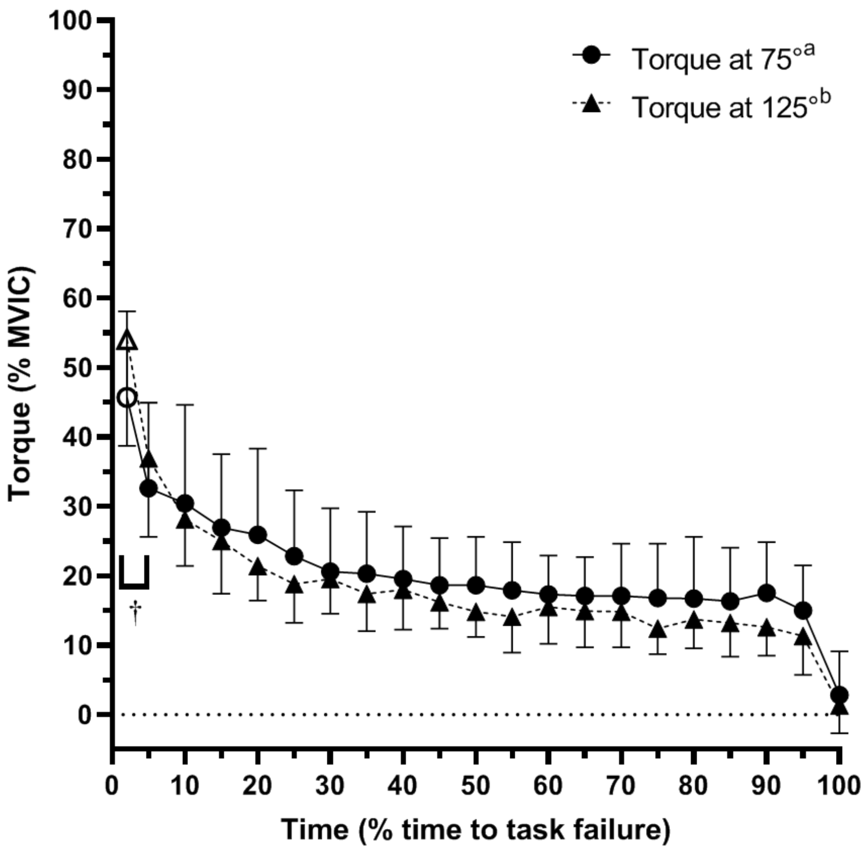

Figure 1. Time course of changes for the normalized (% of pretest MVIC) mean (± SD) torque values during the sustained, isometric forearm flexion tasks at an elbow joint angle of 75° and 125° anchored to RPE = 7 († significantly ($p < 0.001$) greater initial torque value than the torque value at 5% TTF of the sustained, isometric task at an elbow joint angle of 75° and 125°. [a] significantly ($p < 0.0025$, Bonferroni corrected) lower torque values than the value at 5% TTF of the sustained, isometric task from 20 – 100% TTF at an elbow joint angle of 75°. [b] significantly ($p < 0.0025$, Bonferroni corrected) lower torque values than the value at 5% TTF of the sustained, isometric task from 15 – 100% TTF at an elbow joint angle of 125°)

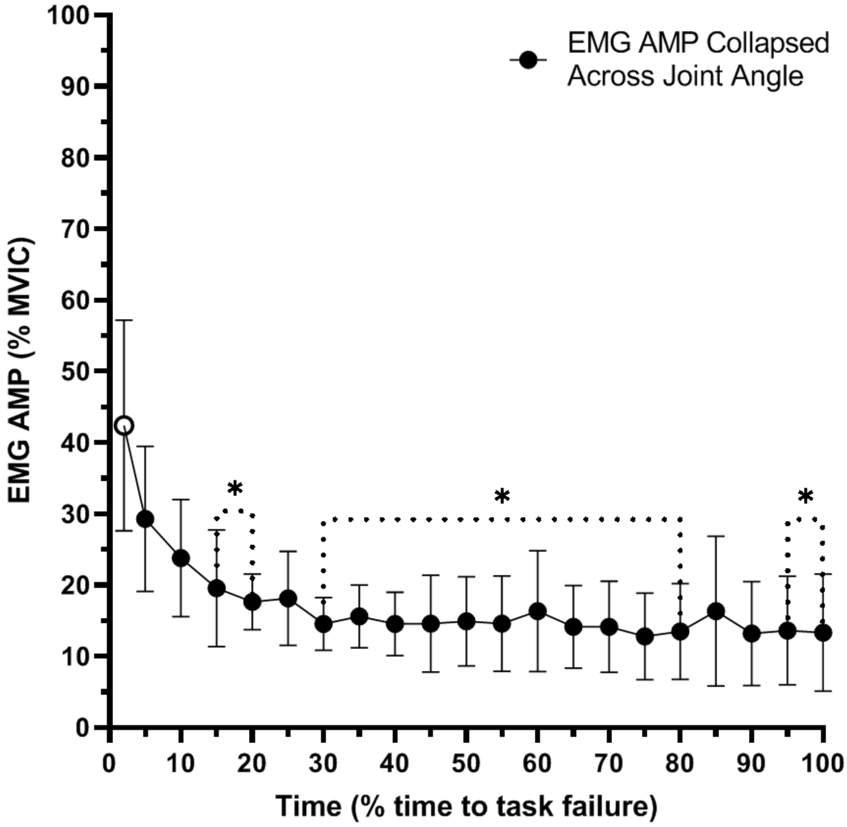

Figure 2. Time course of changes for the normalized (% of pretest MVIC) marginal mean (± SD) EMG AMP values (collapsed across Joint Angle) during the sustained, isometric forearm flexion tasks anchored to RPE = 7 (* significantly ($p < 0.0025$, Bonferroni corrected) lower EMG AMP values than the value at 5% TTF of the sustained, isometric task)

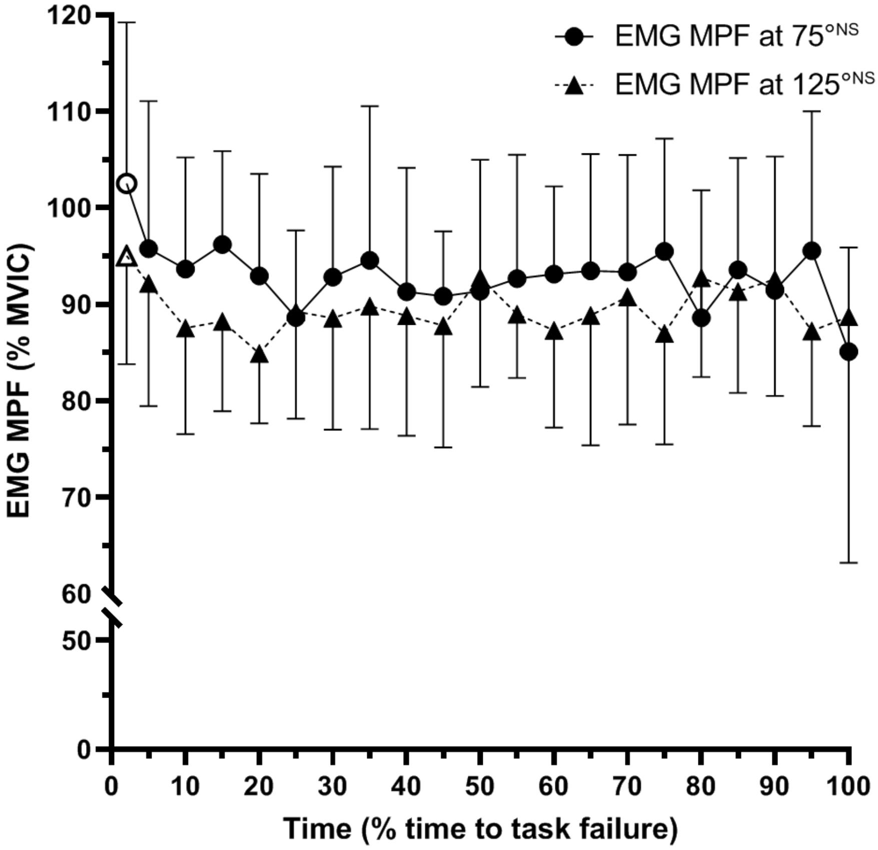

Figure 3. Time course of changes for the normalized (% of pretest MVIC) mean (± SD) EMG MPF values during the sustained, isometric forearm flexion tasks at an elbow joint angle of 75° and 125° anchored to RPE = 7 (^{NS} No significant ($p > 0.05$) differences for EMG MPF during the sustained, isometric forearm flexion tasks at an elbow joint angle of 75° and 125° anchored to RPE = 7)

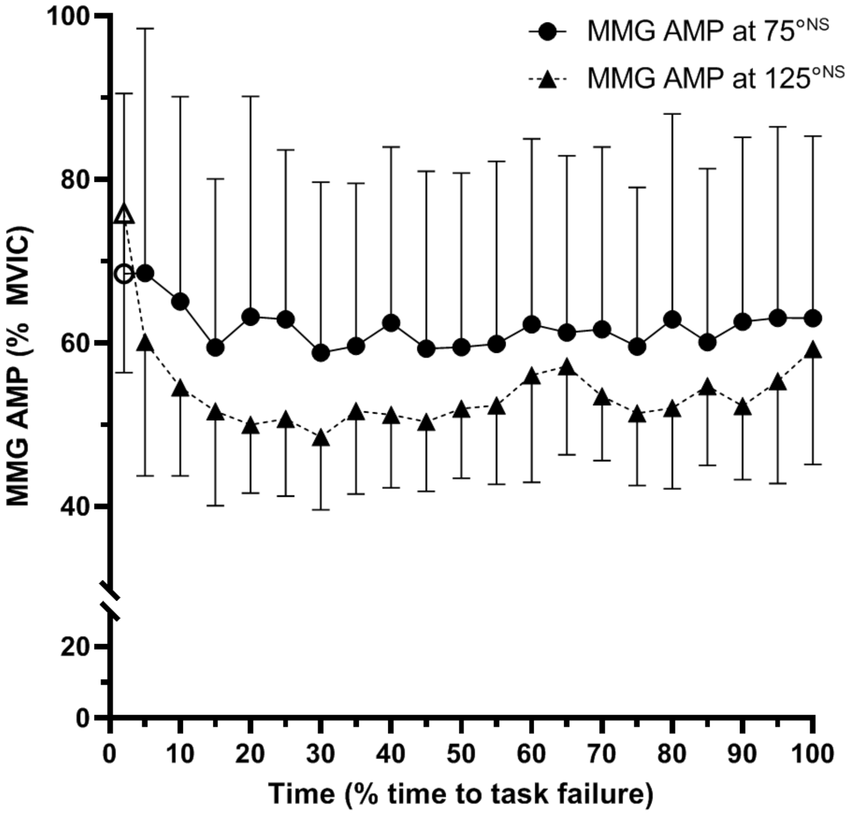

Figure 4. Time course of changes for the normalized (% of pretest MVIC) mean (± SD) MMG AMP values during the sustained, isometric forearm flexion tasks at an elbow joint angle of 75° and 125° anchored to RPE = 7 (^{NS} No significant (p > 0.0025, Bonferroni corrected) differences for EJ_{75} or EJ_{125} between the initial MMG AMP value and the value at 5% TTF, no changes across time, or between EJ_{75} and EJ_{125} at the initial MMG AMP value or the mean MMG AMP values from 5 to 100% TTF)

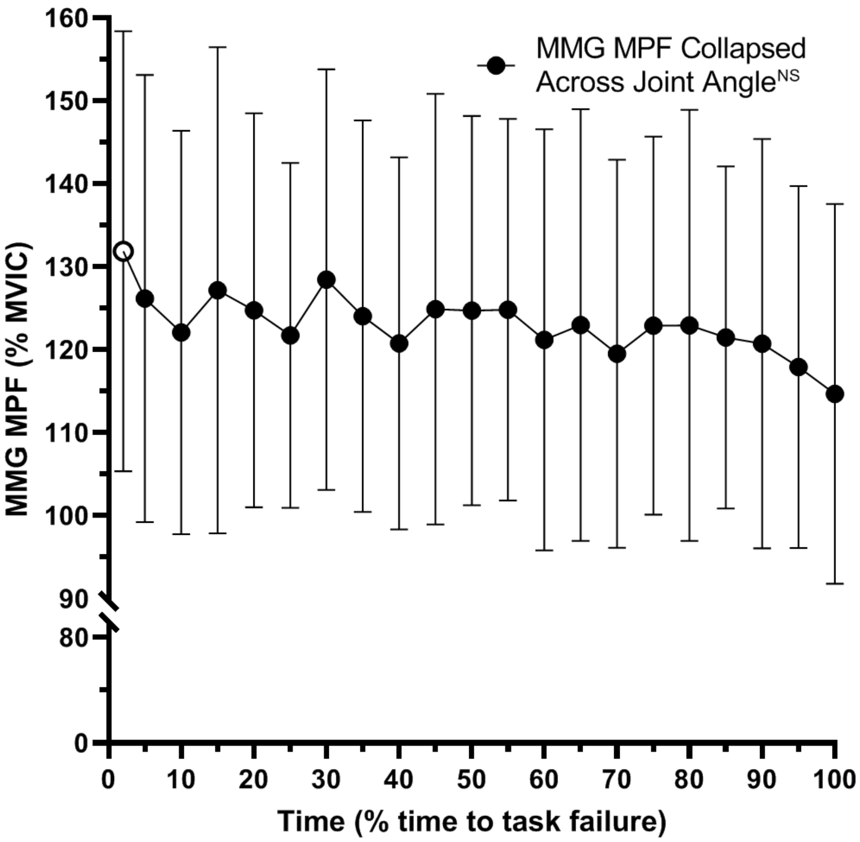

Figure 5. Time course of changes for the normalized (% of pretest MVIC) marginal mean (± SD) MMG MPF values (collapsed across Joint Angle) during the sustained, isometric forearm flexion tasks anchored to RPE = 7 (^{NS} No significant (p < 0.0025, Bonferroni corrected) differences for EJ_{75} or EJ_{125} between the initial MMG MPF value and the value at 5% TTF and no changes across time)

3.3. EMG Responses

The mean (± SD) EMG AMP and EMG MPF values are presented in Figure 2 and Figure 3, respectively. For the EMG AMP responses during the sustained, isometric tasks, there was no significant ($p = 0.103$, $\eta_p^2 = 0.139$) Joint Angle × Time interaction. There was, however, a significant ($p = 0.036$, $\eta_p^2 = 0.403$) main effect (collapsed across Time) for Joint Angle and a significant ($p < 0.001$, $\eta_p^2 = 0.656$) main effect (collapsed across Joint Angle) for Time. A Bonferroni correct dependent t-test indicated that there was no difference ($p = 0.015$, $d = 0.943$) between the initial EMG AMP value and the value at 5% TTF. Bonferroni corrected dependent t-tests indicated, however, that the EMG AMP values from 15 – 20%, 30 – 80%, and 95 – 100% TTF ($p < 0.0025$, d range: 1.323 – 1.887) were less than the value at 5% TTF. In addition, Bonferroni corrected dependent t-tests indicated that there were no significant ($p > 0.0025$) differences between EJ_{75} and EJ_{125} at the initial EMG AMP value or the mean EMG AMP values from 5 to 100% TTF. For the EMG MPF responses, there was no significant ($p = 0.311$, $\eta_p^2 = 0.113$) Joint Angle × Time interaction, main effect ($p = 0.459$, $\eta_p^2 = 0.062$) for Joint Angle, or main effect ($p = 0.052$, $\eta_p^2 = 0.153$) for Time.

3.4. MMG Responses

The mean (± SD) MMG AMP and MMG MPF values are presented in Figure 4 and Figure 5, respectively. For the MMG AMP responses during the sustained, isometric tasks, there was a significant ($p = 0.011$, $\eta_p^2 = 0.180$) Joint Angle × Time interaction. Bonferroni corrected dependent t-tests indicated, however, there was no significant ($p > 0.0025$) differences for EJ_{75} or EJ_{125} between the initial MMG AMP value and the value at 5% TTF, no changes across time, or between EJ_{75} and EJ_{125} at the initial MMG AMP value or the mean MMG AMP values from 5 to 100% TTF. For the MMG MPF responses, there was no significant ($p = 0.437$, $\eta_p^2 = 0.102$) Joint Angle × Time interaction and no main effect ($p = 0.802$, $\eta_p^2 = 0.007$) for Joint Angle. There was, however, a significant ($p < 0.001$, $\eta_p^2 = 0.265$) main effect (collapsed across Joint Angle) for Time. Bonferroni corrected dependent t-tests indicated, however, there was no significant ($p > 0.0025$) differences for EJ_{75} or EJ_{125} between the initial MMG MPF value and the value at 5% TTF and no changes across time.

4. Discussion

In the present study, there were no differences between the mean pretest MVIC values for EJ_{75} or EJ_{125} (Table 1). Typically, the highest MVIC value for forearm flexion occurs at an elbow joint angle between approximately 90 – 120°, with lower values at the extremes of the range of motion [21]. It has been suggested [30], that joint angle and muscle length specific differences in MVIC values are due to the degree of overlap of actin and myosin and cross-bridge attachments, with a suboptimal amount of overlap at the longer joint angles and too much overlap at the shorter joint angles. Based on previous studies [19,24,31] and the shape of the typical joint angle versus torque relationship [25], we hypothesized that the MVIC values at EJ_{75} and EJ_{125} would be similar. Although the MVIC at EJ_{75} was 15.1% less than EJ_{125} was, we found no significant differences between the mean MVIC values. Future research should extend the current findings by assessing elbow joint angles that are less than 75° and greater than 125°.

The initial torque values in the present study ranged from 23.0 – 84.9% MVIC, with no difference in the torque output between joint angles (Table 1). These findings were consistent with previous studies that have examined the RPE versus % MVIC relationship and reported that subjects tend to perceptually underestimate the expected torque or force output when anchored to RPE [32,33]. For example, Smith et al. [9] reported that during an isometric forearm flexion task anchored to RPE = 7, the initial torque was 59.7 ± 15.0% MVIC. West et al. [34] hypothesized that subjects may subconsciously under produce force at higher intensities as a protective mechanism to prevent excessive mechanical or metabolic challenges to the muscle. This hypothesis seems unlikely given that the subjects in the present study consciously determined their torque output and were free to reduce torque during the task. Tucker [13] suggested, however, that when exercise is anchored to RPE, there is an anticipatory component that, in theory, is determined by a combination of previous experiences, the task modality, physiological and psychological inputs, which are processed within the brain to set an initial exercise intensity that is perceived to match the prescribed RPE. Thus, based on Tucker [13], the initial torque values (Table 1) at the start of the sustained tasks in the present study, were perceived to match the prescribed RPE = 7 based on an anticipatory component that integrated previous experiences with current physiological status and psychological perceptions of the tasks.

In the current study, 70% (seven out of 10) of subjects reduced torque to zero at EJ_{75} and 90% at EJ_{125}, respectively. Furthermore, there were significant declines in torque (> 90%) at each joint angle (Table 1). These findings agreed with previous studies that utilized the RPE Clamp Model [13] during isometric leg extensions [22,23] as well as isometric forearm flexion [9]. Specifically, Keller et al. [22,23] reported that during sustained, isometric leg extensions anchored to RPE = 5, 70% of women and 60% of men reached a point where continuing the task would require an RPE > 5 even though torque had not reached zero. Recently, Smith et al. [9] reported that 45% of women reduced torque to zero and the mean torque decreased by 95.69 ± 6.54% during a sustained, isometric forearm flexion task anchored to RPE = 7. In addition, decreases in exercise performance have been reported during dynamic tasks anchored to RPE. For example, Cochrane-Snyman et al. [5] reported a decline in treadmill running speed using the RPE-Clamp Model. Similarly, Flood et al. [35] reported a reduction in power output during cycle ergometry at a fixed RPE. The findings of the present study, in conjunction with previous investigations, suggest that when a task is anchored to a constant RPE, it is necessary to reduce exercise intensity to maintain the prescribed RPE across a variety of exercise modalities [5,9,22,23,35].

There were similar torque responses during the sustained tasks, regardless of joint angle, that were characterized by three phases across time (Figure 1). The first phase was from the initial torque and neuromuscular values (average of the first 3 s) to 20% TTF at EJ_{75} and 15% TTF at EJ_{125} (Figure 1). In addition, the first phase included two unique segments (segment 1 and segment 2). During segment 1, there were precipitous decreases from the initial torque values to 5% TTF at each joint angle. These decreases in torque were likely associated with conscious decreases in central drive and de-recruitment of motor units that were mirrored by the neuromuscular responses (Figure 2 – Figure 5). These findings suggested that, regardless of joint angle, the initial torque was almost immediately perceived as too high, and that it was necessary to reduce the intensity of the contraction so RPE would not exceed 7. Group III afferent neurons have been suggested to be sensitive to the mechanical changes within the muscle including the stretch and intensity of the contraction [36]. Perhaps, the decision to decrease torque during segment 1 was informed by a combination of feedback, likely to the supplementary motor area (SMA) [37], from group III afferent neurons [36] and corollary discharge from the premotor and primary motor areas of the brain [10,13,38]. It could be, initially, that the feedback from group III afferent neurons was processed within the SMA and the decision to reduce torque was fedforward to the premotor and primary motor areas of the brain, which resulted in a decrease in central drive to the muscles, and, therefore, resulted in de-recruitment of motor units and reduced torque output [10]. Based on the Corollary Discharge Model [38], the premotor and primary motor areas then generated an efferent copy (i.e., an internal signal that develops from the central motor commands), which provided immediate neural feedback to the SMA to determine whether the reduction in torque was sufficient to match the prescribed RPE = 7. Thus, we hypothesize that the combined feedback from group III afferent neurons and the efferent copies generated from central motor command were integrated within the SMA to determine if the torque output sufficiently matched the prescribed RPE of 7.

After segment 1 of the first phase, torque continued to decrease at a reduced rate throughout segment 2 from 5% to 20% TTF at EJ_{75}, and 15% TTF at EJ_{125} (Figure 1). These findings indicated that there were dissociations between the lack of changes in the neuromuscular responses and RPE, and the decrease in torque. During segment 2, it is likely that the reduction in torque to maintain RPE = 7 was informed by afferent feedback from group III and group IV (primarily metabosensitive) neurons due to increased levels of metabolic byproducts such as inorganic phosphate and hydrogen ions within the active muscle fibers [36,39]. Based on the RPE Clamp Model [13], mechanical and metabolic perturbations within the muscle increase afferent feedback to the brain, which would influence the perception of exertion and result in continuous adjustments to exercise performance to return the conscious RPE to the prescribed level. Previously, Broxterman et al. [40] demonstrated using fentanyl administration (afferent blockage) that group III and group IV afferent neurons act to inhibit intramuscular metabolic perturbations during exercise. Although group III afferent neurons are primarily sensitive to mechanical changes within the muscle, a small number of group III neurons have also been reported to be sensitive to intramuscular metabolic perturbations [36]. Given that the torque values across segment 2 were above the torque (22% MVIC) typically associated with the onset of blood flow restriction during sustained, isometric forearm flexion [41], and that blood flow restriction has been associated with intramuscular metabolic perturbations [39], it is likely that an accumulation of metabolites within the muscle was sufficient to cause inhibitory feedback from group IV neurons. Therefore, like segment 1, during segment 2, there was a similar process of neuronal feedback from the premotor and primary motor areas to the SMA, which integrated the afferent signals from group IV and a small number of group III (metabosensitive) neurons as well as the efferent copies generated from central motor command that likely contributed to the additional reductions in torque.

During the second phase of the sustained, isometric tasks in the present study, torque was characterized by a plateau from 20 – 95% TTF at EJ_{75} and 15 – 95% TTF at EJ_{125} (Figure 1). During this period, each of the neuromuscular parameters and torque remained unchanged and tracked RPE. Furthermore, the consistent neuromuscular responses suggested the subjects were able to sustain RPE and torque across the second phase with no changes in muscle activation (EMG AMP), muscle fiber action potential conduction velocity (EMG MPF), motor unit recruitment (MMG AMP), or global firing rate of unfused activated motor units (MMG MPF). Based on the RPE Clamp Model [13], the plateaus in torque at each joint angle, corresponded to an intensity that the subjects perceived as sustainable at RPE = 7. Given that the average torque values in the present study were lower than the intensity (22% MVIC) typically associated with the onset of blood flow restriction [41], it is likely that blood flow was adequate during phase two, which would allow for clearance of metabolites and the replenishment of energy substrates [42]. Similar force or torque and neuromuscular responses were reported by Keller et al. [22], Keller et al. [23], and Smith et al. [9], which support the findings of the present study and the suggestion that the plateaus in torque and neuromuscular parameters, regardless of joint angle, resulted from the subjects perception that the self-selected torque output was sustainable at RPE = 7.

During the third phase of the sustained, isometric tasks torque decreased from 95 – 100% TTF for both joint angles (Figure 1). These findings indicated that there were dissociations for the torque versus perceptual and neuromuscular parameters which suggested a decrease in neuromuscular efficiency (normalized torque/normalized EMG AMP) [43]. These time-dependent patterns for torque and neuromuscular parameters were consistent with peripheral fatigue and the characteristics of excitation-contraction coupling failure due to the effects of exercise-induced, intramuscular metabolic perturbations on calcium release and re-uptake kinetics, calcium sensitivity for binding with troponin, and actin-myosin binding properties [39,42]. Perhaps, factors associated with perceived fatigability such as mood, and/or motivation also contributed to the decrease in torque to zero during the third phase [7,8]. For example, Kluger et al. [8] and Enoka and Duchateau [7] have suggested motivation may cause a reduction in exercise performance to manage the development of fatigue. Thus, some subjects may have lost motivation to continue the task. It is also possible that the sum of all neural feedback from the primary and synergistic muscles involved with forearm flexion and corollary discharge associated with central command led to voluntary reductions in torque and, ultimately, termination of the task based on the Sensory Tolerance Limit (STL) model [44]. Specifically, Hureau et al. [44] described the Sensory Tolerance Limit as a global model of fatigue where the sum of all neural feedback, from systems directly and/or indirectly involved in the exercise task, along with efferent copies associated with central command are integrated within the brain and cause a reduction in performance or termination of the task. The lateral prefrontal cortex (LPFC) has been proposed as a region of the brain which is primarily responsible for the decision to continue or terminate exercise [45]. Specifically, Robertson and Marino [45] proposed that the LPFC integrates afferent feedback sent from the anterior cingulate cortex and the orbitofrontal cortex. Once this information has been integrated within the LPFC, the decision to modify exercise intensity or terminate exercise is passed through the premotor area and the basal ganglia via feedforward mechanisms [45]. Thus, according to the STL model, it may be, that a combination of peripheral fatigue and feedback from muscles involved with forearm flexion and handgrip caused a reduction in torque and, ultimately, termination of the exercise [44].

In summary, the present study utilized the RPE Clamp Model [13] during sustained, isometric forearm flexion tasks anchored to a high perceptual intensity to examine the effects of joint angle on the time course of torque and neuromuscular responses. Our findings indicated that the torque and neuromuscular responses were not joint angle dependent and that the three phases for the torque versus time relationship occurred independent of joint angle. Furthermore, the present findings indicated that each phase may have been informed by different mechanisms to maintain the prescribed RPE. Specifically, we hypothesized that segment 1 of the first phase, was likely mediated by afferent feedback from group III mechanosensitive neurons and the efferent copies associated with central motor command. Like segment 1, segment 2 was likely mediated by a combination afferent feedback from group IV and a small number of group III metabosensitive neurons, as well as the efferent copies associated with central motor command. During the second phase, the self-selected torque output may have allowed for adequate blood flow and was perceived by the subjects to be sustainable at RPE = 7. For phase three, we hypothesized that the subjects likely reached their STL and that the decision to terminate the task was due to a combination of fatigue-induced feedback from muscles involved with forearm flexion and handgrip, as well as psychological factors such as motivation that caused the subjects to terminate the exercise. Although motivation and mood may have also contributed to task termination, it was not measured and thus, follow-up studies should assess various psychological factors to determine if they contribute to the decision to terminate the task. Furthermore, varying perceptual intensities such as an RPE = 3, 5, and 9 were not used in the present study which may have resulted in different time-dependent torque and neuromuscular responses than RPE = 7. Finally, women were not included in the present study and the potential outcomes from the sex-comparisons may have indicated time-dependent torque and neuromuscular responses unique to men or women during tasks anchored to a constant RPE. Future research should also incorporate the use of ultrasound or near-infrared spectroscopy (NIRS) to assess potential changes in blood flow and oxygenation of the active muscle(s) and their effects on torque response when anchored to RPE.

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Statement of Competing Interests

The authors have no competing interests.

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