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Perceptual Fatigability and Neuromuscular Responses During a Sustained, Isometric Forearm Flexion Muscle Action Anchored to a Constant Level of Perceived Exertion

Abstract

Objective: The purpose of the present study was to examine the fatigue-induced changes in torque, and the electromyographic (EMG) and mechanomyographic (MMG) responses during a sustained submaximal, isometric forearm flexion muscle action anchored to a constant rating of perceived exertion (RPE). **Methods:** Eleven women (mean ± SD: age = 20.5 ± 1.9 yrs.; height = 169.9 ± 6.6 cm; body mass = 73.2 ± 15.9 kg) performed 2, 3s forearm flexion maximal voluntary isometric contractions (MVIC) before a sustained isometric muscle action anchored to RPE = 7 until task failure (defined as torque that would require RPE > 7, or the torque was reduced to zero). The EMG amplitude (AMP), EMG mean power frequency (MPF), MMG AMP, and MMG MPF signals from the biceps brachii (BB) were recorded. Regression analyses were conducted to examine the torque and neuromuscular responses vs. time relationships. **Results:** The percent decline in torque during the sustained isometric muscle action was 95.69 ± 6.54 %. There was a significant ($p < 0.001$; R = -0.998), negative quadratic EMG AMP relationship and a significant ($p < 0.046$; R = 0.952), positive quadratic MMG AMP relationship vs. Time, but no significant ($p > 0.05$) relationships for EMG MPF or MMG MPF vs. Time. **Conclusion:** The findings suggested that torque was initially regulated by an anticipatory feedforward mechanism and continually adjusted due to afferent feedback. In addition, substantial inter-individual, as well as differences between the individual and composite responses, were observed for the neuromuscular response patterns.

Keywords

Perception, Exertion, Fatigue, Upper Body, Resistance Training, RPE Clamp Model

Cover Page Footnote

Acknowledgments RWS was primarily involved in data collection, analyses, manuscript writing, and accepts responsibility for the integrity of the data analysis. RWS, TJH, RJS, and GOJ conceived and designed the study. RJS and GOJ provided administrative oversight of the study. All authors contributed to the final drafting and approved the final submission of this manuscript. There was no external funding for this project.

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Introduction

The study of fatigue has been extensive, with the earliest reports dating back almost two centuries (Di Giulio et al., 2006). Thanks to recent technological advances, our understanding of the neurological, psychological, and performance-related effects of fatigue has grown exponentially. Although the collective understanding of fatigue has grown, the interpretation of fatigue has been confounded by the various methods and techniques employed to study it and the multitude of definitions used throughout the literature (Enoka et al., 1992). Therefore, Kluger et al. (2013) proposed a unified taxonomy for the definition of fatigue. Kluger et al. (2013) suggested that fatigue is made up of two interdependent attributes: perceived fatigability and performance fatigability. Perceived fatigability refers to subjective sensations of weariness, perception of exertion, or exhaustion that are influenced by homeostatic and psychological factors sensitive to changes in neurotransmitters, oxygenation, motivation, and mood (Enoka et al., 2016; Kluger et al., 2013). Furthermore, Kluger et al. (2013) proposed that performance fatigability should describe changes in the magnitude of a performance-related measure over the course of time or pre-test to post-test measures in mechanical output. Performance fatigability is influenced by factors such as force production, energy substrate levels, and motor neurons that affect contractile function and muscle activation. Recent studies (Cochrane-Snyman et al., 2016; Keller et al., 2018a, 2019) have utilized specific anchoring schemes, including anchoring intensity by rating of perceived exertion (RPE), force, or power output to examine the characteristics of perceived fatigability and performance fatigability (Cochrane et al., 2015; Enoka et al., 2016; Keller et al., 2018a; Keller et al., 2020a; Kluger et al., 2013; Tucker, 2009). Keller et al. (2018a, 2019) hypothesized that the neuromuscular patterns of responses provide information regarding the fatigue-related mechanisms underlying the specific parameter used to anchor the task. That is, when a fatiguing task is anchored to force, the neuromuscular patterns of responses describe the changes in motor unit activation strategies that are required to maintain force. When the task is anchored to RPE, however, the neuromuscular patterns of responses reflect factors that are required to maintain the predetermined level of perceived exertion.

Perceived exertion is the collective integration of afferent factors such as metabolite accumulation, thermoregulation, and heart rate as well as anticipatory mechanisms such as substrate availability, motivation, and past experience that facilitate an individual’s perception of how hard or easy the exercise feels at any given moment (Robertson et al., 1997; Tucker, 2009). According to Robertson et al. (1997), the perception of exertion as assessed by RPE has been suggested to include components such as the subjective intensity of effort, strain, discomfort, and/or fatigue that is experienced during the task. Originally, the RPE scale was developed to provide clinicians and researchers a reliable method to quantify the subject's perception of exercise exertion based on a 6 - 20 scale (Borg scale) during aerobic testing (Borg et al., 1962). In recent decades, a new RPE 0-10 scale (OMNI-RES) was shown to be valid and reliable for quantifying perception of exertion during resistance training (Lagally et al., 2006; Robertson et al., 2003) and it has been suggested that reporting an RPE that is regionalized to the primary muscle action may enhance the accuracy of the perceptually based intensity during resistance training (Robertson et al., 2003).

Recent studies (Keller et al., 2018a, 2019; Keller et al., 2020a; Keller et al., 2020b) have utilized the OMNI-RES scale to apply the RPE Clamp Model (Tucker, 2009) and anchor exercise intensity allowing for the examination of the effects of perceived fatigability on performance-related measures of fatigue and neuromuscular responses during sustained isometric muscle actions. In a series of studies, Keller et al. (2018a, 2019; 2020a; 2020b) examined the changes in maximal voluntary isometric contraction (MVIC) values from pre-test to post-test following sustained isometric leg extensions in men and women across multiple levels of perceived exertion (RPE = 1, 2, 5, and 8). Furthermore, our laboratory group analyzed the composite and individual electromyographic (EMG) and mechanomyographic (MMG) patterns of responses throughout the sustained isometric muscle actions (Keller et al., 2018a, 2019; Keller et al., 2020a; Keller et al., 2020b). In each study (Keller et al., 2018a, 2019, 2020b), it was evident that individuals must reduce force to maintain the prescribed RPE values. Similar responses for power output and running speed have been reported in studies that utilized the RPE Clamp Model to examine neuromuscular responses during cycle ergometry and treadmill running (Cochrane-Snyman et al., 2016; Cochrane-Snyman et al., 2019; Cochrane et al., 2015).

Although inter-individual differences have been reported (Anders et al., 2019; Cochrane-Snyman et al., 2016), examination of the neuromuscular responses during a sustained task anchored to force have resulted in predictable patterns that have been used to describe fatigue-induced changes in motor unit activation strategies. For example, during sustained isometric lower extremity muscle actions anchored to force, fatigue has been characterized by increases in the amplitude and decreases in the frequency contents of the EMG and MMG signals (Farina et al., 2014; Smith et al., 2016). The typical fatigue-related patterns of responses during sustained isometric forearm flexion, however, indicated increases in EMG AMP, decreases in MMG AMP, and decreases in the frequency contents of both the EMG and MMG signals (Beck et al., 2007; Carr et al., 2018; Kimura et al., 2004; Moritani et al., 1986). Furthermore, the findings of Keller et al. (2018a, 2019) have demonstrated that sustained isometric muscle actions anchored to RPE resulted in different fatigue-induced neuromuscular patterns of responses compared to the typical responses when anchored to force during leg extensions. No previous studies have examined the fatigue-induced EMG, MMG, and torque-related patterns of responses during sustained forearm flexion muscle actions anchored to RPE. Therefore, the purpose of the present study was to examine the fatigue-related patterns of responses for EMG, MMG, and torque during a sustained isometric forearm flexion muscle action anchored to RPE = 7. Based on the previous findings for isometric leg extensions (Keller et al., 2018a, 2019), it was hypothesized that subjects would exhibit decreases in torque and EMG AMP, while there would be increases in MMG AMP and no changes in EMG MPF and MMG MPF during the sustained isometric muscle action.

Methods

Subjects

Eleven females (mean ± SD: age = 20.5 ± 1.9 yrs.; height = 169.9 ± 6.6 cm; body mass = 73.2 ± 15.9 kg) volunteered to participate in this study. The subjects were recreationally trained and participated in resistance and/or aerobic exercise at least 3 $d \cdot wk^{-1}$ (American College of Sports, 2018), 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 collected data has been previously published. 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.

Familiarization Visit

During the familiarization visit, 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 (Robertson et al., 2003) and read the standardized OMNI-RES instructions (Keller et al., 2018b; Robertson et al., 2003) that were used during the experimental visit. The subject then completed the standardized warm-up consisting of 6, submaximal (approximately 50-75% of their maximal effort), isometric forearm flexion muscle actions as well as 2, 3 s maximal voluntary isometric contractions (MVICs) to set a perceptual anchor corresponding to RPE = 10. Finally, the subject performed a brief (approximately 1 – 2 min) submaximal isometric muscle action anchored to RPE = 7 on the OMNI-RES scale to familiarize the subject with the anchoring procedures.

OMNI-RES Scale Standardized Anchoring Instructions

The anchoring instructions used in the present study were originally developed by Gearhart et al. (2001) as a standardized method to gauge training intensity during lower body resistance exercise. The instructions were adapted by Keller et al. (2018b) to be utilized as anchoring procedures during an isometric leg extension task and have been modified for use during forearm flexion in the present study. Therefore, to promote the proper use of the OMNI-RES scale, the following standardized anchoring instructions were read to each subject before the RPE trial, "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."

Experimental Visit

During the experimental 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 (based on throwing preference) aligned with the lever arm of the dynamometer and an elbow joint angle at 100°. Once positioned, the subject performed the standardized warm-up consisting of 6, 3 s, submaximal (approximately 50-75% of their maximal effort), isometric forearm flexion muscle actions followed by 1 min of rest. After the warm-up, the subject was read the OMNI-RES instructions pertaining to the anchoring procedures. The subject then performed 2, 3 s forearm flexion MVICs on a calibrated dynamometer (Cybex 6000, Cybex International Inc. Medway, MA). 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. The elbow joint angle of 100° for forearm flexion was selected to reflect the point in the range of motion that approximated maximal isometric torque production (Kulig et al., 1984). Following the MVIC trials, the sustained submaximal, isometric forearm flexion muscle action anchored to RPE = 7 on the OMNI-RES scale was performed at the elbow joint angle of 100°. During the sustained isometric muscle action, the subject was blinded to torque and elapsed time to avoid pacing strategies (Albertus et al., 2005; Keller et al., 2020a). 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 (Keller et al., 2020a). Thus, during the RPE trial, torque was free to decrease to maintain a constant RPE value. Upon task failure, the forearm flexion muscle action was terminated and time to task failure (TTF) was recorded. In addition, during the sustained isometric muscle action, the subject was reminded to be attentive to sensations such as strain, intensity, discomfort, and fatigue felt during the contraction to maintain appropriate levels of exertion (Keller et al., 2018a; Robertson et al., 1997). Furthermore, the subject was continuously reminded that there were no incorrect contractions or perceptions and were reminded to relate levels of exertion to previously set anchors.

Electromyographic, Mechanomyographic, and Torque Signal Acquisition

During the experimental visit, 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 (Hermens et al., 2000). 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, miniature accelerometers (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) were 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 10 % increments and a 1 s epoch from the center of each 10 % 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 including the MPF (in Hz). The MPF was selected to represent the power density spectrum and was calculated as described by Kwatny et al. (1970). 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.

Statistical Analysis

The pretest forearm flexion MVIC with the greatest torque production was used to normalize the torque, EMG, and MMG parameters for each 10 % of the TTF as well as the initial torque and neuromuscular parameters (EMG AMP, EMG MPF, MMG AMP, and MMG MPF) of the first 3 s of the sustained submaximal, forearm flexion muscle action anchored to RPE = 7 (Figure 1). Separate polynomial regression analyses (linear and quadratic) were used to define the individual and composite relationships for the normalized EMG AMP, EMG MPF, MMG AMP, MMG MPF, and torque values versus normalized time (every 10 %) relationships during the sustained submaximal, forearm flexion muscle action anchored to RPE = 7. All calculations and statistical analyses were carried out in IBM SPSS v. 26 (Armonk, NY, USA). A *p*-value ≤ .05 was considered statistically significant and all the data was reported as mean ± SD.

Results

Table 1. Percent decline for torque (N·m) from the initial torque at RPE = 7 during the sustained isometric muscle action to task failure.

Subjects	Initial Torque	Torque at Task Failure	Percent Decline (%)
1	14.25	3.00	78.95
2	17.33	0.00	100.00
3	26.00	0.50	98.08
4	12.83	1.00	92.21
5	20.25	2.00	90.12
6	25.50	0.50	98.04
7	10.50	0.50	95.24
8	14.33	0.00	100.00
9	24.17	0.00	100.00
10	19.67	0.00	100.00
11	15.50	0.00	100.00
Mean ± SD	18.21 ± 5.32	0.68 ± 0.98	95.69 ± 6.54

MVIC, Percent Decline, and Time to Task Failure

The mean ± SD torque during the pre-test MVIC trials was (31.0 ± 8.3 N·m) and the mean ± SD percent of pre-test MVIC from the initial torque while anchored to RPE = 7 was 59.7 ± 15.0 %. The percent decline for the initial torque to task failure for the individual and mean ± SD values are displayed in Table 1. In addition, the mean time to task failure during the sustained isometric muscle action was 787.0 ± 325.2 s (range = 199 – 1152 s).

Table 2. Polynomial regression model, Correlation (Corr.), and p - value for normalized EMG AMP, EMG MPF, and Torque vs. Time during the sustained submaximal, isometric forearm flexion muscle action anchored to RPE = 7.

Subjects	EMG AMP			EMG MPF			Torque		
	Model	Corr.	*p* - value	Model	Corr.	*p* - value	Model	Corr.	*p* - value
1	Linear	-0.976	< 0.001	-	-	NS	Linear	-0.957	0.002
2	-	-	NS	-	-	NS	Linear	-0.75	0.012
3	Linear	-0.984	< 0.001	-	-	NS	Linear	-0.872	0.001
4	Quadratic	0.960	0.021	-	-	NS	Linear	-0.863	0.001
5	Quadratic	-0.962	0.004	-	-	NS	Linear	-0.859	0.001
6	Quadratic	-0.998	< 0.001	Linear	0.647	0.043	Quadratic	-0.981	0.043
7	Quadratic	-0.984	< 0.001	Quadratic	-0.976	< 0.001	Linear	-0.895	< 0.001
8	-	-	NS	Quadratic	-0.979	0.001	Linear	-0.876	0.001
9	Quadratic	-0.997	< 0.001	Linear	0.936	< 0.001	Linear	-0.896	< 0.001
10	-	-	NS	-	-	NS	Linear	-0.667	0.035
11	-	-	NS	Linear	-0.917	< 0.001	Linear	-0.92	< 0.001
Composite	Quadratic	-0.998	< 0.001	-	-	NS	Linear	-0.917	< 0.001

Torque Responses

During the sustained task, the normalized individual and composite torque responses indicated that there were significant negative linear relationships for torque vs. time (r = – 0.667 to – 0.981) for 10 of the 11 subjects, a negative quadratic relationship (R = – 0.981) for 1 subject, and a negative linear relationship (r = – 0.917) for the composite data (Table 2).

EMG Responses

The EMG AMP responses resulted in significant negative linear relationships (r = – 0.976 and – 0.984) for 2 of the 11 subjects; negative quadratic relationships (R = – 0.962 to – 0.998) for 4 of the 11 subjects; a positive quadratic relationship for 1 subject; no significant relationship for 4 of the 11 subjects; and a negative quadratic relationship (R = – 0.998) for the composite data (Table 2).

The EMG MPF responses resulted in a significant negative linear relationship (r = – 0.917) for 1 subject; positive linear relationships (r = 0.647 and 0.936) for 2 of the 11 subjects; negative quadratic relationships (R = – 0.976 and – 0.978) for 2 of the 11 subjects; no significant relationship for 6 of the 11 subjects; and no significant relationship for the composite data (Table 2).

Table 2. Polynomial regression model, Correlation (Corr.), and p - value for normalized MMG AMP and MMG MPF vs. Time during the sustained submaximal, isometric forearm flexion muscle action anchored to RPE = 7, continued.

Subjects	MMG AMP			MMG MPF		
	Model	Corr.	*p* - value	Model	Corr.	*p* - value
1	Linear	-0.978	< 0.001	Linear	0.911	< 0.001
2	Quadratic	0.997	< 0.001	Linear	-0.938	< 0.001
3	Linear	-0.938	< 0.001	Linear	0.786	0.007
4	Quadratic	0.918	0.012	-	-	NS
5	-	-	NS	Quadratic	0.938	0.006
6	Linear	0.771	0.009	Linear	-0.633	0.049
7	-	-	NS	-	-	NS
8	Quadratic	0.972	0.001	Quadratic	-0.988	< 0.001
9	Quadratic	-0.950	0.007	Quadratic	0.918	0.018
10	Quadratic	0.979	< 0.001	Linear	-0.961	< 0.001
11	Quadratic	0.984	0.050	Quadratic	-0.981	0.019
Composite	Quadratic	0.952	0.046	-	-	NS

MMG Responses

The MMG AMP responses resulted in significant, negative linear relationships for MMG (r = – 0.938 and –0.978) for 2 of the 11 subjects; a positive linear relationship (r = 0.771) for 1 subject; a negative quadratic relationship (R = – 0.950) for 1 subject; positive quadratic relationships (R = 0.918 to 0.997) for 5 of the 11 subjects; no significant relationship for 2 of the 11 subjects; and a positive quadratic relationship (R = 0.952) for the composite data (Table 2).

The MMG MPF responses resulted in significant negative linear relationships for MMG (r = – 0.633 to – 0.961) for 3 of the 11 subjects; positive linear relationships (r = 0.786 and 0.911) for 2 of the 11 subjects; negative quadratic relationships (R = – 0.981 and – 0.988) for 2 of the 11 subjects; positive quadratic relationships (R = 0.918 and 0.938) for 2 of the 11 subjects; no significant relationship for 2 of the 11 subjects; and no significant relationship for the composite data (Table 2).

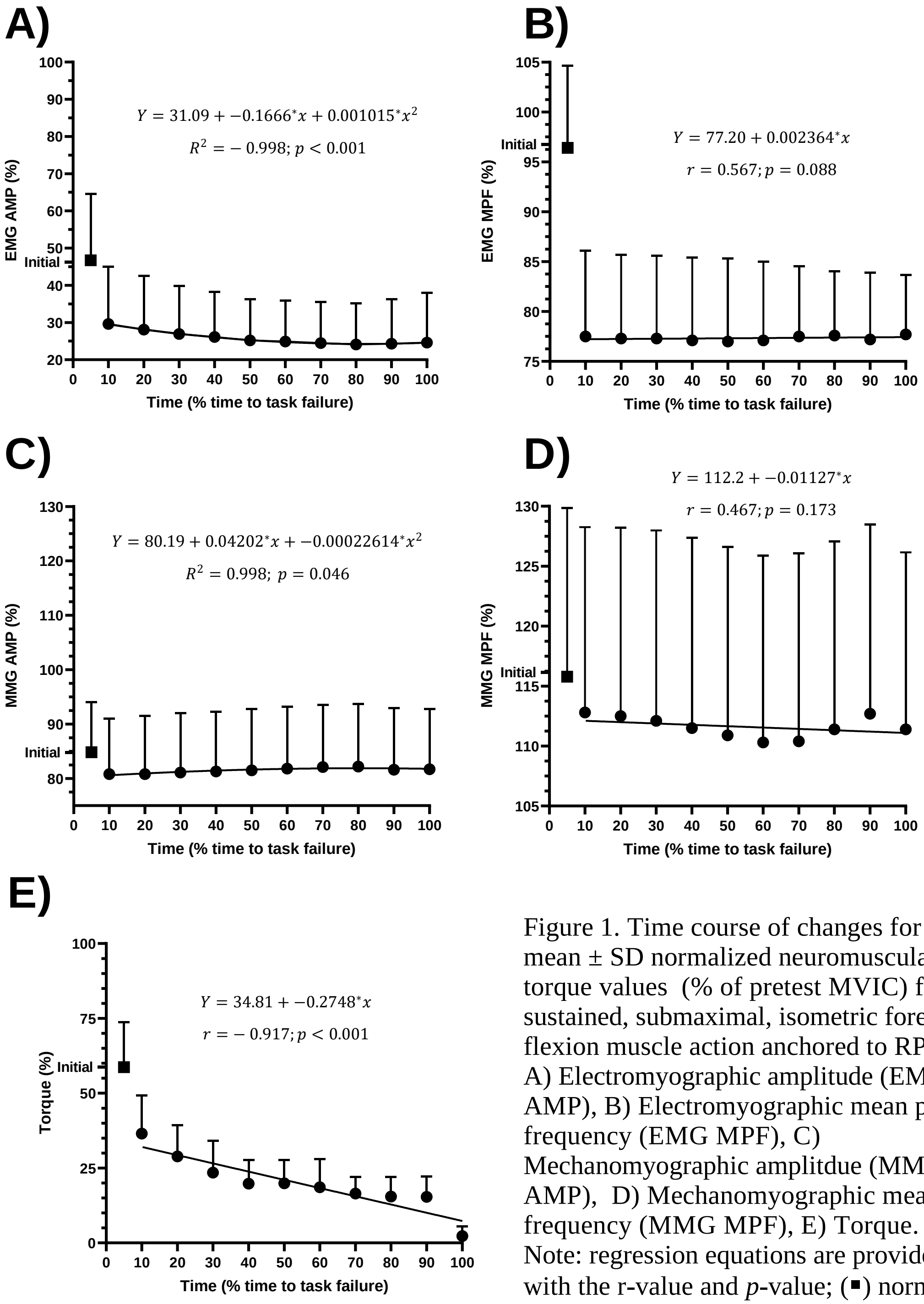

Figure 1. Time course of changes for the mean ± SD normalized neuromuscular and torque values (% of pretest MVIC) for the sustained, submaximal, isometric forearm flexion muscle action anchored to RPE = 7. A) Electromyographic amplitude (EMG AMP), B) Electromyographic mean power frequency (EMG MPF), C) Mechanomyographic amplitdue (MMG AMP), D) Mechanomyographic mean power frequency (MMG MPF), E) Torque. Note: regression equations are provided along with the r-value and *p*-value; (▪) normalized (% of pretest MVIC) initial torque and neuromuscular values of the first 3 s of the sustained isometric muscle aciton.

Discussion

In the present study, the mean ± SD % of MVIC for the initial torque anchored to RPE = 7 was 59.7 ± 15.0 % (Figure 1E). These findings were consistent with previous investigations of isometric forearm flexion (Jones et al., 1982), leg extension (Keller et al., 2018b; Pincivero et al., 2003; West et al., 2005), and leg flexion (Keller et al., 2018b) muscle actions that reported perceived torque (or force) values that were less than the expected % of MVIC. For example, Pincivero et al. (2003) reported that during 5 s, submaximal, isometric, leg extensions, the torque at RPE = 4 to RPE = 9 underestimated the % of MVIC by 2.2 to 14.0 %. Furthermore, Keller et al. (2018b) reported torque values at RPE = 7 of approximately 58.0 – 59.0 % of MVIC during brief, isometric leg flexion and leg extension muscle actions in women. West et al. (2005) hypothesized that the perceptually determined force values when anchored to RPE, underestimated the values associated with an expected % of MVIC due to a subconscious protective mechanism to prevent mechanical and metabolic damage to the active muscle. These studies (Keller et al., 2018b; Pincivero et al., 2003; West et al., 2005), examined the differences in force production anchored to RPE compared to an expected % of MVIC during brief, isometric, muscle actions. Like the present study, however, Keller et al. (2018a, 2019, 2020b) utilized sustained isometric, leg extensions anchored to RPE to examine performance fatigability and neuromuscular patterns of responses. The results indicated that the initial, perceptually determined force values at perceived exertion levels of RPE = 2, 5, and 8, were 3.4 – 13.2 % less than the expected % of MVIC (Keller et al., 2018a, 2019, 2020b). Tucker (2009) suggested that during sustained tasks anchored to RPE, there is an anticipatory setting of the initial torque, mediated through RPE, to promote optimal performance during the task. Therefore, in the present study, the underestimation of torque relative to the expected % of MVIC that occurred at the beginning of the sustained forearm flexion task, was likely due to an anticipatory, self-selected torque to aid in the maintenance of the sustained task (Tucker, 2009).

The present study utilized the RPE Clamp Model of Tucker (2009), and the current subjects under investigation completed a sustained isometric forearm flexion task anchored to RPE = 7 until the torque was reduced to zero, or RPE > 7. During the task, all subjects consciously decreased torque to maintain RPE = 7. For five of the 12 subjects, torque reached zero at task failure (Table 1). It has been suggested that perceived fatigability during sustained exercise is derived from the rates of change in modulating homeostatic and psychological factors such as blood glucose, core temperature, pain, and metabolites that are incorporated into the regulation of exercise intensity to manage fatigue development (Enoka et al., 2016; Kluger et al., 2013). St Clair Gibson et al.

(2006) proposed a model to describe the role of information processing in pacing and perception of exertion during exercise in which the initial anticipatory, self-selected intensity is continuously modulated via feedback from peripheral chemoreceptors and mechanoreceptors that are sensitive to changes in the various modulating homeostatic and psychological factors. St Clair Gibson et al. (2006) further suggested that these modulations of exercise intensity are based on short-term homeostatic responses that occur throughout the task to prevent catastrophic failure during the task and potential damage to the active muscle (St Clair Gibson et al., 2006). It is our interpretation that the risk of catastrophic failure and muscle damage when applying the RPE Clamp Model while the subjects were free to consciously reduce force seems unlikely. Thus, the reductions in torque exhibited in the present study due to perceived fatigability were likely prompted by the adherence of the subjects to the protocol pertaining to the maintenance of RPE = 7, and not fear of catastrophic failure or muscle damage.

The modulating factors underlying perceived fatigability and performance fatigability interact to determine and manage the symptoms of fatigue (Enoka et al., 2016; Thomas et al., 2018). Performance fatigability is dependent on the contractile attributes of the active muscle such as calcium kinetics, metabolism, voluntary activation, activation patterns, and afferent feedback as well as an optimal level of muscle activation to complete the prescribed task (Enoka et al., 2016). In multiple studies, Keller et al. (2018a, 2019; 2020a; 2020b) utilized the RPE Clamp Model to examine performance fatigability and neuromuscular responses during sustained unilateral and bilateral isometric leg extensions in men and women anchored at RPE = 1, 2, 5, and 8. In these studies (Keller et al., 2018a, 2019; Keller et al., 2020a; Keller et al., 2020b), the 13.0 to 47.5 % decrease in MVIC from pre-test to post-test provided a global measure of performance fatigability following the sustained isometric leg extensions. Furthermore, the authors suggested that the fatigue-induced neuromuscular responses may have provided information concerning the underlying mechanisms that were specific to the parameter used to anchor the intensity of exercise (Keller et al., 2018a, 2019). Thus, the application of the RPE Clamp Model by Keller et al. (2018a, 2019; 2020a; 2020b) as well as the present study, together with the assessments of neuromuscular responses, allowed for the examination of the interactions among factors related to perceived fatigability and performance fatigability.

During sustained isometric muscle actions anchored to force, there are predictable patterns of neuromuscular responses that have been used to describe fatigue-induced changes in motor unit activation strategies including increases in EMG AMP (Kimura et al., 2004; Moritani et al., 1986), decreases in MMG AMP (Carr et al., 2018; Kimura et al., 2004), and decreases in the frequency contents of both the EMG and MMG signals (Beck et al., 2007; Carr et al., 2018; Kimura et al., 2004). These neuromuscular patterns of responses have been attributed to the

Onion Skin Scheme and/or Muscle Wisdom (De Luca et al., 1994; Fuglevand et al., 2003). The use of the RPE Clamp Model, however, has demonstrated less consistent patterns of responses than the typical neuromuscular responses when force is used as the anchoring scheme (Beck et al., 2004; Keller et al., 2018a, 2019, 2020b; Smith et al., 2016). Furthermore, the patterns of responses are even less consistent when individual responses are analyzed in combination with composite responses. Previously, Neltner et al. (2020) reported the fatigue-induced neuromuscular responses for both the individual and composite responses based on the recommendations of Anders et al. (2019) due to "…inter-individual variations in fatigue-related changes" (p. 11). The individual patterns of neuromuscular responses in the present study were markedly different from the composite model. For example, the results of the present study indicated a composite negative quadratic relationship for EMG AMP vs. time, although only 4 of the 11 subjects exhibited patterns similar to the composite model. Six of the 11 subjects, however, exhibited an overall decrease (linear or quadratic) in EMG AMP, which suggested there may be a consistent directional change but not a uniform pattern of response when compared to the composite model (Table 2). Furthermore, 1 of 11 subjects exhibited an increase in EMG AMP which was more typical of the fatigue-induced increases in EMG AMP during submaximal isometric tasks (Kimura et al., 2004; Moritani et al., 1986) but was dissociated from the decrease in torque during the sustained isometric muscle action (Table 2). Additionally, 4 of 11 subjects exhibited no changes in EMG AMP which suggested that muscle activation (EMG AMP) was not related to the reduction in torque for those individuals (Table 2). Similar to the present study, Keller et al. (2018a) observed substantial inter-individual variability in EMG AMP during a sustained isometric task anchored to RPE = 5. Keller et al. (2018a) reported that 3 of the 10 subjects exhibited a negative, quadratic relationship, 2 of the 10 subjects exhibited a positive, quadratic relationship, and 5 of the 10 subjects exhibited no relationship for EMG AMP vs. time. Thus, the inter-individual patterns of responses for EMG AMP in the present study, together with those reported by (Keller et al., 2018a), suggested that during a sustained isometric muscle action anchored to RPE, individuals exhibited a variety of motor unit (MU) activation strategies to maintain a constant perception of exertion. Like EMG AMP, there were inter-individual differences in the patterns of responses for EMG MPF in the present study during the sustained isometric muscle action. For example, 3 of 11 subjects exhibited a general decrease (linear or quadratic), 2 of the 11 subjects exhibited a linear increase, and 6 of the 11 subjects exhibited no change in EMG MPF (Table 2). Furthermore, the composite model for EMG MPF in the present study indicated no relationship between EMG MPF vs. time (Table 2). During a sustained isometric muscle action anchored to force, the typical fatigue-induced decreases in EMG MPF have been associated with decreases in muscle fiber action potential conduction velocity due to increased metabolic byproduct buildup within the muscle (Broman et al., 1985). For the 3 of 11 subjects that exhibited decreases in EMG MPF, it may be that a sustained isometric forearm flexion muscle action anchored to an RPE of higher intensity could result in reductions in action potential conduction velocity (Table 2). In previous studies utilizing the RPE Clamp Model during cycle ergometry time trials, Cochrane et al. (2015) and Cochrane-Snyman et al. (2016) observed no relationship in the composite model for EMG MPF vs. time as well as individual variability in the neuromuscular responses. Cochrane-Snyman et al. (2016) hypothesized that the reductions in power output required to maintain a constant RPE may have resulted in decreased metabolic-related but not perceptual-related intensity at RPE levels associated with higher intensities. Therefore, the lack of change in the individual responses for EMG MPF may have been a result of reductions in torque output which caused an overall decrease in metabolic byproducts for those individuals. Similar to previous studies (Cochrane-Snyman et al., 2016; Cochrane et al., 2015), Keller et al. (2018a, 2019) observed substantial inter-individual variability in EMG MPF and reported no changes in the composite models for men and women during sustained isometric leg extensions anchored to RPE = 5. Thus, the findings of the present study, in conjunction with previous investigations (Cochrane-Snyman et al., 2016; Cochrane et al., 2015; Keller et al., 2018a, 2019), highlight the importance of examining composite and individual responses as well as the overall direction of change when "…formulating hypotheses to explain motor control strategies…" (Cochrane-Snyman et al., 2016) during a sustained isometric muscle action anchored to a constant perception of exertion.

In the present study, individuals demonstrated increases and decreases (linear or quadratic) or no changes in MMG AMP and MMG MPF vs. time, while the composite model reflected an increase in MMG AMP only during the sustained isometric muscle action anchored to RPE = 7 (Table 2). The MMG signal is the mechanical counterpart to motor unit activation as measured by EMG and, under fatiguing conditions, MMG AMP can reflect motor unit recruitment, while MMG MPF qualitatively reflects the global firing rate of unfused, activated motor units (Beck et al., 2007). In a previous study, Keller et al. (2018a) observed increases in the composite MMG AMP model during a sustained muscle action anchored to RPE = 5. Keller et al. (2018a) hypothesized that the reductions in force to maintain a constant RPE required less muscle activation (EMG AMP) which decreased muscle stiffness allowing for greater muscle fiber oscillations which has been associated with increases in MMG AMP (Beck et al., 2004). The hypothesis suggested by Keller et al. (2018a) only partially supported the findings of the present study. The Muscle Wisdom theory (Fuglevand et al., 2003), however, may explain the motor unit activation strategies (increases in MU recruitment and decreases in firing rates) demonstrated by 5 of 11 subjects in the present study (Table 2). Muscle Wisdom theory is considered to be the economical activation of the fatigued muscle by the central nervous system, which is characterized by concurrent reductions in force (or torque), increases in MU recruitment (MMG AMP), relaxation rate, and decreases in the firing rate of active MUs (MMG MPF) (Enoka et al., 1992; Fuglevand et al., 2003). Thus, the Muscle Wisdom theory (Fuglevand et al., 2003) may provide an explanation for the increases in MMG AMP and decreases in MMG MPF expressed by those individuals in the present study. According to the Onion Skin Scheme (Contessa et al., 2016), there is an increase in the firing rates of already activated MUs, while simultaneous recruitment of higher threshold MUs occurs. In addition, the later recruited high threshold MUs have lower firing rates than the MUs that were initially recruited. In the present study, 4 of 11 subjects exhibited decreases (linear or quadratic) in EMG AMP and MMG AMP, along with increases (linear or quadratic) in MMG MPF. Perhaps the decreased muscle activation (EMG AMP) resulted in derecruitment of the higher threshold MUs (MMG AMP) which would have resulted in an increase in MMG MPF due to the increased firing rates of the already activated lower threshold MUs. Thus, the motor unit activation strategies for the 4 of the 11 subjects may be explained, in part, by the Onion Skin Scheme (Contessa et al., 2016). Taken together, the findings of the present study indicated substantial differences in the inter-individual neuromuscular responses and the individual responses compared to the composite models which extends the previous suggestions of Anders et al. (2019) to studies utilizing the RPE Clamp Model. Furthermore, when sustained isometric muscle actions are anchored to RPE to examine the perception of fatigue, there is the confounding influence of a conscious decision to decrease torque. Thus, the neuromuscular patterns of responses associated with the RPE Clamp Model can be affected not only by physiological manifestations from central and peripheral fatigue, but also by the conscious decrease in torque and reductions in central drive which complicates inferences regarding fatigue-induced changes in motor unit activation strategies.

Several models have been proposed to explain the underlying mechanisms responsible for the regulation of RPE during exercise (Marcora, 2009; Noakes, 2012; Pageaux, 2016; Tucker, 2009). In a previous review, Pageaux (2016) discussed the characteristics of three prominent models that are based on feedback (Afferent Feedback Model), feedforward (Corollary Discharge Model), and a combination of feedback and feedforward mechanisms (Combined Model) for the regulation of RPE. Proponents of the Afferent Feedback Model have postulated that during exercise, the feedback from group III/IV free nerve endings relay information to areas of the brain related to central motor command and, that RPE is continuously adjusted based on this feedback from the active muscle (Amann et al., 2020; Pageaux, 2016). The Corollary Discharge Model of Marcora (2009) proposed that an efferent copy of the central motor command (i.e., corollary discharges) is sent from the dorsolateral prefrontal cortex to the premotor cortex and the somatosensory areas to provide the information necessary to generate the corresponding RPE associated with the efferent motor output (Abbiss et al., 2015; Marcora, 2009). Proponents of the combined model have suggested that during exercise, the feedback from group III/IV muscle afferents is integrated with the corollary discharges to provide the necessary information to generate the corresponding RPE (Pageaux, 2016). Furthermore, Pageaux (2016) acknowledged that although experimental evidence supports the validity of the combined model (Amann et al., 2010), perception of exertion remained unchanged with reduced afferent feedback during a cycling task (Kjaer et al., 1999) which suggested that this model was improbable. Previously, Zénon et al. (2015) demonstrated that RPE is regulated within higher brain centers upstream from the motor cortex. During hand-grip contractions at multiple intensities (10, 23, 37, and 50 % of MVC), the authors recorded the subjects perception of exertion while disrupting neural activity in the primary motor cortex (M1) and supplementary motor areas (SMA) of the brain with continuous theta-burst transcranial magnetic stimulation (Zénon et al., 2015). The authors reported that disruption of the SMA activity, but not M1 activity, resulted in alterations in the perception of exertion which suggested that the SMA and surrounding regions upstream were responsible for the generation of perception of exertion, not the motor cortex (Zénon et al., 2015). Thus, the Corollary Discharge Model was not supported by the findings of the present study due to the voluntary reduction in torque required to consciously maintain a constant RPE in addition to the experimental evidence that suggested RPE is regulated upstream from the motor cortex (Zénon et al., 2015). Furthermore, afferent feedback, which has been discussed previously (Keller et al., 2018a; Marcora, 2009; St Clair Gibson et al., 2006; Tucker, 2009), likely contributed to the regulation of RPE during the sustained isometric muscle action in the present study but did not completely generate the perception of exertion.

The findings of the present study supported the RPE Clamp Model of Tucker (2009) which suggested that the initial torque and RPE were generated from an anticipatory feedforward mechanism, and then the torque was consciously adjusted due to afferent feedback to maintain the predetermined RPE. During the sustained isometric forearm flexion muscle action anchored to RPE = 7 in the present study, a similar level of muscle activation (EMG AMP) was required to produce less torque throughout the task which has been suggested to be caused by excitation-contraction coupling failure within the muscle (Enoka et al., 1992). Therefore, we hypothesized that torque and RPE were derived from an anticipatory feedforward mechanism and then continuously adjusted throughout the sustained isometric task by afferent feedback to maintain the predetermined

RPE, and the task was ultimately terminated due to a disruption between activation of the muscle and torque production (Enoka et al., 1992; Tucker, 2009).

A limitation in the present study included utilizing only one perceptual level of intensity (RPE = 7). Perhaps, multiple levels of perceived exertion (e.g., RPE = 1, 3, 5, and 7) would have resulted in more consistent individual patterns of neuromuscular responses. In addition, the present study only examined the neuromuscular responses in women. It is possible that there could be sex differences in both the torque and neuromuscular responses. Finally, there were no pre-test vs. post-test assessment MVICs in the present study. The examination of the percent difference in MVIC from pre-test to post-test would have provided an objective global measure of fatigability.

In summary, the findings of the present study demonstrated that, during a sustained isometric forearm flexion muscle action anchored to RPE = 7, EMG MPF, MMG AMP, and MMG MPF remained unchanged. There were, however, mean decreases in torque and EMG AMP with an increased ratio between muscle activation (EMG AMP) and torque output that was likely the result of excitation-contraction coupling failure (Enoka et al., 1992). Caution should be used, however, when interpreting the composite neuromuscular responses due to the substantial individual differences as evidenced in the present study. In addition, the present findings supported the RPE Clamp Model of Tucker (2009) and suggested that torque production was initially regulated by an anticipatory feedforward mechanism and then continuously adjusted by afferent feedback to maintain RPE = 7. Future studies should seek to examine performance fatigability and perceived fatigability as well as the individual and composite neuromuscular responses during a sustained isometric muscle action anchored across multiple levels of perceived exertion.

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