Variation of Force Amplitude and its Effects on Muscle Fatigue

by

Marcus Yung

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Authors Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Abstract

Current trends in industry are leaning towards specialized production systems and sedentary computer work tasks that are associated with low and less varying mechanical exposures. It has been suggested that physical variation is an effective intervention to reduce local fatigue and potentially musculoskeletal disorders. However, little is known in how the differences between physical variation patterns affect physiological and psychophysical responses. The general purpose of this thesis was to explore the biophysical effects of varying force amplitudes using forces, cycle times, and duty cycles that are relevant to occupation and longer-term health outcomes.

Fifteen healthy males performed an elbow extensor sustained isometric exertion at 15% Maximum Voluntary Force (MVF), an intermittent contraction between 0% MVF and 30% MVF (On/Off), an intermittent contraction between 7.5% MVF and 22.5% MVF (MinMax), an intermittent contraction between 1% MVF and 29% MVF (1 Percent), and a sinusoidal contraction between 0% and 30% MVF (Sinusoidal). Eight commonly used measurement tools recorded biophysical responses as participants performed each condition for up to 60 minutes or until exhaustion, during 60 minutes recovery, and at 24 hours post-exercise. Measures included electromyography of the triceps muscles, mechanomyography, blood flow, heart rate, stimulated tetani and twitch responses, maximum exertions, and perceived exertion. The rate of response during exercise and comparisons between baseline, cessation, and recovery values were used to assess fatigue responses.

First, this thesis addressed whether physical variation delayed fatigue response when compared to a sustained low-force sustained isometric contraction. The On/Off contraction led to a slower fatigue response in 25 of 32 measurement parameters. On/Off (Median = 3600 seconds, 25th = 2274, 75th = 3600) led to a longer endurance time than the sustained isometric (Median = 579 seconds, 25th = 408, 75th = 1190.50), p = 0.003. MMG root mean square values revealed a greater rate of response during a sustained isometric (M = 8.514%/min, SD = 8.525) than On/Off (M = 0.979%/min, SD = 1.669), p = 0.002. In recovery, the sustained isometric exertion led to a long-

lasting fatigue effect. The On/Off intermittent contraction also led to long-term fatigue response despite the condition not completed until exhaustion.

Second, this thesis investigated the effects of varying intermittent contractions (MinMax, 1 Percent, Sinusoidal) and their relationship to sustained isometric and On/Off intermittent contraction conditions. The MinMax condition showed exercise responses that were between those of sustained isometric and On/Off conditions, with 8 of 32 measurement parameters that were significantly different from sustained and 12 of 32 measurement parameters that were significantly different from On/Off. The 1 Percent condition resulted in exercise responses that were similar to the On/Off contraction during exercise, with 16 of 32 measurement parameters that were statistically different from sustained. The Sinusoidal contractions resulted in delayed fatigue response during exercise as there were 16 of 32 measurement parameters that were significantly different from sustained and 2 of 32 measurement parameters that were statistically different from On/Off. Both MinMax (Median = 1474 seconds, 25^{th} = 694, 75^{th} = 2901, $p = 0.003$) and Sinusoidal (Median = 2205 seconds, 25^{th} = 711, 75^{th} = 3600, $p = 0.006$) led to shorter endurance times when compared to On/Off. 1 Percent (Median = 3202 seconds, 25^{th} = 650, 75^{th} = 3600, $p = 0.005$) and Sinusoidal ($p = 0.009$) led to longer endurance times than sustained isometric. 1 Percent (M = 6.501%/min, SD = 7.756, $p = 0.000$), MinMax (M = 10.853%/min, SD = 9.446, $p = 0.003$), and Sinusoidal (M = 9.484%/min, SD = 11.108, $p = 0.001$) led to slower rates of perceived exertion than the sustained isometric (M = 18.237%/min, SD = 12.873). MinMax ($p = 0.022$) led to a quicker increase in ratings of perceived exertion when compared to On/Off (M = 5.287%/min, SD = 6.195).

This research shows that implementing physical variation, at the same mean amplitude, may provide reduced fatigue rate and that the magnitude and shape of the intermittent force variations affect exercise and recovery measures. Time varying forces may therefore provide the necessary mechanism to encourage beneficial physiological responses that would improve long-term health and well being of workers at low-load jobs.

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Thesis Aims and Outline

The general theme of this thesis was to investigate mechanical variability and the effects of diversity in delaying or preventing the onset of fatigue. The introduction (Chapter I) will describe the rationale for the study and introduce the purpose and hypotheses. The review of literature (Chapter II) will facilitate the understanding of the need of mechanical variation in occupational tasks and lay the foundation for the current issues and limitations from past research. The third chapter will discuss the common methodology in Chapters IV & V. Specific details will be discussed in each corresponding chapter. Finally, an overview and integration of the main findings will be presented in the final section (Chapter VI). Additionally Chapter VI will discuss the implications and potential contributions of the findings in a general context. Specific description of the chapters is as follows:

Chapter I — An introduction to the rationale and current issue of mechanical variation as applied to occupational settings.

Chapter II — Reviews the importance of mechanical variation in occupational tasks, lays the foundation for the current issues and limitations from past research, and reviews literature important for the choice of independent and dependant variables.

Chapter III — This section presents the general methodology used in Chapters IV and V. Specific details will be discussed in the relevant chapter.

Chapter IV — This section examines the central postulate that mechanical variability differs from an isometric static exertion. The effects of an isometric contraction will be compared to the classical intermittent contraction pattern (activation/relaxation) for a 1-hour exercise period followed by 1 hour of recovery and a 24-hour post-exercise follow-up.

Chapter V — Response measures are compared between exercise protocols of diverse force levels. The central theme of this chapter is to understand the effects of varying the force between an isometric contraction and classical intermittent contraction pattern based on statistical expressions of diversion (standard deviations). Additionally, a comparison between a mechanical sinusoidal wave pattern and a square wave pattern will be discussed to answer the question, "Does the shape of the force pattern influence muscle response?"

Chapter VI An overview and integration of the major findings as well as the implications and potential contributions to occupation and prevention of work related musculoskeletal disorders.

Chapter I

Introduction

Mechanical variation, through research by Mathiassen (2006), has been described as "the change in exposure over time." It is associated with the quantity and the frequency of changes - that always occur across time in occupational settings - and whether recurring elements are exhibited. Introducing mechanical variation to a task or job is neither a new concept nor practice in the prevention of fatigue and Musculoskeletal Disorders (MSD) at work. In fact, interventions that promote variation include job enlargement, job rotation, changing work patterns, and increased break allowances. But what is the basis for such interventions and similar preventative strategies thought to reduce MSDs?

Work-related musculoskeletal disorders (MSD) are an overwhelming concern in both Canada and the United States. The Workplace Safety Insurance Board of Ontario reported 43% of all lost time injuries as a MSD in 2007. Similarly, according to the US Bureau of Labor Statistics, 29% of all cases involving days away from work were WMSD related. Consequently, over the past few decades, much research has been devoted to guidelines and strategies to reduce MSDs.

In general, preventative measures are based on exposures that optimize an acute physiological, psychological, and biomechanical response (Westgaard & Winkel, 1996). A key aspect of these approaches is the definition of acceptable exposures, breaks, and work rest patterns. Mathiassen (2006) suggested that breaks might not be related to rest but rather influences the overall exposure variation. Accordingly effective breaks should be redefined as varying exposure rather than complete rest. A new paradigm is thus required to provide appropriate physical stresses that would benefit the worker with loads sufficiently vigorous to trigger a positive adaptation.

Purpose

The specific purposes of this research study were:

1. To compare fatigue responses of contraction patterns consisting of variation about a mean force amplitude based on a metric of diversity.
2. To compare fatigue responses of a sinusoidal contraction pattern and a square wave pattern.

Sub-Problems

1. A challenging area is the development of fatigue in exposure patterns of great complexity, as extrapolation of simple exposure patterns typically tested to occupational tasks may not be straightforward. These laboratory tests also typically occur over short time scales and many of the acute responses recover rapidly (de Oliveira Sato & Cote Gil Coury, 2009; Nussbaum et al., 2001). As such, this study will require participants to exercise at determined forces until exhaustion or up to 60 minutes.
2. Typical levels of force used in laboratory studies are within higher ranges. These may not be able to create physiological effects observed in occupationally relevant load levels that are maintained for longer durations. Westgaard and Winkle (1996) suggest that low force ranges, most indicative of fatigue and longer-term health outcomes, have infrequently been the focus of laboratory studies. For instance, Tami et al. (2003) induced fatigue fracture on the rat ulna bone using a single bout of repetitive loading; cycling at mean sub-threshold levels. Although the study replicated the classical material fatigue response, it did not replicate a physiological bone fatigue fracture. This study will use forces, frequencies, cycle times, and duration profiles that are common to work and the development of occupational-related injury.
3. Various tools and procedures have been used to assess fatigue; each may provide limited information about specific processes and mechanisms in the activation chain for fatigue (Vøllestad, 1997). A 'gold standard' to assess human muscle fatigue may be difficult to identify as to whether

muscle fatigue occurred or not. This study will use a number of measures, including electromyography, mechanomyography, muscle blood flow kinetics, low frequency fatigue, muscle twitch force, perceived exertion, endurance time, and force output to assess and measure different mechanisms of fatigue.

Hypotheses

Mechanical Variation and Diversity:

1. The classical intermittent isometric contraction pattern, a repeated cycle of zero mechanical force and 30% of the participant's maximum voluntary force, +/-1 from mean force amplitude, will show a lower rate of fatigue response when compared to a submaximal sustained isometric contraction. (One-tailed test)
2. Activity with +/- ½ mean force amplitude (between 7% and 22.5% of the participant's maximum voluntary force) will show:
    a. A slower rate of fatigue response when compared to a submaximal sustained isometric contraction. (One-tailed test)
    b. A quicker rate of fatigue response than the classical intermittent contraction pattern. (One-tailed test)
3. Exposure to forces between 1% and 29% of the participant's maximum voluntary force will result in:
    a. A slower rate of fatigue response when compared to a submaximal isometric condition. (One-tailed test)
    b. A quicker rate of fatigue response than classical intermittent on/off contraction pattern
4. Sinusoidal wave patterns will show:
    a. A slower rate of fatigue response when compared to a submaximal sustained isometric contraction. (One-tailed test)

b. A quicker rate of muscle fatigue response when compared to the classical intermittent contraction pattern. (One-tailed test)

Additional Issues

The data collected should also allow a discussion of the following issues:

1. Based on the definition of muscle fatigue described by Vøllestad (1997), as the "exercise-induced reduction in the maximal capacity to generate force or power output", maximum voluntary force and endurance time may be a good direct indicator of fatigue.
2. Muscle twitch and low-frequency force assessments may be a good indicator of the loss of force generating capacity. These assessments may coincide with a reduction in maximum voluntary force production.
3. Electromyography and mechanomyogaphy have both been used as indirect tools to measure physiological responses accompanying fatigue. Changes in the time domain and frequency spectra have been widely used parameters during fatigue from prolonged exercise. It is generally accepted that mechanomyography is a more sensitive measure than electromyography. Both electromyography and mechanomyography will show shifts in both amplitude and frequency.
4. Ratings of perceived exertion and heart rate will show increased responses with time for all conditions.

Chapter II

Review of Literature

Prevention of MSDs

Guidelines have also been established based on thresholds for tissue injury. In vitro studies on isolated tissue specimens have been conducted to identify mechanisms of failure under both static and cyclic loading (Parkinson & Callaghan, 2007; Tampier et al., 2007). Mueller & Maluf (2002) suggest a range of physical outcome (tissue death, decreased tolerance, maintenance, increased tolerance, injury, tissue death), a composite value defined by magnitude, time, and direction of application that incur a particular outcome.

Ideally, a primary prevention approach, by applying optimal designs to the working environment and equipment or task should be used. These design guidelines may be developed from psychophysical, physiological, or biomechanical studies. Engineering solutions, if properly implemented, should reduce the risk of injury or illness (Amell & Kumar, 2001). However, systematic reviews (Brewer et al., 2006) and intervention studies (Gerr et al., 2005; Amick et al., 2003) on computer workstations showed mixed results for the effectiveness of workplace interventions. Despite evidence of the beneficial effects of engineering solutions, compliance towards its use may be compromised (Evanoff et al., 2003). For instance, nursing personnel did not routinely use mechanical lifting and transfer devices, even if available (Evanoff et al., 2003). The lack of compliance may be attributed to the additional time, physical effort, training, and lack of space required to use the intervention (Evanoff et al., 2003).

Optimal design at the task level may involve a reduction or change in the mechanical exposure. An exposure reduction of at least 14% resulted in a concomitant improvement in musculoskeletal health (Lotters & Burdof, 2002). As such, this discussion returns to the implementation of mechanical exposure variation as a preventative strategy to reduce WMSDs.

Guidelines for Prevention

A popular and frequently used set of guidelines to create so-called acceptable exposures is based upon the load-adjust psychophysical approach. This approach has been used extensively in manual materials handling literature (Snook et al., 1995) and in the upper limbs (Ciriello et al., 2001; Potvin et al., 2006). Electromyography (EMG) has been used to create maximum voluntary contraction (MVC) upper limit guidelines (Jonsson, 1978). For instance, studies have shown a high risk of musculoskeletal disorders in repetitive work tasks at median load levels less than 6% MVC (Jensen et al., 1993a; Veiersted et al., 1990). Jensen and colleagues (1993a) observed pain symptoms in the upper trapezius muscle at a median load of 5.3% MVC and 4.1% MVC for production and office workers, respectively. Veiersted et al., (1990) conducted a field study to investigate patterns of upper trapezius muscle activity during standardized and machine-paced packing tasks. According to Veiersted and colleagues (1990), workers with neck and shoulder pain demonstrated a static median level of 1.9% MVC. Bystrom & Kilbom (1990) suggested an upper limit of 16.7% MVC for 60 minutes at an intermittency of 10 + 5 second contraction-relaxation periods. However, intermittent contraction patterns are inconsistent from one task to another and different time history patterns may affect the development of fatigue (Mathiassen, 2006; Visser & van Dieen, 2006) or low back pain (Krajcarski & Wells, 2008).

Mechanical Exposure Variation

Past literature suggests the importance of the force, duration, and frequency profile in the development of musculoskeletal disorders. This is particularly true in industry as outsourcing (Mathiassen, 2006) and automation of work processes (de Looze et al., 2009) leads to tasks with lower and less varying mechanical exposure levels. Process strategies, i.e. lean manufacturing, modify the nature of work, changing the physical exposure patterns with work schedules tailored to production demands while increasing the versatility of the workers (de Oliveira Sato & Cote Gil Coury, 2009). As the number of workers decrease, the repetitive and less varying load on workers increase (de Oliveira Sato & Cote Gil Coury, 2009). The risk of musculoskeletal disorders is attributed to the increased occurrence of short-cycle and repeated tasks.

One effective ergonomic intervention is the implementation of more mechanical exposure variation to improve performance and muscle adaptation. Initiatives often include, job enlargement, job rotation, increase break allowances, and more diversified tasks/multifunctional jobs.

Job Enlargement

Job enlargement allows increased physical variation by exposing workers to a greater number of different tasks, each that may be considered repetitive, similar-level, and not recommended for long periods of time (Moller et al., 2004; Campion et al., 2005). An alternative use of job enlargement is to increase the motivational value of a job, leading to greater self-efficacy (Campion et al., 2005). Contrasting results in literature give rise to confusion as to whether job enlargement is an effective implementation. Enriched and enlarged jobs may result in greater job satisfaction and mental health (Lin et al., 2007; Søgaard et al., 2006) and cycle-to cycle physical exposure variability (Moller et al., 2004). On the other hand, job enlargement may not be sufficient in reducing overall physical workload if the strain profiles on the body are similar between tasks (Søgaard et al., 2006). From a practical perspective, job enlargement may not lead to improved health among all workers if individuals benefiting from more diversified tasks do so at the expense of others (Moller et al., 2004). Additionally, job enlargement may not be supported if it affects salary systems and other workplace structure (Moller et al., 2004).

Job Rotation

Similar to job enlargement, job rotation is often implemented to alleviate physical stress and fatigue by rotating workers between jobs/tasks that use different muscle groups and exposure patterns (Wells et al., 2009; de Oliveira Sato & Cote Gil Coury, 2009; Frazer et al., 2003). Spreading the load over several workers, the loads are averaged but the peak is experienced by all and the overall risk in the working population is increased (Frazer et al., 2003). Potential benefits also include mitigating boredom, employee learning and improved versatility, reduced absenteeism, improved worker retention, morale building, increased production, and employer learning (Eriksson & Ortega, 2006; Jorgensen et al., 2005; Frazer et al., 2003). Job rotation was shown to decrease physical workload based on %VO2 max and perceived exertion and decrease mental workload based on excretion rate of adrenaline (Kuijer et al., 2004). However, this is not often the case, where the redistribution of risk is not uniform and may not be sufficient to reduce high peak and cumulative risk factors (Frazer et al., 2003; Kuijer et al., 2004). If there are improvements to physical and mental workloads, it may not directly imply a reduction in workload due to a reduction in work demands (Hsie et al., 2009). Similarly, there is very limited support for the employee motivation hypothesis (Eriksson & Ortega, 2006). Implementation of job rotation has limitations including rotating workers with medical restrictions, limited number of jobs to rotate to, and a decrease in product quality (Jorgensen et al., 2005).

Rest Allowance Models

Much research has focused on rest allowance models to reduce fatigue in muscle groups thereby reducing the risk of musculoskeletal disorders. Introducing micropauses of minimal or zero activity, without prolonging exposure time, or changing the work pattern may prevent musculoskeletal disorders (Mathiassen, 2006; Bystrom et al., 1991), however Mathiassen (2006) and Bystrom et al., (1991) comment that there is a lack of evidence to support this. Ideally, a rest allowance model should provide a safeguard from overexertion while considering time constraints (Hsie et al., 2009). Different rest allowances will result in varying physiological and perceptual responses that in turn may or may not result in a musculoskeletal disorder. Cutlip et al. (2009) suggested that both short (10 seconds) and long (5 minutes) rest times resulted in injury and significant deficits in performance. Insufficient rest between work bouts hindered the recovery of a particular tissue as the rate of damage exceeds the rate of repair. On the other hand, long rest times allowed tissues to generate higher forces during subsequent work bouts, increasing the risk of injury (Cutlip et al, 2009). In a 20-minute protocol of intermittent exercise, a long work to rest duration elicits greater metabolic and perceptual strain than a short work to rest duration (Price & Moss, 2007). It was speculated that work duration was related to blood lactate concentration and perceived effort was associated with both blood lactate and bicarbonate concentrations (Price & Moss, 2007). El ahrache and Imbeau (2009) compared and addressed the practical applications of four commonly used rest allowance models: Rohmert, 1973, Milner, 1985, Rose et al., 1992, and Bystrom and Fransson-Hall, 1994.

The study found substantial discrepancies in the time required for adequate rest between the four models, citing Rose et al. model as the most conservative and the Rohmert recommending the least amount of rest. Although the Rohmert model is the most cited, it is based on the hypothesis of infinite endurance at exertions less than 15% MVC (El ahrache, 2006). Determining rest allowances is not straight forward as models differ in muscle specificity, approach (subjective, physiological, psychophysical), and sampling procedures and populations (El ahrache, 2006). The application of a work-rest model is often limited to specific task conditions (Iridiastadi & Nussbaum, 2006). It has also been suggested that if exposure to work is varying, breaks may have marginal effects (Mathiassen, 2006). And have been argued that using such work-rest ratios may not effectively protect workers from the risk of developing a WMSD (Mathiassen & Winkel, 1992). This may add to the confusion as to the appropriate rest allowance model for a particular task.

Exposure Amplitude Variation

According to Veiersted et al. (1990) the pattern of muscle activity during tasks are probably more important that the total duration of the working period in the development of musculoskeletal complaints. Chronic injuries are typically observed at forces less than 10% MVC for long durations (Westgaard & Winkel, 1996) and have led to hypotheses to the stereotyped recruitment patterns. One such theory is the Cinderalla hypothesis where a fraction of motor units are active at sub-maximal levels. Low threshold motor units may become overloaded, leading to injury during prolonged and low-level static work (Hägg, 2000). Technological advancements have led to lower physical workloads and an increase in sedentary computer work tasks, which may be detrimental to short- and long-term health (Straker & Mathiassen, 2009). Mathiassen (2006) suggested that breaks might not be related to rest but rather influences the overall exposure variation. Accordingly effective breaks should be redefined as varying exposure rather than complete rest. A new paradigm is thus required to provide appropriate physical stresses that would benefit the worker with loads sufficiently vigorous to trigger a positive adaptation.

Mathiassen and Christmansson (2004) speculate that workers that vary their posture and/or loads between task cycles are less prone to develop musculoskeletal disorders. This variation is about an average amplitude that occurs during the task. Mechanical exposure variation is implemented to allow motor units, that would be otherwise overloaded, an opportunity to relax. This can be achieved by transferring the load to other muscles or to other motor units in the same muscle or by changing the timing and size of the load. Exposure variation analysis (EVA) quantifies the relative time that EMG activity is distributed within specified amplitude intervals or classes (Mathiassen, 2006) and has shown that variation reduces the risk of developing injury (Jensen et al., 1999).

The alternative is to include microbreaks of minimal or zero activity as part of the Cinderella hypothesis and EMG gaps theory (Eksioglu, 2006; Veiersted et al., 1990). Microbreaks may be quantified as EMG activity lower than 0.5% MVC for 0.2 seconds or longer (Veiersted et al., 1990). The low detection level was chosen due to the hierarchal recruitment patterns of motor units where at low level muscle activity some motor units may be continuously activated (Veiersted et al., 1990; Hägg & Aström, 1997). Veiersted and colleagues (1990) found that muscle activity below 1% MVC, observed with EVA, may be a good indicator of muscular rest. Patterns with EMG gaps of short duration led to fewer musculoskeletal complaints than patterns with continuous activity (Veiersted et al., 1990; Hägg & Aström, 1997; Jensen et al., 1999).

However, overall, past literature has revealed weak empirical evidence of increased physical variation in both intervention and epidemiological studies. Very few studies focus on longer-term effects of job rotation or job enlargement while existing research have shown inconsistent results (Mathiassen, 2006). Evidently, there is a need to understand the optimal variation to be applied into occupational settings. "Healthy" patterns of work, based on force amplitude, must be determined and prescribed to improve long-term health outcomes (Straker & Mathiassen, 2009; Christensen et al., 2000). Wells and colleagues (2009) devised an approach to quantify functional similarity of tasks that can be used to create sufficient task diversity within a job or job rotation scheme. Using three isometric gripping tasks, Wells et al., (2009) found alternation between lateral pinch and power grips to be sufficient in creating varying mechanical exposure at 30% MVC.

What is less known is how diversity, the extent in which exposure entities differ, between physical variation patterns affect both physiological and psychophysical responses. Further investigation is also required to understand the effectiveness of mechanical variation at low-level forces that are attributable to longer-term health effects (Nussbaum et al, 2001; Westgaard & Winkel, 1996; Mathiassen & Winkel, 1992; Mathiassen, 1993). Thus the following study was undertaken to address the question of diversity and variation and its effectiveness in delaying or preventing the onset of fatigue using a number of response measures.

So How Does Muscle Fatigue Relate to Longer Term Health Outcomes?

The etiology of WMSD can be conceptualized with a progressive stepwise model (Figure 2.1) that begins with an external exposure and manifests itself as an injury or adaptive response that may be acute, cumulative, or chronic in nature. This model has been adapted from Wells et al. (2004), Sjogaard & Søgaard (1998), and Westgaard & Winkel (1996).

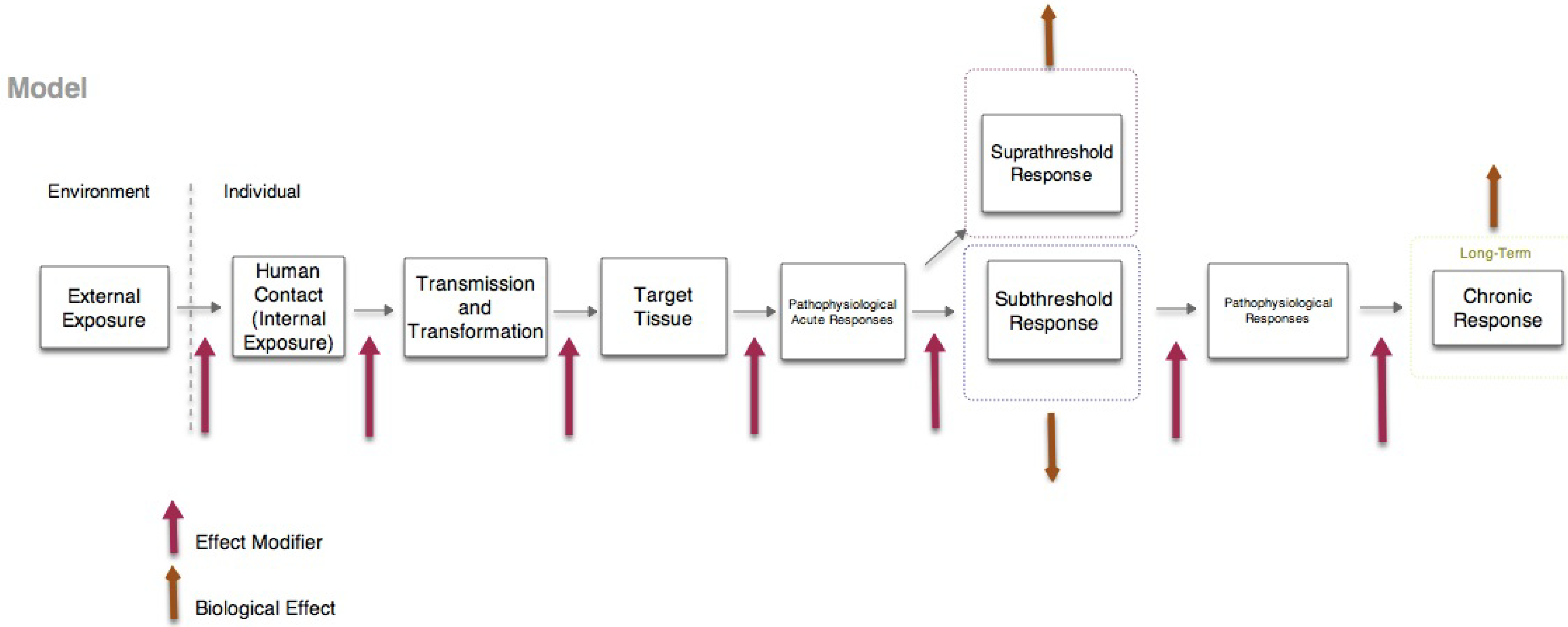


Figure 2.1 Exposure-Tissue-Response Model
Model conceptualizing the progression of a WMSD from external exposure to target tissue to response. A supra-threshold response occurs when tolerance threshold is exceeded after a single incident/exposure. Sub-threshold stimuli may accumulate and result in a cumulative response. These responses can be measured. Prolonged exposure of subthreshold stimuli may incur a long-term outcome or a chronic response. Responses can be adaptive (increased/decreased tolerance, no functional changes) or result in an injury. Diagram adapted from Wells et al. (2004), Sjogaard & Søgaard (1998), and Westgaard & Winkel (1995).

External and internal exposures are transmitted and transformed through the body to a biological tissue (Wells et al., 2004). These target tissues exhibit injury or adaptive effects. Søgaard et al., (2003) suggested that acute responses are possible precursors to pain and disorder. Such acute responses include increases in motor unit recruitment, motor unit firing, increases in metabolite accumulation, reduced local tissue oxygenation, etc. These responses may lead to yielding, breaking, rupturing, and deformation of tissue if exposure surpasses a threshold (supra-threshold). Sub-threshold exposures that are continuous or repetitive may result in reduced performance, discomfort, pain, or fatigue. Longer-term health effects include cumulative fiber microtrauma that may result in pain, inflammation, edema, and ultimately disorder or malfunction (Sandrey, 2000).

In this study, acute responses are biomechanical, physiological, and neurophysiological in nature and can be measured using a variety of tools. These acute responses may predict a longer-term health outcome such as localized muscle fatigue. In turn, localized muscle fatigue may be an indicator of a long-term physiological process that may result in musculoskeletal disorders (Nussbaum et al., 2001) and considered a representative or surrogate indicator of injury risks (Iridiastadi & Nussbaum, 2006). Subsequently, predicting the onset of fatigue by identifying and relating the acute responses may be used for the basis for ergonomic guidelines (Nussbaum et al., 2001).

Although the conceptual model applies to all biological tissues, discussion will be limited to tissues identified as most relevant to work-related musculoskeletal disorders. Statistical data compiled from four government agencies overseeing workplace health and safety is shown in Table 2.1. The five tissues that comprise the majority of WMSD in both Canada and the United States are: (1) Tendon, (2) Muscle, (3) Nerve, (4) Bone, and (5) Ligaments (associated with sprain). Responses of all five tissues are shown in Table 2.2.

Table 2.1 Percent of lost time claims/cases based on nature of injury/illness. By type of tissue and agency.

Tissue	Injury/Illness	WSIB	US BLS	WorkSafe BC	Wash. BLS
Tendon	Tendonitis	2.3%	0.4%	2.0% (related)	2.1%†
	Tenosynovitis	0.2%	-		-
	Ganglion	0.1%	-	-	-
Muscle	Myositis	0.0% (5 cases)	-	-	-
	Rotator cuff tear/syndrome	0.4%	-	-	-
	Back pain, hurt back	0.0% (26 cases)	3.2%	22.4% (strain)	2.9%†
	Hernia	0.9%	4.6%†	0.8%	-
Nerve	Carpal Tunnel Syndrome	0.9%	1.0%	-	2.5%
	Sciatica	0.2%	-	-	-
Bone/Joint	Capsulitis	0.0% (20 cases)	-	-	-
	Facet Syndrome	0.0% (37 cases)	-	-	-
	Fractures	7.0%	8.2%	7.1%	6.6%
	Intervertebral herniated, slipped disc including disc syndrome	1.1%	-	-	-
	Dislocation	1.0%	-	0.6%	-
	Epicondylitis	0.7%	-	-	-
Other	Sprain and Strain*	49.1%	38.7% (+ tears)	32.3% (Other)	48.4%
	MSK system and connective tissue, diseases and disorders unspecified	0.5%	2.8% †	5.9% (Occupational disease)	-
	Soreness, pain, hurt, except the back	0.1%	6.8%	0.1% (other injuries)	4.0%

Workplace Safety & Insurance Board of Ontario - % of allowed lost time claims. May include both traumatic and repetitive motion types of injuries and diseases. 2007 Annual Report.
US Department of Labor Bureau of Labor Statistics - % of all cases involving days away from work. 2007 Annual Report.
WorkSafe British Columbia - % of short-term disability, long-term disability, and fatal claims first paid in 2007.
Washington State Department of Labor & Industries - % of MSD by nature of illness, Washington private industry

* Sprain: Stretching/tearing of ligaments. Strain: Stretching or tearing of muscle or tendon.
† % of MSD cases. Percentage is an overestimation.

Table2.2Acute, cumulative, chronic responses according to tissue type.

Tissue	Supra-Threshold Response	Cumulative Sub-Threshold Response	Cumulative/Chronic Adaptive Response Increased Tolerance	Cumulative/Chronic Adaptive Response Decreased Tolerance	Cumulative/Chronic Adaptive Response Maintenance	Chronic Injury Response
Muscle	- Strain	- Strain - Fatigue - Tenderness	↑ Contractile Protein ↑ Fiber Diameter ↑ Peak Tension ↑ Peak Power	↓ Contractile Protein ↓ Fiber Diameter ↓ Peak Tension ↓ Peak Power	No Functional Change	- Myositis - Myofascial Pain Syndrome - Myofiber degeneration
Tendon	- Strain	- Tenderness - Pain - Swelling - Stiffness	↑ Cross-sectional area ↑ Stiffness ↑ Strength	↓ Cross-sectional area ↓ Stiffness ↓ Strength	No Functional Change	- Tendonitis - Tenosynovitis
Nerve	- Axonal demyelination and degeneration - Temporary or persisting paralysis - Motor impairment - Sensory dysfunction - Decrease in conduction velocity	- Tingling - Numbness - Irritated nerve tissue - Neurapraxia - Paresthesias - Weakness	↑ Maximum discharge rate ↑ Activation during MVC neurogenesis (hippocampus) ↑ Motor unit synchronization ↑ Dendritic arborization ↑ Serotonergic neural activity ↑ Synaptic transmission ↓ Recruitment threshold	↓ Maximum discharge rate ↓ Activation during MVC neuron loss ↑ Recruitment threshold	No Functional Change	- Axonotmesis - Neurotmesis - Carpal Tunnel Syndrome - Sciatica
Bone	- Fracture	- Stress - Fractures - Pain - Tenderness - Crescendo of pain	↑ Bone Mineral Density ↑ Strength	↓ Bone Mineral Density ↓ Strength	No Functional Change	- Dislocation - Epicondylitis - Degenerative Disc - Herniated Disc - Facet Syndrome
Ligaments	- Sprain	- Pain - Swelling - Instability	↑ Cross-sectional area ↑ Stiffness ↑ Strength	↓ Cross-sectional area ↓ Stiffness ↓ Strength	No Functional Change	- Ligament Tear

In general, the failure or injury of tissue occurs when applied loads exceed the failure tolerance of the tissue. The structure of the tissue, influenced by its history of recent physical stresses and its accumulation, may be a factor to injury. The type and extent of tissue damage is dependent on the characteristics of the load (rate and mode) and tissue properties (McGill, 1997). Over time, repetitive movements, especially performed at a high frequency, will not allow sufficient time for repair. Repetitive strain can lead to cumulative fiber microtrauma and may result in pain, inflammation, and edema (Sandrey, 2000). Cutlip et al. (2009) demonstrated the effect of increased repetition on myofibre necrosis and myositis. Injury tolerance was compromised with increasing repetitions. If reparative tissue is overwhelmed, there may be a reduction in functional health, reduction in performance, persistent pain, and long-term tissue damage. During the healing process, connective-tissue replacement occurs instead of regeneration. The resulting tissue may be weaker and inflexible and may not be able to adapt to external and internal stresses (Sandrey, 2000). Even if sub-threshold stresses are subsequently applied, tissue regeneration is prevented and may cause pain and further tissue damage (Mueller & Maluf, 2002).

In this research, attention will be paid to muscle responses. According to the statistical data (Table 2.1), a large proportion of lost time claims/cases due to injury are muscle related.

What Are The Mechanisms of Muscle Injury and Repair?

Muscle injury is marked by reduced calcium (Ca^{2+}) sensitivity of myofilaments and reduced Ca^{2+} release from the sarcoplasmic reticulum (Westerblad & Allen, 1991). The influx of Ca^{2+} triggers a cascade of events and an activation of intrinsic proteases that autodigest the myofiber. This may lead to necrosis and myofibral gaps (Gates & Huard, 2005). Furthermore, an increased concentration of inorganic phosphate (P_i) in the myoplasm is observed with muscle injury. Increased P_i reduces crossbridge force and Ca^{2+} sensitivity of myofilaments. Inorganic phosphate may also combine with Ca^{2+} to form an insoluble precipitate of calcium phosphate (CaP_i). The formation of CaP_i may lead to a decrease in Ca^{2+} release from the sarcoplasmic reticulum and a decrease in muscle performance (Allen & Westerblad, 2001). With muscle injury, there is a decrease in pH that is strongly correlated with a decrease in force production. A low pH reduces the number of high force crossbridges in fast fibers and force per crossbridge in both fast and slow fibers (Fitts, 2008). An increase in potassium (K^+) concentration was observed due to a reduction in K^+ and sodium (Na^{2+}) concentration gradient. This may result in an impaired tetanic force and attenuation of the M-wave area (Sejersted & Sjogaard, 2000). Kent-Braun (1999) insists that muscle injury leads to a reduced excitability of neuromuscular transmission.

Muscle injury repair occurs in four phases: necrosis/degeneration, inflammation, regeneration, and fibrosis. Necrosis/degeneration occurs immediately after injury. As response to muscle injury, capillaries in the muscle belly are disrupted, leading to hematoma formation. Growth factors, native myogenic cells, cytokines, and inflammatory interact. Neighbouring healthy myofibers are affected, producing multiple distinct bundles of ruptured myofilaments (Gates & Huard, 2005). The inflammation phase follows and is marked by neovascularization. Neutrophils, activated macrophages, and myogenic cells secrete cytokines. Myogenic cells also secrete growth factors, which are important in regeneration and fibrosis. The regeneration phase is characterized by the activation of satellite cells that proliferate and differentiate into multinucleated myotubes, immature myofibers, and later mature myofibers. Growth factors enhance proliferation and degeneration. The final phase is fibrosis. In this stage, there is an overproliferation of extracellular matrix components. The regeneration process is impaired and the normal tissue architecture is distorted. Fibrotic tissue increases in size over time and is seen as the end product of muscle repair. Increased fibrotic tissue impairs functional outcome, causing a loss of strength, flexibility, and injury (Gates & Huard, 2005). Muscle injury mechanism and recovery phases are summarized in Table 2.3.

Table 2.3 Injury and recovery mechanisms and recovery time of muscle

Tissue	Injury Mechanism	Recovery Mechanism	Recovery Time
Muscle	- ↓Ca^{2+} sensitivity of myofilaments - ↓Ca^{2+} release from SR - ↑ Concentration of inorganic phosphate in myoplasm - ↓Excitability of neuromuscular transmission - ↓pH - ↑ K^+ concentration	- Necrosis/Degeneration - Inflammation - Regeneration - Fibrosis	- Necrosis/Degeneration: After injury to 1 week - Inflammation: within 24 hours after injury and continues until 72 hours after injury. Peaks at 24 hours - Regeneration: Begins 7 days after injury. Satellite cells proliferate and differentiate as long as 10 days. - Fibrosis: Begins 2 weeks after injury and accelerates for as long as 4 weeks after injury.

Past Prediction Models and Studies

A number of prediction models have been developed to relate variables with fatigue. Localized muscle fatigue and its associated factors (e.g. sustained muscle tension and working posture) have been shown to contribute to musculoskeletal disorders (Iridiastadi & Nussbaum, 2006). By using a wide range of intermittent static tasks, Iridiastadi and Nussbaum (2006) established quantitative models that predicted endurance time and muscle fatigue using three input parameters of intermittency (contraction level, duty cycle, cycle time). The outputs were based on ratings of perceived discomfort and EMG parameters including mean power frequency, median power frequency, and root mean square. Models were developed using linear regression and the slopes of both MVC and outputs with regard to time. Stepwise linear regression was used to establish models for endurance and each measure of localized fatigue based on the input variables. There were a few study limitations to ponder. Physiological responses were obtained from the middle deltoid and the contributions from synergistic muscles. There was an assumption of consistent load sharing between muscles. Very short and very long cycle times were not investigated and warrant additional study.

Endurance and fatigue limits for overhead tasks were investigated by Nussbaum and colleagues (2001). Fatigue onset was determined by the rate of change of maximum voluntary exertions while manipulating target height, duty cycle, and hand orientation. Nussbaum et al. (2001) concluded that duty cycle is a critical parameter on endurance and subjective discomfort. Muscle activity varied 10-30%, leading Nussbaum and colleagues to believe that existing guidelines for intermittent static work limits may not be applicable to complex and realistic tasks that involve intermittent dynamic levels of muscle activity within work cycles. Additionally, they question whether myoelectric measures were the best indicators for fatigue during dynamic intermittent tasks.

Sood et al (2007) investigated the reliability of several subjective and objective localized muscle fatigue indices during intermittent overhead work. Fatigue indicators included MVCs, EMG-based measures, and ratings of perceived discomfort. Studying low levels of exertions that were intermittent and dynamic, EMG measures exhibited lower reliability and more variability between muscles while perceived discomfort demonstrated excellent reliability. However, surface EMG (sEMG) was obtained from the anterior deltoid, medial deltoid, and upper trapezius, but excluded the supraspinatus due to its inaccessibility by sEMG. The supraspinatus is often the focus in occupational shoulder injuries. These results may also be limited to overhead tasks and require future considerations of other muscles in other actions to generalize the results.

A model was developed by Wood et al (1997) to predict fatigue of workers as they performed repetitive jobs knowing only the force and temporal patterns. The model is based on the U-shaped relationship between fatigue and level of desired force when physiological work is held constant (Wood et al., 1997). There was potential to accurately predict changes in fatigue across variations of work-rest but was met with several limitations. Limiting subject selection to male participants will provide challenges when applying the model to the current workforce. In 2006, women accounted for 47% of the employed workforce (Statistics Canada, Women in Canada: Work Chapter Updates). Secondly, three constant work-rest ratios were tested which may not be practical in real work tasks.

Fatigue-related changes during low-level force activities on the shoulder were reviewed by de Looze et al., (2009). The authors argue that in low-level activity, subtle intramuscular changes occur and maximal muscle strength and performance measures may not detectably decrease. However, they agree with past literature that musculoskeletal health risks were observed at contraction levels less than 15% MVC (Rohmert, 1973), between 2-5% MVC (Jonsson, 1988), and as low as 0.5 to 1% (Veiersted et al., 1990; Jensen et al., 1993b). Of the 13 papers reviewed, the EMG amplitude ranged from 3-6% MVC up to 16% MVC and had a task duration between 1 to 9.5 hours. De Looze et al. (2009) found that EMG-based fatigue measures were observed in low-force activities at intensities 15 – 20% MVC and there were consistent increases in subjective ratings with fatigue. However, further investigations to assess fatigue at prolonged, low-force activities using measures other than EMG are needed.

Muscle of Interest: Triceps Brachii

This study investigated the responses of muscle using the triceps brachii. The triceps brachii muscle has been shown to contribute to upper limb motions common to tasks found in industry (Louis & Gorce, 2009). Additionally, studies have investigated the development of fatigue of the lateral head of the triceps while performing continuous and intermittent contractions (Bilodeau, 2006). Triceps brachii response and activity have been measured by near-infrared spectroscopy (Lusina et al., 2008), surface electromyography (Marusiak et al., 2009; Louis & Gorce, 2009; Jaskolski et al., 2007), mechanomyography (Marusiak et al., 2009; Jaskolski et al., 2007), and muscle biopsy (Gjovaag & Dahl, 2008). One benefit in using the triceps brachii is the relatively known contributions during isometric elbow extension. Unlike elbow flexion where load sharing is distributed over multiple muscles (M. biceps breve, M. biceps longum, M. brachioradialis, M. brachialis, M. pronator teres), elbow extension is distributed over a smaller number of muscles: the medial, lateral, and long heads of the triceps brachii as well as the anconeus (Zhang & Nuber, 2000; van Bolhuis & Gielen, 1997). Most importantly, the primary contributor for elbow extension is the medial and lateral head of the triceps, contributing about 70 – 90% of total elbow extension moment with the elbow at 90 degrees and the forearm in neutral position (Zhang & Nuber, 2000). The dominant extensor between the medial and lateral heads is specific to the individual. On the other hand, the primary contributor in isometric elbow flexion at 90 degrees is the brachialis with a maximum contribution of 40.73% (van Bolhuis & Gielen, 1997). The second most dominant flexor at 90 degrees in isometric elbow flexion is the biceps longum at 26.59% (van Bolhuis & Gielen, 1997). Relative contributions are dependent on position and task (van Bolhuis & Gielen, 1997; Naito et al., 1998). This may suggest that the exposures of force in this study will be better reflected by collecting activity and responses from the three heads of the triceps rather than multiple muscles required for elbow flexion. According to An et al. (1981), the physiological cross-sectional area for the medial, lateral, and long heads of the triceps are 6.1, 6.0, 6.7 cm^2 respectively.

The composition of the muscle, particularly fibre type, may affect muscle twitch force and recovery process. Adamo and colleagues (2009) suggest that a muscle composed of a greater proportion of slow, fatigue-resistant type I muscle fibres may be less sensitive to K+ loss, leading to a quicker recovery process after sustained exertions. There is also a strong correlation between the percentage of type I muscle fibres and muscle belly response to a twitch stimulus contraction time (Dahmane et al., 2000). To measure the percentage of type I muscle fibres, Dahmane and colleagues (2000) extracted muscle samples from 15 male cadavers between 17 and 40 years old and found that the triceps brachii is composed of 35 +/- 8% of type I muscle fibres with a contraction time of 30 +/- 6 milliseconds. This may suggest that triceps brachii muscles are type II fibre dominant and may have a slower recovery process than type I fibre dominant muscles. However, Adamo and colleagues (2009) argue that the threshold of fatigue resistance due to increased proportion of slow twitch type I fibres may be challenged by greater demands of the work tasks, such as force level, cycle time, and duty cycle. Gjovaag & Dahl (2008) supports the use of triceps brachii to investigate exercise response due to its type II fibre dominancy. Gjovaag & Dahl (2008) argue that postural muscles, which are type I fibre dominant, receive much stimulation from normal ambulatory and postural activity. They suggest that daily simulation of these muscles may affect the responses following exposure to different force patterns and influence the comparisons between experimental conditions. This may be particularly important when experimental conditions/exercise patterns are measured on different days. A non-postural or type II fibre dominant muscle such as the triceps brachii, on the other hand, will exert less force and activity between conditions (Gjovaag & Dahl, 2008) and its response may better reflect changes in amplitude variation.

The size of the muscle and gender effects has been shown to influence the fatigability of muscle. Mannion and colleagues (1997) suggest that increased size not only generates the greatest total force in the sagittal plane but also negatively affects the capacity to sustain a contraction. A large muscle working at the same percentage of maximum contraction as a smaller muscle will occlude circulation to a greater extent and fatigue more rapidly (Wust et al., 2008). However, the ratio of type I fibres and type II fibres of a particular muscle of different sizes are fairly consistent (Mannion et al., 1997). Gender differences have been reported where men typically have larger fibres while females have greater type I fibres, as a consequence of higher type I:II fibre size ratio. Wust et al., (2008) describe previous studies that show a 5% greater percentage of type I fibres in premenopausal women than men. When matched for age and physical activity, males were less fatigue resistant than females during a series of intermittent contractions (Wust et al., 2008). Wust and colleagues (2008) suggest that the contractile speed and rate of energy utilization underlie this sex-related difference.

Although it has been shown that gender differences may affect the fatigability of the muscle, this study observed the muscle responses of 15 male participants. Given the difference in upper arm circumference and muscle size between genders and the number of assessment tools, it was not feasible to measure all responses from female participants.

How is fatigue assessed?

Surface Electromyography

Surface electromyography (EMG) records the algebraic sum of all motor unit action potentials along a muscle fibre at a particular point in time (Winter, 2005). The gross signals of electrical activity are detected from the motor units within the pickup area and provide an estimation of the neural drive (Basmajian, 1985; Vedsted et al., 2006). The EMG amplitude is affected by the number and firing rate of active motor units, the shape of the action potential, and the cross correlation of motor unit discharge (Vedsted et al., 2006). EMG has been an accepted method to indicate muscle fatigue due to changes in EMG amplitude and frequency (Søgaard et al., 2003). An increase in EMG amplitude signifies an increase in central drive in order to compensate for the loss of force producing capacity. The shift in the frequency spectra is attributable to a decrease in muscle fibre conduction velocity (Ebersole et al., 2006; Iridiastadi et al., 2008). Vollested (1997) demonstrated a parallel fall in MVC force with EMG amplitude. Although the cause and effect relationship is unclear, EMG may be a good marker for fatigue development (Vollested, 1997). EMG may also be used as an indicator of long-term post-exercise fatigue in both static and intermittent contractions at levels as low as 10% MVC (Blangsted et al., 2005).

There are potential limitations when using EMG, in both amplitude and frequency domains, to measure acute responses of muscles. Changes in EMG amplitude reflect either a change in number of active muscle fibres or excitation rates; currently it is not possible to distinguish between the two factors (Vøllestad, 1997). Changes were also inconsistent among subjects (Iridiastadi et al., 2008). Load sharing between muscles may affect the observed muscle activity. As a result, fatigue development for a particular muscle may be slowed; disturbing the relationship between endurance time and EMG based fatigue indicators (Iridiastadi et al., 2008).

According to Clancy et al. (2005), EMG recordings should be avoided at forces below 20-30% MVC. At these low forces, additive background noise was accentuated, compromising the spectral analysis in the frequency domain. Mean and median frequencies below 25% were artifactually inflated with mean frequency estimates more susceptible to noise (Clancy et al., 2005). More recently the effects of muscle temperature have been shown to influence the time required to reach fatigue. With increasing contractions there is a subsequent increase in muscle metabolism and endogenous heat production. Temperatures can rise 3 to 4 degrees Celsius and reach a value as high as 41 degrees (Reardon & Allen, 2009; Place et al., 2009). This may be due to the reduction of potassium and hydrogen levels at elevated temperatures (Reardon & Allen, 2009). However, Place et al. (2009) showed that there were no differences in fatigability between normal and high muscle temperature in fatigue-resistant muscles of mice in vivo. Muscle temperature may also affect the frequency characteristics of the EMG signal as temperature affects the motor unit potential conduction velocity and influences the lower range of the power spectrum (Fenwick et al., 2010). Fenwick and colleagues (2010) found a variable effect of temperature, between participants, on the mean power frequency (MPF) of the EMG signal between 34° and 39° Celsius. Overall, an increase in temperature led to an increase in MPF but its effects on the EMG signal could be eliminated by a three-fold decrease in MPF during fatiguing contractions. Changes in RMS amplitudes were minimally affected by muscle temperature (Fenwick et al., 2010).

Mechanomyography

Muscle mechanomyography (MMG) is used to detect intrinsic mechanical properties of active muscle fibres (Søgaard et al., 2003). There is potential use of MMG to measure changes due to fatigue (Søgaard et al., 2003). MMG measures the skin surface oscillations during contractions. Contractions are composed of pressure waves that result in pressure fluctuations that are propagated to the skin surface (Shinohara & Søgaard, 2006). Such pressure waves are due to radial thickening and lateral movement of the active fibres that results in dimensional changes of fibres of each active motor unit (Vedsted et al., 2006). The lateral movement transfers force to the series elastic elements and tendon (Shinohara & Søgaard, 2006). The resulting amplitude of MMG represents the magnitude of force fluctuations during voluntary contractions (Shinohara & Søgaard, 2006). MMG also measures the underlying cross-bridge cycling (Blangsted et al., 2005; Vedsted et al., 2006) and during low-force contractions, it may indicate impairments in force generation at the myofibrilliar level (Blangsted et al., 2005).

MMG responds differently than EMG and is dependent on the mode of contraction (Vedsted et al., 2006; Kawczynski et al., 2007). Similar to EMG, changes in both amplitude and frequency domains may be an indicator of fatigue. Changes in amplitude may reflect motor unit recruitment, increases in firing rate, and synchronization of active motor units. Changes in its frequency spectra reflect the global firing rate of unfused, activated motor units (Ebersole et al., 2006; Blangsted et al., 2005). These pain-related neuromuscular changes may be associated with nociceptive afferents feedback (Kawczynski et al., 2007). Vedsted and colleagues (2006) showed an increase in MMG amplitude with a progressive recruitment of motor units. Similarly, there was an increase in MMG amplitude with increasing torque production (Ebersole et al., 2006). As a result, there is potential to use MMG as an indicator of fatigue and long-term post-exercise fatigue after intermittent and static contractions at submaximal levels as low as 10% MVC (Blangsted et al., 2005). In fact, Blangsted et al., (2005) suggested that MMG is more sensitive than EMG when detecting fatigue in both 10% and 5% MVC contractions. The increase and decrease in time and frequency domains, respectively, were more pronounced in MMG than EMG. MMG has been validated for monitoring muscle activity with kinemyography (KMG) and phonomyography (PMG) (Trager et al., 2006).

Like EMG, there are limitations and assumptions when using MMG. It is assumed that during voluntary contractions, MMG is intrinsic to the muscle and not due to artifact such as skin displacement (Vedsted et al., 2006). The response of MMG may be also modulated by an increase in intramuscular pressure, making it difficult to differentiate between a response due to motor unit strategy and intramuscular pressure (Shinohara & Søgaard, 2006). However in subsequent research, it was shown that intramuscular pressure had no influence on MMG amplitude (Søgaard et al., 2006).

Near-Infrared Spectroscopy

Near-infrared spectroscopy (NIRS) is a non-invasive optical tool that monitors the balance between oxygen supply and utilization (Pereira et al., 2007; Felici et al., 2009) and blood volume levels of tissue to measure changes in glucose, lactic acid, hematocrit, and body temperature (Lafrance et al., 2004; Mancini et al., 1994). Emitting light photons in the 700 – 1000 nm spectrum (Pereira et al., 2007), light is delivered through pulsed laser diodes at multiple wavelengths (McGorry et al., 2009). As oxyhemoglobin is deoxygenated, absorbance at 760 nm increases while it decreases at 850 nm. During re-oxygenation, the opposite occurs. The NIRS signal is primarily derived from small blood vessels (arterioles, capillaries, venules) rather than large blood vessels that contain a large molar quantity of blood (Pereira et al., 2007). Additionally, NIR light is scattered and largely absorbed by denser internal tissues and minimally affected by bone refractions. Incident light is transmitted back to the photodetectors, with all wavelengths except 760 nm and 850 nm filtered out. The subsequent light intensity changes reflect changes due to hemoglobin.

NIRS has been found to correlate well with blood flow, venous oxygen saturation (Pereira et al., 2007), and EMG (McGorry et al., 2009). Such significant test-retest correlations were reported for the erector spinae muscle during static contractions, the vastus lateralis during knee extensions performed at slow and fast movement velocities (Pereira et al., 2007) and the forearm during repetitive gripping tasks (McGorry et al., 2009). In addition, NIRS has been shown to be sensitive to subtle changes in oxygenation kinetics (McGorry et al., 2009) and minimally affected by changes in skin blood flow and body temperature between 760 and 800 nm (Mancini et al., 1994).

Despite the positive reviews there are limitations to using NIRS as a tool to measure blood flow and oxygenation. NIRS provides qualitative data and does not yield absolute levels of hemoglobin deoxygenation. It is assumed that the penetration depth is 2.5cm (McGorry et al., 2009), which is required to convert the absorption of light to the absolute concentration of hemo- and myoglobin. However, the path can be affected by skin, subcutaneous fat, and other factors (Mancini et al., 1994; McGorry et al., 2009). There are suggestions of the influence in hand dominance and participant's past history in NIRS-derived measurements but this could be minimized by low work intensities (McGorry et al., 2009).

Vascular Response

Similar to NIRS is the measurement of blood flow of larger blood vessels (arteries) using laser-doppler or Multigon doppler ultrasound. The laser-doppler is used to investigate blood-flux responses within the infrared wavelength (Roe & Knardahl, 2002; Larsson et al., 1995) while the doppler ultrasound images arteries to be used to calculate mean blood velocity (Saunders et al., 2005; Rogers et al., 2006). In general, blood flow increases in response to both contractile activity and metabolic work (Hamann et al., 2005) and can be used to measure physiological biomarkers.

There is ample literature that suggests a relationship between increased blood flow and muscle contractions (Roe & Knardahl, 2002; Larsson et al., 1995; Hamann et al., 2005). Even at low-level muscle activity, there is an increase in blood flux with the largest increase immediately after the end of activity (Roe & Knardahl, 2002). Blood flow returns to normal levels within a 10-minute rest period (Larsson et al., 1995). Blood flow is dependent on working velocity, contraction frequency, power output, and oxygen consumption (Hamann et al., 2005). For instance, metabolic demand and a resulting increase in blood flow correlates well with high frequency and short-duration contractions (Hamann et al., 2005). These tools to measure blood flow were found comparable to surface EMG while statically holding a 1 kg load, particularly to the fall of mean power frequency (Roe & Knardahl, 2002; Larsson et al., 1995).

Despite evidence towards blood flow and muscle contractions, there are studies that have shown a mismatch between metabolic requirements and blood flow at low levels of muscle activity (Roe & Knardahl, 2002). High blood flow velocities may also cause measurement errors as well as movement artifacts (Roe & Knardahl, 2003). Contributions from muscle, skin, and non-nutritive blood flow may also be a potential source of error. Changes in tissue blood volume and erythrocyte concentrations may also influence measurements. Lastly there may be large inter-individual variations and a lack of absolute values of blood flow (Roe & Knardahl, 2003). However, although unavoidable, the magnitude of these errors is very small (Walker et al., 2007).

With the laser-doppler method, static muscle contractions were associated with insufficient blood flow, particularly in the presence of pain (Roe & Knardahl, 2002). However, laser-doppler is an invasive measurement using a single optic fibre placed percutaneously within the muscle (Larsson et al., 1995).

Using transcranial doppler ultrasound, vasoregulatory mechanisms showed symmetry in their response to increases and decreases in contraction intensity and were identical under fast and slow contraction intensity (Rogers et al., 2006). Transcranial doppler can also reveal rapid vasoregulatory mechanisms under repeated step oscillations of force or when the exposure consists of 2 or more identical oscillations per contraction (Rogers et al., 2006). However, the vasoregulatory mechanism may not be able to respond rapidly when the contraction intensity is alternated every contraction. Transcranial doppler requires a probe that is fixed to the skin, over the artery. As such, experimental conditions must be at a cool temperature to reduce the oscillations in blood flow due to the rhythmic opening and closing of temperature-sensitive artheriovenous anastomoses. This artifact may result in a four-fold increase in blood flow (Walker et al., 2007). Doppler ultrasound has been used to measure blood velocity through the profunda artery within the triceps brachii muscle during sustained isometric contractions of 20% MVC (Griffin et al., 2001). As the arterial diameter did not change with contractions, blood velocity through the profunda artery was an indicator of blood flow (Griffin et al., 2001).

Ratings of Perceived Exertion

Another short-term indicator for complex work patterns is perceived exertion (de Oliveira Sato & Cote Gil Coury, 2009). Ratings of perceived exertion (RPE) are a subjective indicator of work intensity and are likely mediated by central mechanisms where the corollary provides a copy of motor commands to the somatosensory cortex. The central mechanism ensures muscle energy homeostasis is maintained and critical energy depletion does not occur (Allman & Rice, 2003; Noakes, 2004). Higher RPE may be a consequence of a greater proportion of maximal neural drive (Allman & Rice, 2003).

Past literature has identified a linear relationship between RPE and duration of exercise. RPE is positively related to fatigue and reduced work capacity (de Oliveira Sato & Cote Gil Coury, 2009) and is found to increase during 10% MVC (Blangsted, 2005; Helene Garde et al., 2003). Noakes (2004) has surmised that the time left to exhaustion at a constant workload can be reflected by the rate at which the RPE increases.

Since RPE may be mediated by central mechanisms, it is possible that age-related hyperactivity may elevate perceived exertion in older aged individuals (Allman & Rice, 2003). Familiarity to the required tasks may also affect psychological dues related to RPE. This is true if the greater proportion of input cues to perceived exertion are psychological opposed to physiological (Mital et al., 1994). RPE may also be more strongly coupled to high frequency fatigue than low frequency fatigue (Adamo et al., 2002; Adamo et al., 2009). Adamo and colleagues (2002) found dissociation between subjective and objective measures of fatigue as subjects perceived muscle fatigue during and immediately after the work task, whereas LFF measures showed the greatest reduction in twitch force 30-60 minutes post work. The work task was composed of intermittent and sustained grip exertions of 5% MVC.

Muscle Stimulation

Low-frequency fatigue (LFF) is low force exertions, sustained or intermittent, that is generated over time. LFF may be a precursor for musculoskeletal disorders (Hagberg et al., 1995) and its recovery process may exceed 24 hours (Adamo et al., 2002). This fatigue may be associated with the failure in the excitation-contraction coupling mechanism (Adamo et al., 2002) and depression in Ca^{2+} release (Green et al., 2004). According to Green and colleagues (2004), at low frequencies, a small reduction in cytosolic free Ca^{2+} ($[Ca^{2+}{}_f]$) can lead to reductions in force given the steep nature of the force-frequency curve. A method to measure LFF is by electrical stimulation of the muscle and its twitch response. The magnitude LFF during exercise and recovery can be calculated as the force response ratio (response at 20 Hz divided by response at 100 Hz). The relative decline in muscle's response to low (20 Hz) vs high (100 Hz) frequency electrical stimulation is indicative of low-frequency fatigue (Adamo et al., 2009).

Eliciting 2 Hz muscle twitch forces, Adamo and colleagues (2009) found that a decrease in twitch force persisted after 60 minutes post-work after sustained contraction (8% mean force MVC) and recovery of twitch force at 17% mean MVC intermittent grip force pattern after 15 minutes post-exercise. LFF measures detected low-frequency fatigue during intermittent and sustained low-force grip exertions (Adamo et al., 2002). Additionally, LFF measures may detect fatigue that cannot be perceived by the subject (Adamo et al., 2002). Fatigue responses detected by twitch force responses were also not significantly affected by age or gender during intermittent grip exertions of 10% MVC and sustained handgrip exertions of 8% MVC (Adamo et al., 2009).

Hamada and colleagues (2004) caution the use of electrical stimulation as an alternative to voluntary contractions to evoke electrophysiological responses. At identical low intensity intermittent force of 10% MVC, electrical stimulation led to a different activation pattern in motor unit recruitment, where type II glycolytic fibres are recruited first (Hamada et al., 2004). Unlike voluntary contractions, electrical stimulation activates axon collaterals with the largest diameter more rapidly than smaller diameter axons. Larger axonal diameter provides less resistance against external electrical current (Hamada et al., 2004). This suggests the possibility of an "inverse size principle" of motor unit recruitment when the muscle is activated by electrical stimulus (Hamada et al., 2004).

As a result, electrical stimulation may not be ideal as an alternative to voluntary contractions when measuring intermittent contraction responses. However, electrical stimulation is often used to measure twitch force responses and low-frequency fatigue.

Maximum Voluntary Force and Test Contractions

According to Vøllestad (1997) and Bystrom & Fransson-Hall (1994), maximum voluntary isometric contraction force is often used to measure force-generating capacity where a decline in maximal force may indicate the occurrence of central fatigue. Vøllestad (1997) points out that maximum voluntary isometric contraction force are useful as an assessment of fatigue if the joint/body segment is kept in a standard testing position to restrict length changes of the muscle, if there is only one direction of force generation/movement, and with appropriate practice and vocal encouragement.

Cycle Times, Forces, and Duty Cycles

Since this study focussed on the response of muscle and the triceps brachii were used to represent this activity, it will be important to use occupationally relevant forces and frequencies for the elbow and shoulder. These forces and frequencies are likely different between muscles and body regions (Kilbom, 1994; You and Kwon, 2005). For the elbow, Kilbom (1994) suggested that during dynamic and intermittent static contractions, there was a high risk of developing musculoskeletal disorders of the upper arm and elbow at frequencies greater than 10/minute (6 second cycle time). The 6-second cycle time was also observed in a screw-driving task using pistol power tools (Ulin et al., 1993), in a study of repetitive wrist flexion and extension (Snook et al., 1995), as an experimental condition for an intermittent isometric torque exertion task at varying forearm joint angles (O'Sullivan et al., 2005), and in a 30-minute fatiguing protocol of intermittent contractions at the elbow (Søgaard et al., 2003). In a study conducted by Silverstein and colleagues (2008), workers were 1.76 times more likely to develop rotator cuff syndrome at an exposure level of 3 to 6 second cycle time. Other cycle times used and identified in past literature include 3 seconds for an elbow flexion/extension fatiguing task (Potvin, 1997), 5 seconds for median arm movements while completing VDT work at a median force level of 4.5% MVC, and 34, 100, and 166 seconds in combination with various mean contraction levels and duty cycles for an intermittent isometric arm abduction protocol (Iridiastadi & Nussbaum, 2006). Sood et al., (2007) used a frequency of 1.1/minute (54.5 second CT) for a slow overhead drill tapping task and 40/minute (1.5 second CT) to provide a moderate psychomotor challenge. According to past literature, a 6 second cycle time may be representative of durations found in work.

Force levels can be estimated from psychophysical and workplace studies. Mathiassen (1993) used a mean load of approximately 14 – 18% MVC for shoulder flexion; however it was noted that the range may have been too high for direct comparison to occupational shoulder loads. Sood et al. (2007) suggested overhead working tasks to be within the range of 15 to 20% MVC but is affected by working height. Søgaard and colleagues (2003) reported fatigue with upwards to 30 minutes of recovery at mean forces of 5 - 10% MVC. The mean force used in an intermittent static contraction protocol with repeated short cycle elbow flexions was 14% MVC (Bjorksten & Jonsson, 1977). Flodgren and colleagues (2006) measured mean electrical activity at 9.3% MVC for 30 and 60 minute button pressing and piston pushing protocols at a work cycle of 30/minute. Westgaard and Winkel (1996) suggest exposure levels below 10% MVC as the most relevant force range for workers performing sedentary or light production work. Below 10%, workers developed musculoskeletal disorders (Westgaard & Winkel, 1996). In order to observe biomechanical, neurophysiological, and physiological changes within 60 minutes of exercise/work, I proposed a force level that is both relevant to occupation and with the development of WMSD.

Iridiastadi and Nussbaum (2006) used a range of contraction levels (10 – 30% MVE), duty cycles (0.2 – 0.8), and cycle times (20 – 180s) for intermittent abductions of the arm. However, significant fatigue was observed with combinations of mean percent MVE, duty cycle, and cycle time in 21%-0.75DC-166s CT, 21%-0.75DC-34s CT, and 16%-0.80DC-100s CT conditions. Sood et al., (2007) exposed participants to a 50% duty cycle and a cycle time of 54 seconds while Nussbaum and colleagues (2001) found that duty cycles of 66% resulted in earlier signs of fatigue compared to participants exposed to 33%. A duty cycle of 50% was used in a comparison between static and intermittent contractions (Vedsted et al., 2006) and 60%

for a 30-minute fatiguing protocol while flexing and extending the elbow (Søgaard et al., 2003). Silverstein and colleagues (2008) observed low duty cycles (0.03 – 0.15) and a high risk of rotator cuff syndrome in a cross-sectional study of 12 manufacturing and health care worksites. However, it is argued that longer cycle times are found in several industrial tasks such as construction, manufacturing, and assembly lines (Iridiastadi & Nussbaum, 2006). Based on past literature, a 50% duty cycle was applied to this study. These work parameters are summarized in Table 2.4 (cycle time), Table 2.5 (force amplitude), and Table 2.6 (duty cycle).

Table 2.4 Cycle time guidelines and observations from past literature

Cycle Time (Seconds)		
Guideline	**Rationale**	**Reference**
0.75 seconds	Overhead tapping motions between two targets separated 50 cm apart with a 0.36 kg wand-like tool.	Nussbaum et al., (2001).
2 seconds	Maintained for either 30 or 60 minutes. Simulated work task of pushing in a piston and pressing down a button while seated at a table.	Flodgren, et al., (2006).
3 seconds	Average frequency (19.1 +/- 4.1 cycles/min) for an elbow flexion/extension task until exhaustion (average 155.3 +/- 64.0 seconds).	Potvin, J.R. (1997).
3-6 seconds	Based on adjusted odds ratios of the frequency of shoulder movements. Workers with rotator cuff syndrome were 1.76 times more likely to have shoulder cycle times of 3-6 seconds.	Silverstein, et al., (2008).
6 seconds	For dynamic and intermittent static contractions, frequencies greater than 10/minute (6 second cycle time) are the limit for high risk of musculoskeletal disorders of the upper arm and elbow.	Kilbom, A. (1994).
	Used in a study of repetitive wrist flexion and extension.	Snook et al., (1995).
	Normal working pace for a screw-driving task using a pistol power tool for various locations	Ulin et al., (1993).
	Experimental condition for an intermittent isometric torque exertion task for both directions at 11 forearm joint angles based on past literature.	O'Sullivan & Gallway (2005).
	Used in a fatiguing protocol for intermittent contractions at the elbow. Fatiguing protocol was 30 minutes.	Sogaard et al., (2003).
4.6 seconds & 12 seconds	Median arm movements observed while completing tasks at a metal can production plant. Median EMG 5.3%MVC and 4.2% respectively for the two duty cycles.	Jensen et al., (1999)
5 seconds	Median arm movements observed while VDT work (CAD-operators) at median level of 4.5% MVC.	Jensen et al., (1999)
< 30 seconds	A cycle time below 30 seconds is highly repetitive and associated with increased risk of developing a hand/wrist disorder.	Silverstein et al., (1986)
34s, 100s, 166s	Cycle times in combination with various mean contraction levels and duty cycles for an intermittent isometric-isotonic arm abduction protocol.	Iridiastadi, H., & Nussbaum, M.A. (2006).

Table 2.5 Force amplitude guidelines and observations from past literature

Force Amplitude (% MVF)		
Guideline	**Rationale**	**Reference**
5% - 10%	Intermittent contractions at 10% MVC resulted in long-term fatigue.	Sogaard, et al., (2003)
5% - 15%	Upper limit for a force that cannot be sustained for more than 1 hour **without rest. As cited in Nussbaum et al (2001).**	Rose et al., (1992). Sato et al., (1984). Sjogaard et al., (1986).
9.3%	EMG activity of trapezius muscle during button pressing and piston pushing task. Fatigue observed in both 30 and 60 minutes (30/min)	Flodgren et al., (2006).
10%	Most occupationally relevant force range for sedentary work or light production at which many workers develop MSDs.	Westgaard & Winkel (1996).
14%	Mean for intermittent static contractions in a repeated short cycle elbow flexion task.	Bjorksten & Jonsson (1977).
15% - 25%	Maximum shoulder strength for overhead work task used for an intermittent tapping task.	Nussbaum et al., (2001)
16% - 21%	Mean contraction dependent on duty cycle and cycle time during intermittent isometric-isotonic arm abductions.	Iridiastadi & Nussbaum (2006).
16% - 43% 12% - 26%	Flexor digitorum superficialis and flexor carpi ulnari (respectively) in wire tying tools.	Li (2001).
17%	Mean for intermittent handgrip contractions were found unacceptable if values above this value.	Bystrom & Fransson-Hall (1994)
20%	**Chosen above lower levels since elbow injuries (epicondylitis) more commonly reported at higher forces than light assembly tasks.**	O'Sullivan & Gallwey (2005) Mukhopadhyay et al., (2009)
22.8% and **12.8%**	Mean respective concentric and eccentric values at onset of elbow flexion/extension cycles until exhaustion	Potvin (1997)
	Comparison between static and intermittent static contractions, not to fatigue. Differences in muscle tissue oxygenation at 20%	Vedsted et al., (2006)

Table 2.6 Duty cycle guidelines and observations from past literature

Duty Cycle (Proportion of 1.0)		
Guideline	**Rationale**	**Reference**
0.03 – 0.15	Based on adjusted odds ratios, duty cycle of forceful exertions (pinch forces >8.9N; power grip/lifting/push and pull >44.1N) resulted in rotator cuff syndrome cases with 3.27 greater odds to be between 3 and 15%	Silverstein et al., (2008)
0.33 and 0.66	The 0.33 duty cycle tasks were completed by majority of participants. Early signs of fatigue 30-60 minutes into the trial. Few participants completed 0.66 duty cycle tasks, 9-35 minutes into the trial. Overhead intermittent tapping task.	Nussbaum et al., (2001)
0.5	Comparison between static and intermittent contractions. Task was not done to fatigue.	Vedsted et al., (2006)
0.6	Used in fatiguing protocol for 30 minutes during elbow flexion-extension	Sogaard et al., (2003)
0.75 and 0.80	Fatigue induced at these duty cycles in combination with contraction levels and cycle times.	Iridiastadi & Nussbaum (2006)

Feedback Modality

This study utilized position control (proprioceptive feedback) rather than force control (visual feedback) while performing intermittent isometric contractions. Past literature showed that proprioceptive feedback provoked higher ratings of perceived exertion, lower force fluctuation, and steeper slopes of EMG and MMG parameters (Madeleine et al., 2002). Position tasks may also lead to shorter endurance times (Hunter et al., 2002; Rudroff et al., 2007) and greater rate of mean arterial pressure (Rudroff et al., 2007). However, Søgaard and colleagues (2003) argued that visual feedback requires a more complex control loop, increasing the amount of variation in the motor unit recruitment pattern affected by the excitatory descending drive and inhibitory input during the intermittent contractions. The long-term response, however, was similar for both visual and proprioceptive feedback modes and the may be not sufficiently large to prevent the impaired mechanical performance during recovery. As such, it is speculated that proprioceptive feedback may better reflect responses of fatigued motor units when compared to visual feedback (Søgaard et al., 2003) and is better representative of everyday tasks.

Chapter III

General Methodology

The following is a detailed summary of the methodology common to the two study sections (Chapter IV – V).

Participants

Fifteen male participants were recruited from a university student population (see Table 3.1). All participants had no current or past injuries of their right elbow, upper arm, and forearm. Participants were right hand dominant. Informed consent to the procedure, approved by the Office of Research Ethics, University of Waterloo, was obtained from participants prior to the study.

Table 3.1 Age and upper arm circumference of 15 male participants.

Participant	Age	Upper Arm Circumference (cm)
1	20	35.6
2	29	25.4
3	23	29.2
4	20	33.0
5	21	30.5
6	22	35.6
7	24	29.2
8	26	31.8
9	21	41.9
10	24	35.6
11	26	40.6
12	21	33.0
13	26	33.0
14	22	38.1
15	35	37.2
Mean	**24.0**	**33.98**
SD	**4.0**	**4.48**

Instrumentation

Surface Electromyography

EMG was recorded by bipolar surface electrodes (Ag-AgCl electrodes, Ambu Blue Sensor N, Denmark) and was placed on the belly of the lateral and medial head of the triceps as recommended by Cram et al., (1998). EMG also recorded antagonistic activity from the biceps brachii to confirm that there were minimal contributions from this muscle during elbow extension. This would ensure that the forces exerted during elbow extension were made primarily by the triceps brachii muscles. The inter-electrode distance was 20 mm. Prior to mounting the electrodes, the skin was abraded with ethanol and hair was removed by razor. NuPrep Gel (Weaver and Company, CO, USA) was applied to further lower impedence and improve conductivity. The EMG signal was collected using an 8-channel data system (Bortec Octopus, Calgary AL; common mode rejection ratio > 115 dB; band-pass filtered 10 – 1000 Hz). Muscle activation in all maximal trials was calculated from the middle three seconds of the five-second activation (Mathiassen et al., 1995). The root mean square (RMS) amplitude was calculated and normalized to the maximum voluntary force exerted during baseline activity of that trial. Fast Fourier Transform was applied to obtain the frequency spectrum from which mean power frequency (MnPF) and median power frequency (MdPF) of the raw signal were determined. Both mean and median power frequencies were chosen as there were potential differences noted in previous literature (Iridiastadi & Nussbaum, 2006; Ebersole et al., 2006; Clancy et al., 2005).

Mechanomyography

Mechanical aspects of the muscle were monitored using a uniaxial accelerometer (Bruel & Kjaer, 4507) with a measurement range of +/-70 G. The accelerometer had the following technical specifications: sensitivity - 100 mv/g, weight - 4.8 g, frequency range - 0.3 – 6000 Hz. The accelerometer was placed on the belly of the medial head of triceps distal to the EMG electrodes. The medial head of the triceps was chosen due to the accessibility and length relative to the lateral and long heads (An et al., 1981). MMG signals were digitally band-pass filtered at 5 – 100 Hz using a 2nd order Butterworth filter.

Vascular Response

Blood velocity through the profunda brachii artery supplying the triceps brachii muscle was monitored by a single-gated pulsed-Doppler ultrasound (Multigon 500, Multigon Industries, Mt. Vernon, NY). The flat 4.0

MHz probe was positioned over the brachii artery, at the superior portion of the upper arm, with the embedded crystal at a 45° angle to the artery in single testing mode. This setup allowed a clear Doppler signal at pre-exercise, during exercise, and in recovery. Blood velocity was captured at a sampling rate of 2048 Hz onto a PC computer data acquisition system (NIAD; version 1.0.0.10, University of Waterloo, 2001). Brachial artery diameter measurements were performed proximal to the site of the mean blood velocity measurements using ultrasonography.

Heart Rate

A three-lead EMG electrode arrangement, measuring heart rate, was attached to the participant's chest: below the right clavicle, below and slightly to the left of the sternum, and above and to the left of the umbilicus.

Ratings of Perceived Exertion

Participant's perception of effort was obtained using a 10 cm visual analogue digital scale with a linear potentiometer. The scale ranged between 1 and 100 where the former represents no effort and the latter as maximum effort.

Muscle Stimulation

Electrical impulses were delivered to the triceps brachii muscle by two fabricated electrodes using a Grass model S48 stimulator with isolation unit. Electrodes were placed on the proximal and distal portions of the triceps brachii muscle belly to recruit the largest number of muscle fibers. A constant current stimulator with a pulse width of 1 ms and a 2 second train of supramaximal stimulation was used for the 1 Hz twitch. Twitch force was measured as the difference between individual peak and baseline force levels averaged over each train (Adamo et al., 2009). Electrical stimulation testing to measure low-frequency fatigue was administered by delivering single supramaximal impulses at 20 Hz and 100 Hz using pulse durations of 50 s and train durations of 1 s. The stimulation was then adjusted for the 100 Hz train to produce, on average, approximately 15% MVC. At the same voltage eliciting 15% MVC, the triceps were stimulated at 20 Hz. These stimulations triggered a triceps brachii muscle contraction that extended the elbow. A force transducer attached to the apparatus then measured the subsequent contraction force. A typical tetanic and twitch response is shown in Figure 3A.

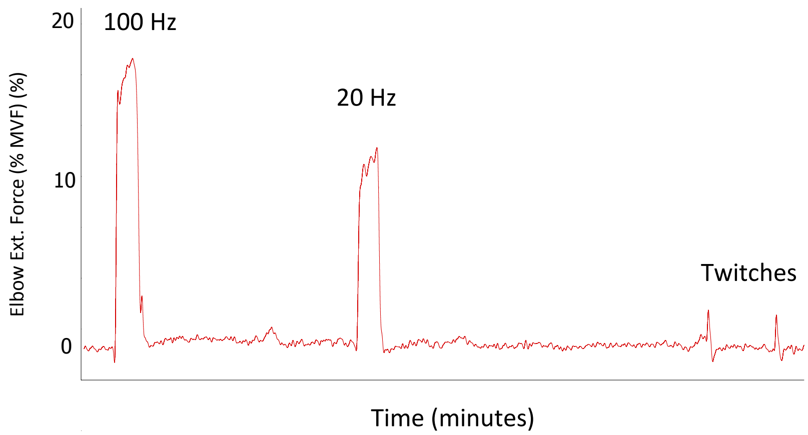


Figure 3.1 Muscle Electrical Stimulation Response
Typical extension force output response for tetanic stimulation at 100 Hz and 20 Hz and 1 Hz twitch. Forces are expressed as % MVF of the pre-experiment MVF.

Maximum Voluntary Force and Test Contractions

Maximum voluntary forces were collected after electrical stimulation and test contractions for a 5-second duration. The middle 3 seconds was analyzed. Any changes in MVF that occur during and after the exercise segment were related to the baseline MVF. The participant performed a test contraction at 15% of the force exerted in the pre-experiment MVF trial before each exercise protocol. Test contractions were collected for 12 seconds. Force measurements were collected from a force transducer attached between the apparatus arm and motorized shaft. Measurements was low-passed filtered at 10 Hz (Butterworth, 2nd order) based on a cutoff frequency determined from residual analysis.

Programmable Force Generator System

A fabricated apparatus (Figure 3B) was designed to support the arm and provide an external resistance that can be modulated to follow a chosen work profile. The system was composed of a brushless 220 VAC servomotor (Kollmorgen AKM64R, Danaher Motion, Washington USA) attached to a custom designed armrest (Department of Kinesiology, University of Waterloo). The servomotor was connected to a servo amplifier and encoder (Danaher S21260-VTS, Danaher Motion, Washington USA) and controlled by a programmable motion controller (DMC-1417, Galil Motion Control Inc., California). The servo motor system had a capacity of 19.257 Nm continuous stall torque and a safe operating range of +/- 9.7 volts. The servo motor system was manipulated using both position and torque controls using WSDK32 Galil software to change torque limits according to the participant's pre-session MVF. A force transducer was connected between the armrest and the shaft of the motor. A light switch was attached between metal stoppers along the armrest to provide visual feedback to the participant if they deviated from the range of operation. Padding was added for comfort and rubber stoppers were included for safety. The participant exerted resistive forces against loads manipulated by the researcher using this force generator system.

Counter-weights were added to the arm apparatus to ensure the participant's arm is positioned at the targeted level. At this level, forces were calibrated to zero.

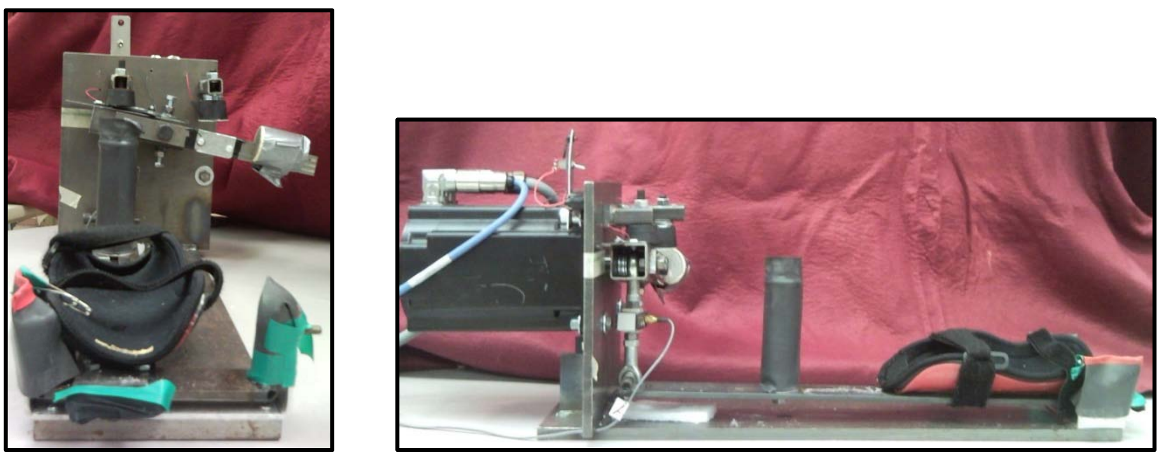


Figure 3.2 Front and Side Profiles of Programmable Force Generator System

Data Collection

Data was collected at 2048 Hz using NIAD Collection software (version 1.0.0.10, University of Waterloo, 2001). For every condition, 10 minutes of baseline activity, 60 minutes of exercise or until exhaustion, and 60 minutes of recovery were continuously collected. Subsequent processing was completed using Chart 4.0 (ADInstruments, Colorado Springs, CO, US) to window test contractions for a 10 second duration (middle 10 seconds for 12 second collection), MVF for a 3 second duration (middle 3 seconds of 5 second collection), 8 minutes into baseline for 30 seconds, exercise every 2 minutes for a 30 second duration, and recovery at 1, 2, 5, and 10 minutes for a 30 second duration.

Data Collection Protocol

Each condition consisted of two experimental sessions that occurred on two consecutive days. The first experimental session consisted of an extensive testing of an intermittent isometric contraction pattern and test batteries (Figure 3.3). Test batteries consist of muscle stimulation (LFF and twitches), test contraction at 15% MVF, and a MVF. The second session was a 24-hour follow-up that was composed by 30-second baseline activity and one test battery. This was repeated for the five experimental conditions (exercise contraction patterns), requiring the exercise and 24-follow up sessions for each condition.

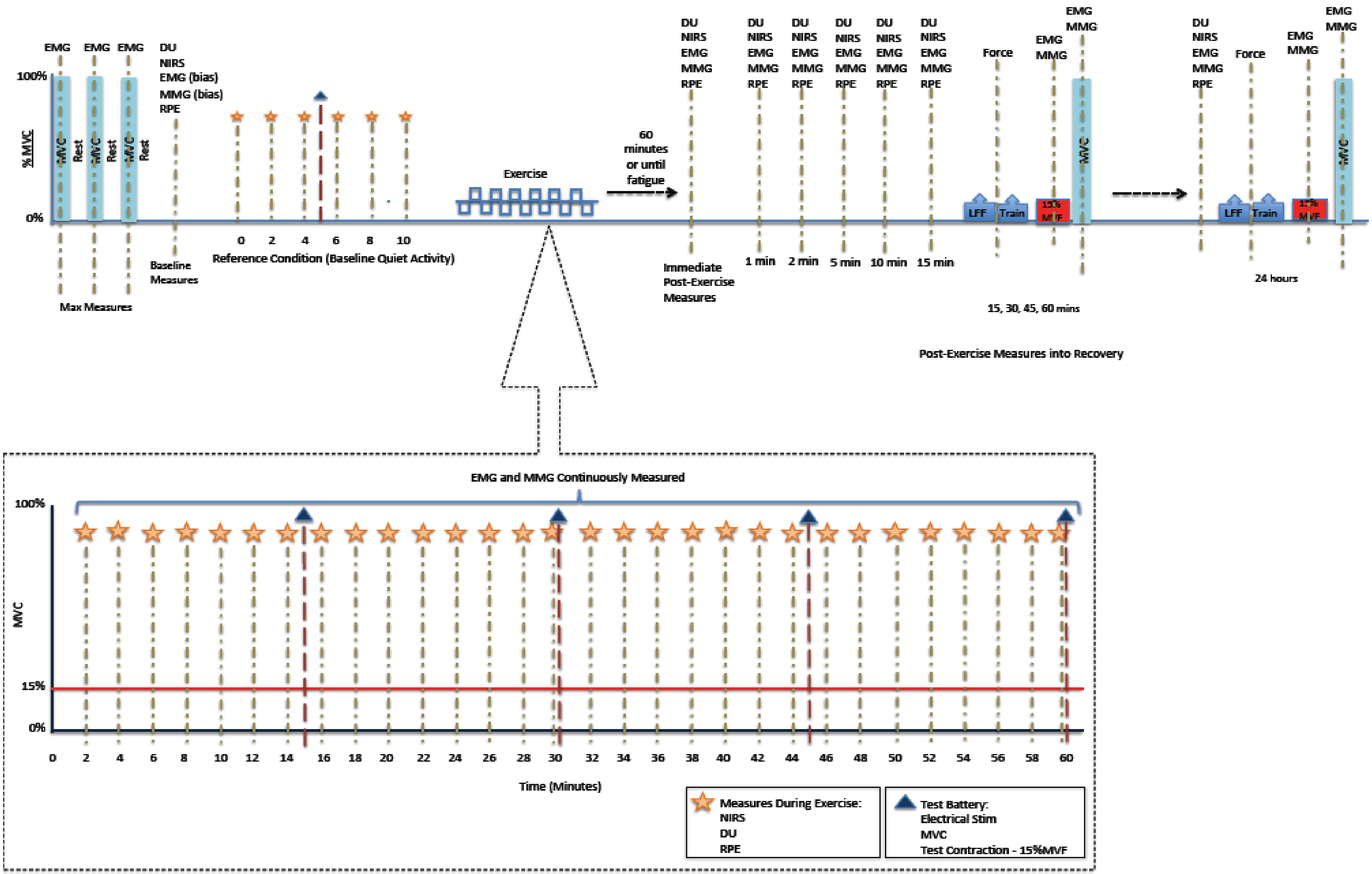


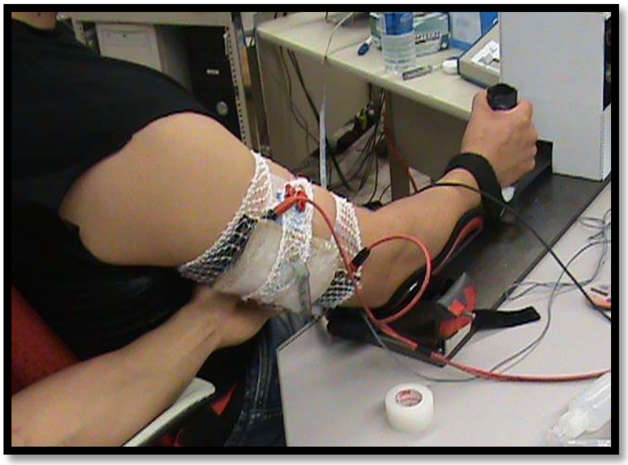
Figure 3.3 Data Collection Protocol
DU – Doppler Ultrasound, NIRS – Near Infrared Spectroscopy, EMG – Electromyography, MMG – Mechanomyography, RPE – Ratings of Perceived Exertion.

No caffeine or alcohol was allowed for at least 24 hours before sessions. Participants were also asked to refrain from exercise for at least 24 hours before the study. EMG, MMG, NIRS, doppler ultrasound, and electrical stimulation electrode placements was marked with an indelible felt-tip pen to ensure consistent placements between test sessions. Photographs were taken and participants were fitted with a saran wrap sheath, marked with the electrode placements, to further ensure consistent placement. The participants were comfortably seated with their right arm at 90 degrees elbow flexion and at a neutral forearm position. The right arm was supported by an armrest and held by the apparatus (Figure 3D). Three maximum voluntary contraction trials were collected with 2 minutes of rest between each maximum contraction. Maximum trials were collected for 5 seconds. The largest maximum voluntary contraction was used to determine the participant's force levels for all conditions, in order to maintain absolute force values.

After the three maximum voluntary contraction trials, voltages to elicit appropriate low-frequency fatigue twitches and 1Hz twitches were determined. Participants were then instructed to relax in the test position at which baseline activity was collected for 10 minutes. Baseline values were collected for EMG, MMG, NIRS, RPE, and Doppler ultrasound. At the middle juncture of the 10-minute collection, a test battery consisting of low-frequency fatigue twitches at 100Hz and 20Hz, two twitches at 1Hz for 1 second, a 15% MVF test contraction, and maximum voluntary force, was collected. Test batteries were performed during exercise every 15 minutes, at the cessation of exercise, and 15, 30, 45, and 60 minutes and 24 hours into recovery.

Participants were randomly selected to perform one of the five experimental conditions (variation about the mean +/- ½, variation about the mean with a minimum level of 1% MVF, variation about the mean with rest, i.e., +/- 1, no variance about the mean, i.e., static load, and sinusoidal waveform +/- 1). The variation about the mean condition has amplitudes within +/- 7.5% MVF and +/- 14% MVF from the mean. Variation with rest/no activity consisted of amplitudes at -% MVCF and 30% MVF (+/- 1 from the mean) that account for at least 50% of the number of contractions in the exercise/working cycle. Details of these test conditions will be discussed in Chapters II and III. Each experimental condition consisted of pre-, during, and post-test batteries and lasted up to a maximum of 60 minutes or until they could not continue. Previous literature has shown increased lactate concentrations after 20 – 30 minutes of repetitive low-load work (Ashina et al., 2002; Rosendal et al., 2004) and symptoms of fatigue within 60 minutes of repetitive low-load work (Flodgren et al., 2006).

Four measures of the effect of the experimental conditions was used: the value of the dependant measure during, at the cessation, and recovery after intermittent exercise, the value during, at termination, and recovery after sustained isometric exercise (Mathiassen, 1993), the value during, immediately after, and recovery of the sinusoidal waveform pattern, and the rate of change of the variable (Iridiastadi and Nussbaum, 2006). During these measures, all previously mentioned measurement tools monitored muscle responses. Ratings of perceived exertion and brachial artery diameter using ultrasound were collected every 2 minutes while EMG, MMG, Doppler ultrasound, and NIRS were collected continuously during the entire exercise segment (Figure 3.4). Combinations of doppler ultrasound, NIRS, EMG, MMG, and RPE were collected immediately post-exercise, and in recovery (1, 2, 5, 10, 15 minutes and 24 hours). Participants continued to follow the contraction work pattern until the participant could not complete the trial or until 60 minutes had elapsed. Participants received non-threatening verbal encouragement throughout the exercise duration. The room temperature was kept consistent (22 to 24°C) to minimize effects seen in NIRS, as previously mentioned.



A

B

Figure 3.4 Data Collection Set-Up
Typical collection set-up with participant’s arm in the programmable force generator system (A and B). Electrodes and probes are attached to the right upper arm and secured with hypafix and surgifix tape. The Multigon probe is held in place by the researcher on the brachial artery during collection (A). Participant exerts forces while watching standardized television programming (B).

Data Analysis

Data was excluded from analysis if participants did not meet the following criteria:

1. Sustain an isometric force for less than 60 minutes

2. Exert extension forces by isolating their triceps

3. Avoid intense physical activity involving their triceps 24 hours prior to their test session

4. Avoid alcohol and caffeine 24 hours prior to the test session

According to Mathiassen and Winkle (1992), forces between 13 and 18% MVF could be sustained for a median duration of 13 minutes and 2 seconds. As such, participants who exerted 15% MVF for the full 60 minute exercise protocol were excluded as it is possible that the elicited "MVF" forces were not truly maximums.

To determine if the conditions lead to fatigue, a test battery consisting of electrical stimulation, test contractions, and maximum voluntary of force contractions (MVF) was used to determine the effects of the contraction pattern. Electrical stimulation measures contractile function of muscle by comparing its relationship to force response. Previous literature has documented this relationship for limb and respiratory muscles by establishing a frequency-force curve (Moxham et al., 1982). A shift to the right (decrease in LFF ratio) implicates a reduction of force response to electrical stimulation at high frequencies and/or a reduction of forces at low-frequency stimulation (Moxham et al., 1982). The changes in force and low-frequency fatigue ratio were analyzed by finding the peak values during tetanus and twitch. Test contractions were elicited to measure EMG RMS, MnPF, and MdPF and MMG RMS, MnPF, and MdPF values at a consistent level of force (15% MVF). Test contractions were 12 seconds in duration and the middle 10 seconds were analyzed. Participants also exerted MVFs for a 5-second duration. Peak force was identified during the middle 3 seconds. Forces were low passed filtered using a Butterworth filter at 10 Hz (based on residual analysis).

It is unclear whether test battery carry-over effects contributed to the fatigue response or imposed a recovery break during continuous exercise. Using the reference condition, pre- and post-test battery responses for MMG and EMG were measured to determine if test batteries contributed to the fatigue response. There were no statistical differences between 60 seconds before and 60 seconds after the test battery (Table 3.2). To ensure that test batteries didn't influence continuous responses, the first two data points (3-4 minutes), immediately after test batteries, were removed.

Table 3.2 Pre- and post- test battery MMG and EMG Responses to determine test battery carry-over effects.

Measure	Time Interval	Mean (SD)	T(25)	P
MMG RMS	Pre Test Battery	8.448 mV (4.185)	T = 1.142	P = 0.265
	Post Test Battery	9.598 mV (5.699)		
EMG RMS Lateral	Pre Test Battery	2.568 mV (2.643)	T = 0.809	P = 0.426
	Post Test Battery	2.733 mV (2.316)		
EMG RMS Medial	Pre Test Battery	2.435 mV (0.667)	T = 1.114	P = 0.276
	Post Test Battery	2.661 mV (1.036)		

Electromyography was analyzed in both time and frequency domains. Amplification of signals was done using gain settings (x1000) to achieve peak signals of +/-5 V. Raw signals were hardware-filtered at 10-1000 Hz and sampled at 2048 Hz. Root mean square (RMS) values reflect changes in muscular activity amplitude and mean and median power frequencies in the frequency spectra. RMS values were based on raw EMG data in the time domain. Mean (MnPF) and median (MdPF) power frequencies were both based on the mean value of 20 or 5 (sinusoidal), 0.5-second epochs during the 30 second window after a Fast Fourier Transformation analysis. Power frequencies were analyzed at the higher levels of force (of an intermittent contraction) to ensure that the stationarity assumption is met. To quantify the shift of frequency that is indicative of fatigue, the Hi-Lo power ratio (area of high power/area of lower power) was calculated. This technique has been used to quantify the changes in the power spectrum associated with muscular fatigue, as there is a reduction in magnitude of the high-frequency component and an increase in low-frequency component (Moxham et al., 1982). High frequency components were defined as 130 – 238 Hz and low-frequency components were within the range of 20 – 40 Hz (Bigland-Ritchie et al., 1981; Moxham et al., 1982). EMG gaps analysis for both medial and lateral heads of triceps brachii muscle was completed to identify whether there was motor unit activity during the silent period of 0% MVF. EMG gaps were defined as EMG activity less than 1% MVC (Nordander et al., 2000) for durations longer than 0.2 seconds. To identify EMG gaps, signal biases (from quiet baseline) were removed in both the MVF during baseline and EMG activity during exercise. EMG data was then full-wave rectified before being passed through a digital Butterworth low-pass filter at a cut-off frequency of 5 Hz (Zahalak & Pramod, 1985; Gowland et al., 1992) to produce a linear-enveloped EMG. The EMG signal was then normalized to the maximum value obtained at that session's maximum voluntary contraction. EMG gaps analyses were based from these normalized values (Beach et al., 2005). Software was used to identify these gaps for every 15-minute interval and for the entire exercise duration.

Mechanomyography, similar to EMG, was analyzed for RMS amplitude as well as mean and median power frequencies. Signals were sampled at 2048 Hz and bandpassed filtered with a Butterworth filter between 5 and 100 Hz. The lower cut off frequency of 5 Hz extracted artifacts associated with upper limb movement. The upper cut off frequency of 100 Hz was set since no significant frequency components have been reported beyond 100 Hz (Al-Zahrani et al., 2009). Analysis techniques to find RMS and MnPF/MdPF were identical to those to analyze EMG.

Multigon data was used to calculate blood flow to the triceps brachii. Steady-state triceps blood flow (TBF) was calculated using the mean blood velocity (MBV). Mean blood velocity is measured in cm/s, brachial diameter (using sonography) in centimeters, and TBF in milliliters per minute. Triceps blood flow can be defined as:

$$TBF = MBV \times 60 \text{ seconds} \times \text{min}^{-1} \times \pi \times (\text{brachial artery diameter}/2)^2.$$

Mean blood velocity was later adjusted for the 45° embedded angle in the 4 MHz probe. Brachial artery diameter was measured every 2 minutes using both internal landmarks and doppler mode to identify the location of the brachial artery. An indelible marking was drawn on the skin to ensure consistent placement of the ultrasound probe. An intra-rater reliability test was completed using custom ultrasound phantoms of known dimensions. Custom phantoms were composed of precisely measured tubing suspended in a collagen-based gelatin mixture. The CV between measurements was r = 0.961. Brachial artery diameters remained fairly consistent throughout the entire protocol for every individual with a SD range of 0.01 to 0.03 cm for every individual in each condition. As a result, mean blood velocity was used to reflect triceps blood flow (and classified as mean triceps blood flow velocity).

Ratings of perceived exertion were calculated pre-exercise, during exercise, and post-exercise. The rating was collected every two minutes during exercise and expressed as rating between 0 (no exertion) and 100 (maximum/intense exertion).

Prior to analysis, biases identified during baseline trials were subtracted from force values. Mechanical force outputs were measured during the entire exercise duration. The true mechanical force at "0%" and "30%" was measured by identifying the middle 2 seconds of every 3-second interval (50% duty cycle of a 6 second cycle time). The 2-second period was within the plateau phase of the desired force. During the sustained isometric condition, the mean force of a 2-second window every 3 seconds was measured. In the Sinusoidal condition, 0.5 second windows were used to measure mechanical force at peak values. Window lengths were based on programmed force inputs and duty cycle and cycle time parameters.

The parameters analyzed for each measurement are summarized in Table 3.3.

Table 3.3. Measurement Parameters For Each Fatigue Assessment Tool

Measurement Parameters	
Measurement	**Parameter Analyzed**
EMG	Root Mean Square Amplitude (Normalized to Trial MVF) Hi/Lo Frequency Ratio Mean and Median Power Frequencies EMG Gaps (< 1% EMG, > 0.2 seconds)
MMG	Root Mean Square Amplitude (Normalized to Trial MVF) Mean and Median Power Frequencies
RPE	Ratings (1 – 100%)
Doppler Ultrasound	Mean blood velocity of brachial artery (cm/s)
Sonosite Ultrasound	Brachial artery diameter (cm)
Electrical Stimulation, MVF, Test Contractions	Forces (Normalized to Pre-Experiment MVF) Low-frequency fatigue ratio (20/100 Hz) Twitch Force (Normalized to Pre-Experiment MVF)
Mechanical Force	Force measured during theoretical high and low levels of force in 2 second windows Force measured during sinusoidal peaks in 0.5 second windows Force measured with 2 second windows every 3 seconds during isometric condition

Chapter IV

Effects of Mechanical Exposure Variation of Force (Intermittent On/Off) in Comparison to Sustained Isometric Holds

Introduction

A rather new concept in physical ergonomics is the design of work for the purpose of inducing positive health benefits rather than designing work to avoid ill health (Straker & Mathiassen, 2009). Traditionally, the physical ergonomics paradigm was to reduce high physical workloads but has since shifted to "more can be better" due to an increased prevalence of low physical stresses found in sedentary office work. Straker and Mathiassen (2009) argued that the "more can be better" strategy might provide appropriate physical stresses that would benefit workers (overall job performance and satisfaction), employers (improved work quality and productivity), and society (reduced costs associated with job absenteeism and more attractive jobs when recruiting a new workforce). Optimal work patterns may be in the form of motor variability, providing sufficient demands on the neuromuscular and cardiovascular system to maximize positive effects while minimizing the risk of injury.

In a recent cross-sectional study conducted by Madeleine (2010), increased variation in muscle activation was associated with reduced muscle fatigue development. At chronic stages of injury, healthy workers exhibited a larger variability in their motor patterns than workers who reported pain. It was speculated that there is an increased risk of developing WMSD due to less variable motor patterns. In addition, highly experienced workers displayed greater motor variability than novice workers with lower work experience in the same tasks. Madeleine (2010) argues that motor variability may delay muscle fatigue development and may prevent the onset of a work-related musculoskeletal disorder. Although Madeleine (2010) demonstrated evidence for the beneficial value of motor (physical) variability in an occupational setting, there is still a lack of evidence for the physiological and psychophysical effects of physical variation (Mathiassen, 2006).

Westgaard and Winkel (1996) asserted that sustained exertions at low force levels may rapidly lead to fatigue effects and reduced performance and may result in occupational musculoskeletal disorders. By allowing motor units, that would be otherwise overloaded, an opportunity to relax, it is postulated that physical variation will result in reduced rate of fatigue response in biophysical and psychophysical factors. This may be particularly true by having periods of time with minimal or zero activity. Hägg (1991) formulated the Cinderella hypothesis, which suggest that low-threshold motor units are first to be recruited (taking into account Hennemann's size principle, 1965) and are accordingly at risk for metabolic overload, resulting in muscle pain and strain. Research conducted by both Thorn et al., (2002) and Zennaro et al., (2003) support the Cinderella hypothesis. At zero activity, low-threshold motor units may be sufficiently de-recruited to avoid fibre damage during long-term muscle activations. Veiersted and colleagues (1990) suggest that these breaks may be quantified as EMG activity lower than 0.5% MVC for 0.2 seconds or longer. Westgaard (1988) suggested that even at muscle activity less than 1% MVC, there was a high frequency of myalgia, and has since been used as the threshold to determine muscular breaks (Nordander et al., 2000).

This chapter will therefore investigate the effects of physical variation, with periods of zero physical activity and high activity, compared to a sustained isometric contraction at a constant force level. It is hypothesized that the classical intermittent contraction, a repeated cycle that includes zero loading, will show a slower rate of physiological and psychophysical fatigue response when compared to a sustained isometric contraction.

Methods

Participants

Fifteen healthy males participated in the study. All participants completed a self-report checklist and no participants had past or previous health problems. All participants were non-smokers and exercised on an irregular to regular basis. Individuals who smoked or had any current medical problems related to cardiovascular disease, kidney disease, or diabetes were excluded from participation in the study due to its possible effects to blood flow. Participants were fully informed of all experimental procedures and associated risks before written consent was obtained using an information and consent form approved by the UW Office of Research Ethics.

Sustained Isometric vs. On/Off

Participants performed two conditions: a sustained isometric elbow extension contraction at 15% MVF (maximum voluntary force) and an intermittent elbow extension isometric contraction at 0% MVF and 30% MVF. Each condition was performed on separate days, at least 7 days apart to reduce possible carry-over effects from the previous condition. To exclude order effects, the sustained isometric and On/Off conditions were performed in a random order. Participants were instructed to avoid alcohol, caffeine, and exercise involving the upper extremities 24 hours prior to collection. Sufficient practice time for each condition was given to participants to reduce learning effects. Proprioceptive input to accurately trigger joint rotations in a movement sequence requires little or no training with the greatest reduction in variable spatial error in the first 30 trials (Cordo et al., 1994).

Given occupationally relevant forces observed and suggested in past literature, as reviewed in Chapter II, this study used a mean load of 15% MVC, a duty cycle of 50%, and a cycle time of 6 seconds. The 15% MVF was expressed as the proportion of MVF exertion from the largest magnitude of the three pre-experiment MVF contractions. The On/Off condition consisted of forces at 0% and 30% MVF while the sustained isometric condition required participants to sustain a force of 15% MVF (Figure 4.1).

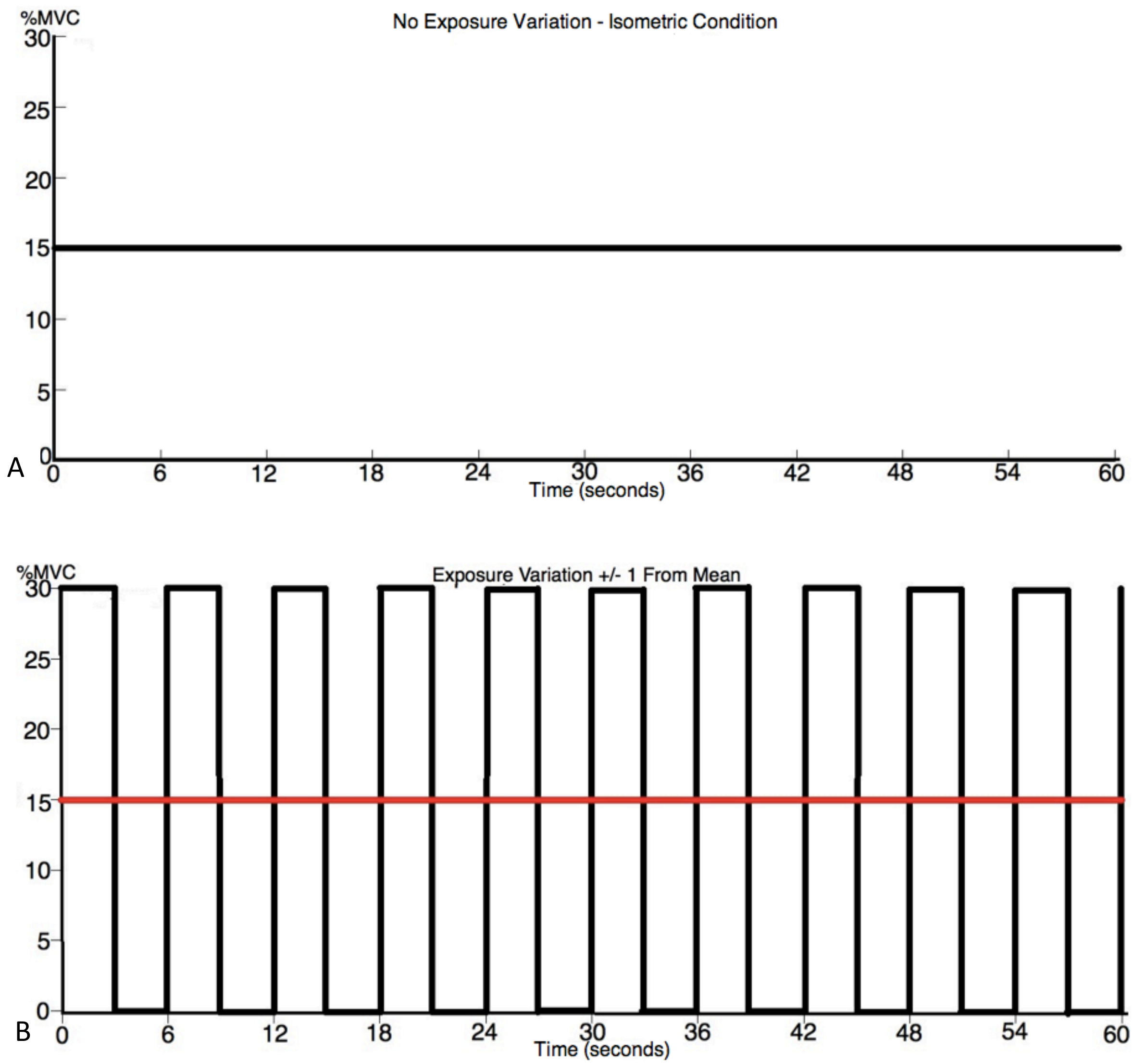


Figure 4.1 Test Conditions: sustained isometric contraction at 15% MVF (A) and On/Off pattern (B) consisting of forces at 0% and 30% with a duty cycle of 50% and cycle time of 6 seconds.

To achieve these force patterns the force generator system was programmed using WSDK Galil software. The programmable force generator system was described in Chapter III. Participants were required to exert resistive forces against loads manipulated by the researcher. As such, this study required position control (proprioceptive feedback) rather than force control (visual feedback). The MVF was converted to a torque value and adjusted to define 0% and 30% of the MVF. During transitions between 0% and 30% of force, the motor was programmed to increase or decrease in seven steps, providing 700 milliseconds to transition and 2.3 seconds at each 0% and 30% plateaus. The transition time of 700 milliseconds allowed participants to neuromechanically adjust to the defined levels as the minimal sensory conduction and processing delay time from a proprioceptive trigger is 210 milliseconds (Cordo et al., 1994). The 700 milliseconds ramp prevented over and under-shooting of the targeted force inputs.

Procedure

General procedures are outlined in Chapter III. To summarize:

In both conditions, 10 minutes of baseline rest activity was recorded while participants quietly sat with their arm in the programmable force generator system. Muscular activity was measured using EMG and MMG. EMG signals were recorded from the lateral and medial heads of the triceps brachii and the biceps brachii muscle. MMG was recorded from the medial head of the triceps, distal to the EMG electrodes.

Blood flow response was monitored by Transcranial Doppler ultrasound and Sonosite Ultrasound. Other biophysical activity to measure fatigue includes muscle stimulation, force production, and ratings of perceived exertion. Two fabricated electrodes were placed on the proximal and distal ends of the triceps to measure low frequency fatigue (LFF) and twitches using electrical stimulation. Forces were measured with a force transducer attached between the motor shaft and armrest. Ratings of perceived exertion, a psychophysical measure, were collected using a visual analogue digital scale.

Each condition was completed for 60 minutes or until exhaustion (Figure 4.2). If participants could not maintain the desired force levels, the trial was terminated at the discretion of the researcher. As outlined in Chapter III, EMG, MMG, blood velocity, force, and heart rate measures were collected continuously but were later sampled for a 30 second window, every 2 minutes. Ultrasound images of the brachial artery diameter were measured at the 2-minute intervals. At the cessation of exercise, further measures of EMG, MMG, and blood velocity were collected for the first 15 minutes of recovery. Test batteries (muscle stimulation, test contractions, maximum voluntary force contraction) were collected every 15 minutes during exercise and every 15 minutes during recovery. A 24-hour post exercise trial was collected, with only the quiet activity and a test battery.

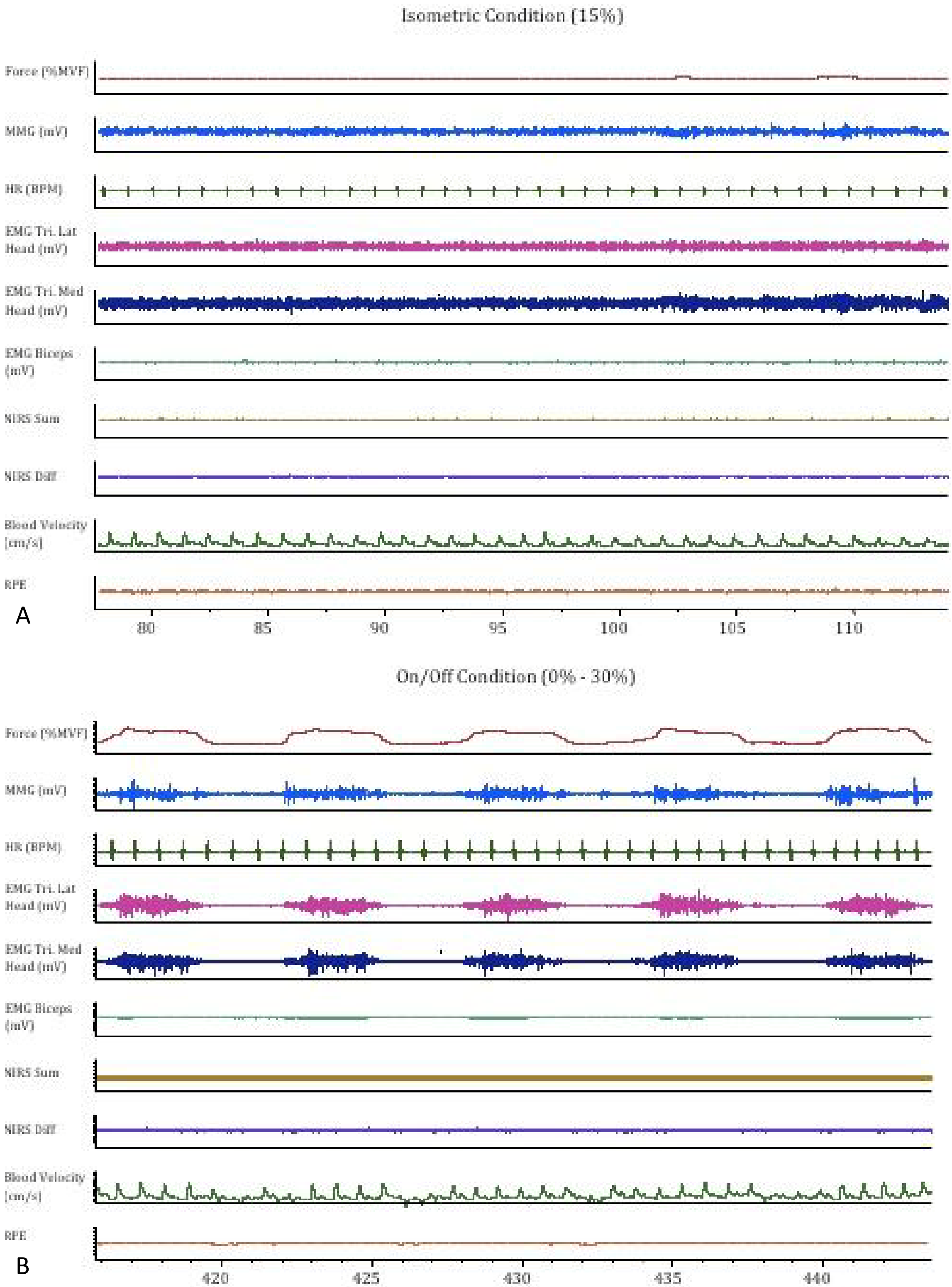


Figure 4.2 Screenshot Data Collection Profiles

Continuous measurements of force (% MVF), MMG (mV), HR (BPM), EMG Lateral (mV), EMG Medial (mV), EMG Biceps (mV), NIRS Sum and Difference, Blood Velocity (cm/s), and RPE (mV) for sustained isometric (A) and On/Off intermittent contraction (B). N.B. Horizontal axes are not "0" in magnitude of response.

Measurement Parameters

Detailed descriptions of the measurement parameters are found in Chapter III. To summarize:

Electromyography was analyzed in both time and frequency domains. Root mean square (RMS) values reflect changes in muscular activity amplitude and mean and median power frequencies in the frequency spectra. To quantify the shift of frequency that is indicative of fatigue, Hi-Lo power ratio (area of high power/area of lower power) was calculated. EMG gaps identified whether there was motor unit activity during the silent period of 0% MVF.

Mechanomyography was analyzed for RMS amplitude as well as mean and median power frequencies. Analysis techniques to find RMS and MnPF/MdPF were identical to those to analyze EMG.

Steady-state triceps blood flow (TBF) was calculated using the mean blood velocity (MBV). Mean blood velocity was used to reflect triceps blood flow.

Ratings of perceived exertion were calculated pre-exercise, during exercise, and post-exercise. The rating was collected every two minutes during exercise and expressed as rating between 0 (no exertion) and 100 (maximum/intense exertion).

To determine if the conditions lead to fatigue, a test battery consisting of electrical stimulation, test contractions, and maximum voluntary of force contractions (MVF) was used to determine the effects of the contraction pattern. Test contractions were elicited to measure EMG RMS, MnPF, and MdPF and MMG RMS, MnPF, and MdPF values at a consistent level of force (15% MVF).

Although participants exerted forces that were manipulated by the researcher, the true mechanical forces that were exerted may differ. The true mechanical force at "0%" and "30%" was measured by identifying the middle 2 seconds of every 3-second interval (50% duty cycle of a 6 second cycle time). The 2-second period was within the plateau phase of the desired force. During the sustained isometric condition, the mean force of a 2-second window every 3 seconds was measured.

Statistical Analysis

Normality of the data was assessed prior to statistical testing using p-p and q-q plots. Measured parameters were plotted against time and fitted with either a linear or a non-linear regression line. Determination of the model to fit the response was based on achieving highest r squared values in all conditions to allow for equitable comparisons. Paired t-tests were used to compare the regression coefficient (slopes) between conditions. Since hypothesis driven questions were undertaken, a one-tailed *a priori* analysis was used to compare sustained isometric and On/Off conditions.

Force, test contractions, low-frequency fatigue, and twitch force were measured at baseline (pre) and compared to values at cessation (post) of exercise, at 15-minute intervals into recovery, and 24 hours post exercise. These test batteries were also measured during exercise, in 15-minute intervals, and both rate of response during exercise and recovery were analyzed using linear and non-linear regression as described above. Forces during the test session and 24-hours post exercise were normalized to the baseline maximum voluntary force elicited at the beginning of the entire protocol. These comparisons were analyzed using one-way repeated measures ANOVA and a Dunnett's test to compare cessation and recovery values against baseline. Mauchley's test of sphericity determined if the sphericity assumption was met. If the assumption was not met, the appropriate follow-up test (i.e. Greenhouse Geiser or Huynh-Feldt analysis) was undertaken. Partial eta squared values (η_p^2) were reported to determine the effect size of the repeated measures ANOVA analysis.

Test contraction values collected during exercise and 60 minute recovery were normalized to baseline. When comparing values 24 hours post exercise and at baseline, both were normalized to the pre-experiment maximum voluntary force contractions.

Completion times, the number of EMG gaps per minute, and mean and median duration of each gap for both sustained isometric and On/Off conditions were compared using Wilcoxon signed ranks test. Mechanical force outputs were compared between the beginning and end of exercise. The beginning of exercise was defined as the first 10 contractions (for each of the "on" or "off" levels of force) or the first 30 seconds of sustained isometric exercise. The end of exercise was regarded as the last 10 contractions prior to exercise cessation for the intermittent condition or the last 30 seconds for sustained isometric. The average mechanical force outputs were analyzed for the entire exercise protocol.

Full statistical results during exercise are presented in Appendix A and during recovery in Appendix C. Values during cessation and recovery that are statistically significant than baseline are marked with a star.

Measurement Results and Preliminary Discussion

Endurance Time Results

Participants were able to exert necessary forces to attain an intermittent square wave isometric contraction between 0% and 30% and a static sustained isometric contraction at 15% MVF. Participants performed both conditions for 60 minutes or until exhaustion. Endurance times are shown in Figure 4.3. A Wilcoxon signed-ranks test indicated that the On/Off condition (Median = 3600 seconds, 25^{th} = 2274, 75^{th} = 3600) led to longer completion times than the sustained isometric condition (Median = 579 seconds, 25^{th} = 408, 75^{th} = 1191.50), Z = 2.93, p = 0.003.

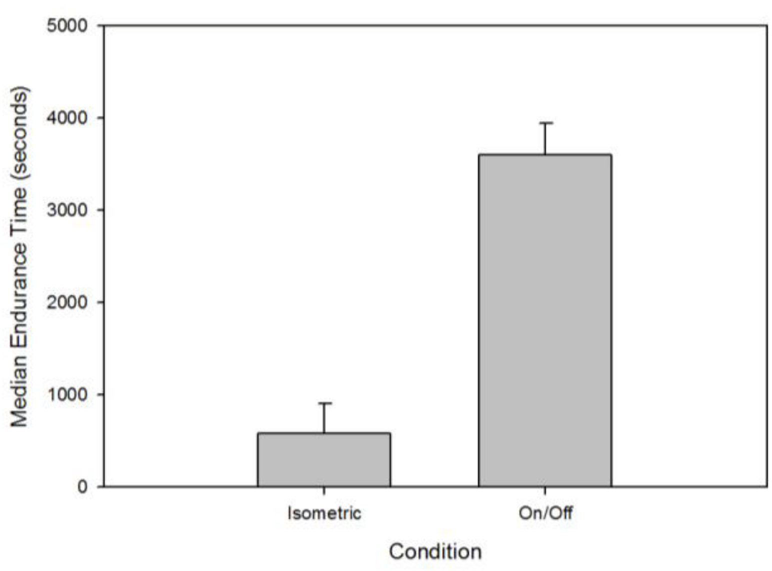


Figure 4.3 Median Endurance Times for Sustained Isometric and On/Off.

Endurance Time Preliminary Discussion

Endurance times for both conditions suggest that participants were able to perform the desired intermittent contraction pattern longer than the sustained isometric hold. Participants were able to exert a sustained isometric contraction at 15% MVF for a median duration of 579 seconds (9 minutes, 39 seconds). This result is similar to observations by Mathiassen and Winkel (1992) who found a median endurance time of 13 minutes and 12 seconds for an isometric load between 13% and 18% MVC. Krogh-Lund & JØrgensen (1992) identified an endurance time of 15.1 minutes (906 seconds) for a 15% MVC elbow flexion. This result is in contrast to Rohmert's (1973) endurance limit of 15% MVF that could be sustained for an "unlimited" period of time without reduction of force. However, Rohmert (1973) assumed that an isometric contraction, which could be sustained for 10 – 15 minutes, could be sustained for an "unlimited" period of time. Interpreted differently, a sustained isometric force of 15% MVF can be elicited for an endurance time of 10 to 15 minutes.

The intermittent contraction had a median endurance time of 3600 seconds (60 minutes) indicating that more than half of the participants were able to sustain this contraction for the entire protocol. In a study by Bjorksten and Jonsson (1977), various intermittent contractions were performed for 60 minutes. It was found that participants were able to perform an intermittent contraction pattern (50% duty cycle, 10 second cycle time) with a mean force of 13.9% MVC for the 60 minute duration. The results of this study therefore agree with findings by Bjorksten and Jonsson (1977).

Force Results

Maximum voluntary force measurements were collected at baseline, during exercise test batteries, and during recovery. The rate of force decrement was analyzed between conditions. Responses were fitted by linear regression (sustained isometric: mean $r^2 = 0.948$, On/Off: mean $r^2 = 0.622$) and compared. A typical response with regression fit is shown in Figure 4.4. On/Off condition (M = -6.754% MVF/Test Battery, SD = 8.7159) had a significantly slower rate of force decrement than the sustained isometric condition (M = -14.408% MVF/Test Battery, SD = 14.279), $t(13) = 2.681$, $p = 0.01$ (one-tailed), $d = 0.647$.

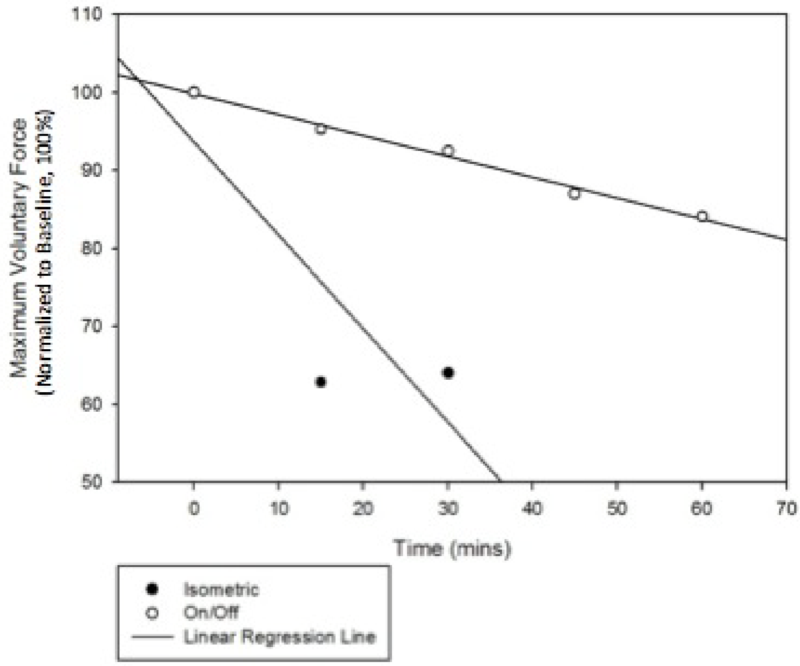
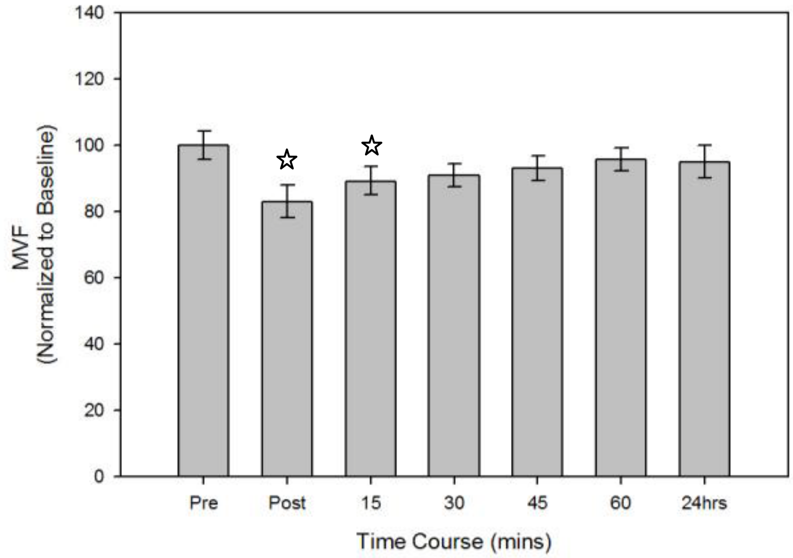


Figure 4.4 Typical Maximum Voluntary Force Response During Exercise for Sustained and On/Off Conditions

Baseline force values were compared to recovery at 15, 30, 45, 60 minutes, and 24 hours (Figure 4.5). For graphical purposes, forces at cessation (post), during recovery, and 24-hours post exercise were normalized to the condition's baseline (pre) value.



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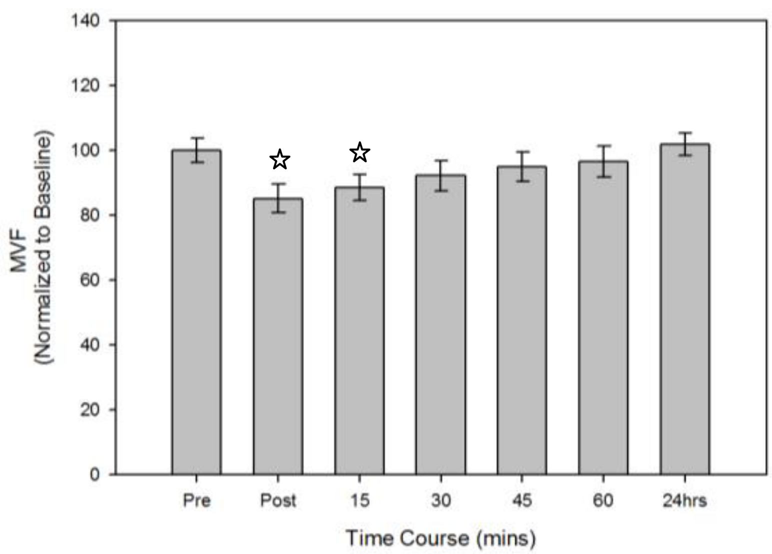


B

Figure 4.5 Max Voluntary Force Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B)

Most notably, both sustained isometric ($F(0,6) = 3.465$, $p = 0.041$, $\eta_p^2 = 0.224$; baseline (M = 80.470% MVF, SD = 15.143); cessation (M = 66.749% MVF, SD = 14.278) and On/Off conditions ($F(0,6) = 4.510$, $p = 0.001$, $\eta_p^2 = 0.273$; baseline (M = 78.512% MVF, SD = 13.268); cessation (M = 66.891% MVF, SD = 66.891) led to a decrement of maximum voluntary force at the end of the exercise session. Likewise, both sustained isometric (M = 72.119% MVF, SD = 10.074) and On/Off (M = 69.465% MVF, SD = 11.123) conditions had statistically significant decreased force values at 15 minutes into recovery.

Force Preliminary Discussion

These results are consistent with past literature that has observed a reduction in maximal force from the start of exercise (Vøllestad, 1997) and after sustained low-intensity contractions (Søgaard et al., 2006). According to Vøllestad (1997), the balance of Na^+ and K^+ ions over the sarcolemma and *t*-tubule membrane was compromised resulting in impairment in the propagation of action potentials. Consequently, there was a decrease in the amount of Ca^{2+} released from the sarcoplasmic reticulum into the cytosol, further decreasing the binding between Ca^{2+} and troponin C (Westerblad & Allen, 1991). Søgaard and colleagues (2006) suggest that muscle fatigue resulting from low-level forces may be largely attributed to a decrease in calcium release. Additionally, a reduction in tension may be due to reduced myofibrillar Ca^{2+} sensitivity. As a result, fewer cross-bridges are formed between actin and myosin molecules, leading to a decrease in force production. It has also been shown that the accumulation of metabolites can reduce the affinity of Ca^{2+} binding to troponin (Vøllestad, 1997; Westerblad and Allen, 1991). For instance, lactic acid may produce intracellular acidosis that reduces the maximum Ca^{2+} -activated tension (Westerblad & Allen, 1991).

The varying degree of these metabolic and physiological changes may be dependent on the different work regimes or fatigue protocols and may be explained by the motor unit activation pattern (Vøllestad, 1997; Søgaard et al., 2006). The sequence of motor unit recruitment is commonly described as slow twitch (type I) to fast fatigue resistant (type IIA) to fast fatigable (type IIB) and may have intermediate fiber types (IIAB or IIX). When progressively exerting submaximal contractions, type I fibres are generally recruited first and later type II fibres. At exhaustion all motor units are active. In the sustained isometric condition, contractile slowing is commonly observed and motor unit excitation rate decline (Vøllestad, 1997). However, during repetitive isometric contractions, Vøllestad (1997) suggest that an oscillating force is generated at the motor unit level as the interval between excitation pulses are longer than the rise time of the force. As force rises and falls according to each excitation pulse, there may be a higher energy demand than an isometric mean force. Although intermittent isometric contractions may lead to higher energy demand, the fall in maximum force generation is distributed over a longer period of time, as reflected in the rate of force decrement and median completion times. This was also not apparent during recovery where there was no difference between conditions in both rate and baseline-recovery comparisons. This may imply that the force rise time was sufficient to allow excitation pulses to follow the contraction inputs. Alternatively, the active recruitment and de-recruitment of motor units, inducing successive rotation of motor units, supplemented with sufficient rest periods may have played a role in decreased rate of force response during exercise and no difference in the rate of recovery between conditions. Another consideration is the extent (or lack) of "fatigue" in the intermittent contraction. Possibly the decrement of fatigue during recovery may be prolonged after an intermittent contraction if the On/Off condition was done until exhaustion in which case a higher demand due to the discrepancy between excitation pulse and contraction inputs.

Twitch Results

Twitch force during exercise was fitted with a linear regression line (Figure 4.6) with mean goodness of fit $r^2 = 0.943$ (sustained isometric) and $r^2 = 0.693$ (On/Off). Sustained isometric (M = -22.460% MVF/Test Battery, SD = 26.216) contractions had a quicker decreasing rate of twitch force compared to On/Off (M = -9.757% MVF/Test Battery, SD = 14.668) contractions, $t(13) = -2.229, p = 0.023, d = -0.598$.

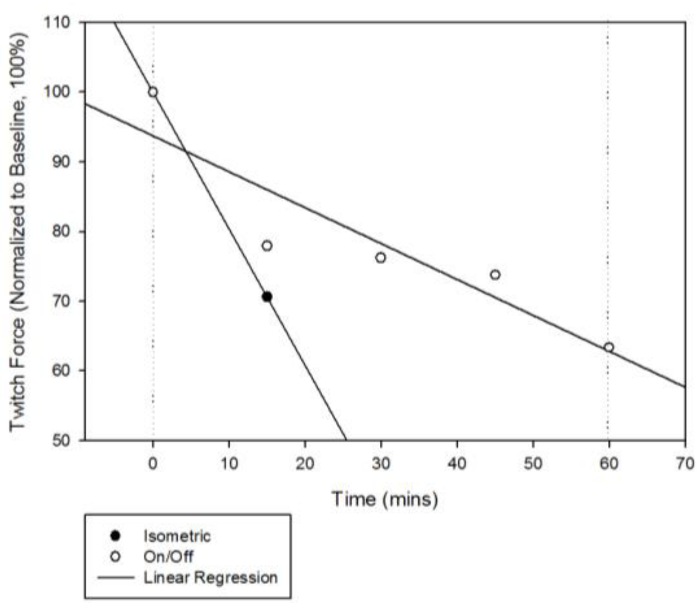


Figure 4.6 Typical Twitch Force Response During Exercise for Sustained and On/Off

Twitch values between baseline and cessation revealed statistically significant differences for sustained isometric [F(0,6) = 7.346, P = 0.000, $\eta_p^2 = 0.380$, baseline (M = 5.234% MVF, SD = 2.271), cessation (M = 3.222% MVF, SD = 1.570) and the On/Off condition [F(0,6) = 3.047% MVF, p = 0.057, $\eta_p^2 = 0.203$, baseline (M = 5.609% MVF, SD = 2.639), cessation (M = 4.326% MVF, SD = 4.152). At 15 minutes recovery, there was a decrease in twitch force in the sustained isometric condition (M = 3.465% MVF, SD = 1.882) but no significant difference after On/Off exercise (M = 4.557% MVF, SD = 4.055). At 24-hours post exercise, sustained isometric condition had a significant decreased twitch force when compared to baseline [24 hour (M = 4.316% MVF, SD = 2.390)]. Comparisons between baseline and recovery are shown in Figure 4.7.

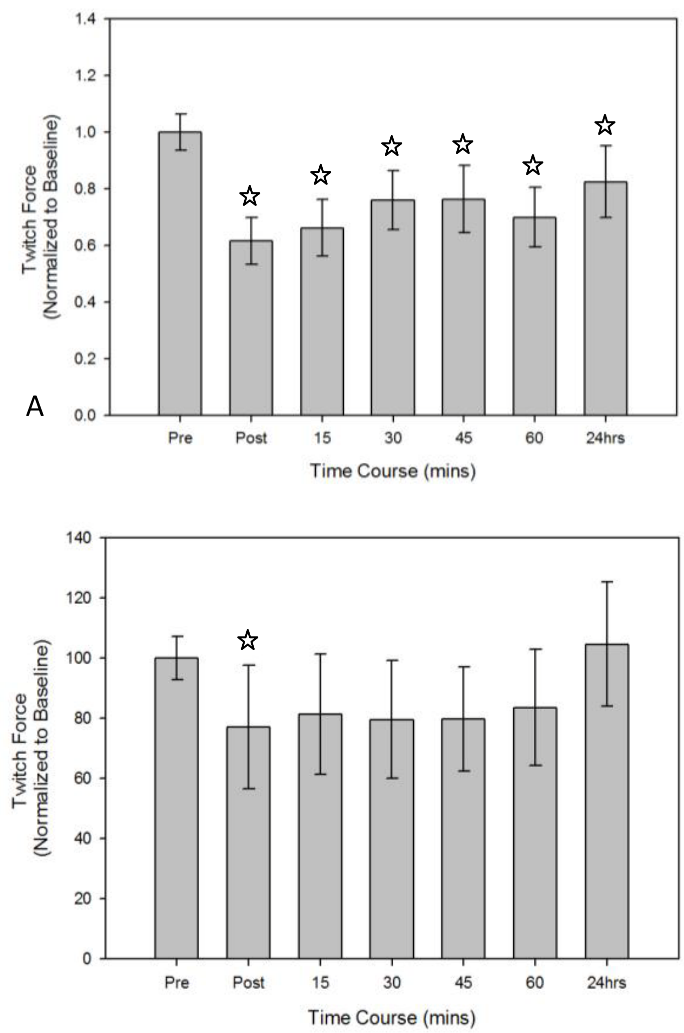


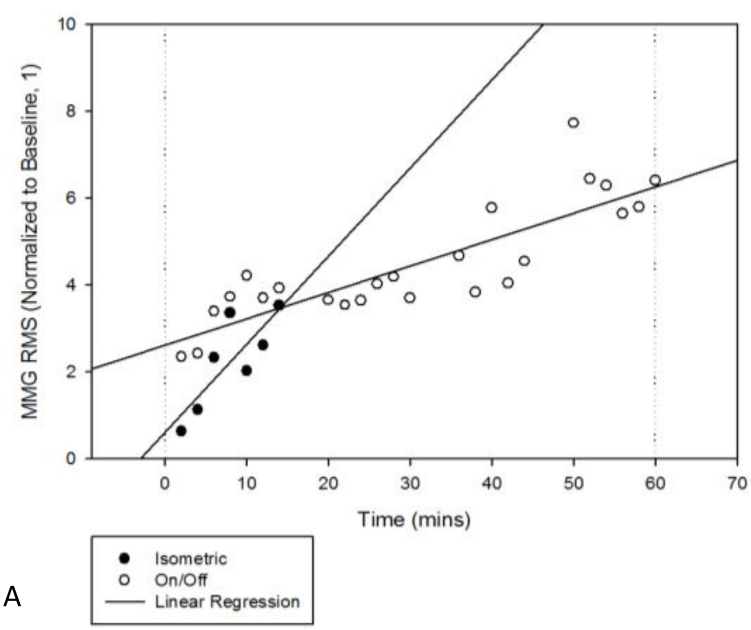
Figure 4.7 Twitch Force Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B)

Twitch Preliminary Discussion

Twitch is often used to assess force-generating potential. Like MVF, twitch may be an indicator of the excitation of motor units, the release of Ca^{2+} into the cytosol and it's binding to troponin, and the cross-bridge turnover and ATP utilization/regeneration processes (Vøllestad, 1997). Twitch force may also be as good an indicator as the response elicited by low frequency tetani of long-term fatigue (Edwards et al., 1977). In both sustained isometric and On/Off conditions, force potential decreased at the conclusion of exercise with a substantial rate of decrease in the sustained isometric condition. However, during recovery, although there was no statistical difference between the rates of recovery, sustained isometric contractions led to depressed twitch force up to 24 hours post-exercise. This is similar to findings observed by Blangsted and colleagues (2005) who found a decrease in peak twitch force stimulation up to 150 minutes after a 10% MVC isometric wrist extension and up to 60 minutes by Søgaard and colleagues (2006) after an isometric elbow flexion at 15% MVC. Bystrom and Fransson-Hall (1994) found a decrease in electrically stimulated force 24 hours after exposure to a sustained contraction at 25% MVC. Since forces exerted by participants selectively fatigue lower threshold motor units, the peak twitch force, evoking both a mixture of low and high-threshold motor units, may reflect the decreased force contribution from fatigued low-threshold motor units. The On/Off condition, on the other hand, revealed depressed, but not statistically different, peak twitch force after 15 minutes recovery. This may suggest that a sustained isometric contraction leads to both a quick rate of decreased force production and sustained depression during recovery, mostly attributed to an impairment of the excitation-contraction coupling (Blangsted et al., 2005). These results appear to be in contradiction with results found by Bystrom and Fransson-Hall (1994) who observed persistent muscle fatigue of the extensor digitorum communis 24 hours after exposure to intermittent (≥ 17% MVC) contractions. This discrepancy may be explained by the intensity of the exercise protocol. In the Bystrom and Fransson-Hall (1994) study, at 40% MVC intermittent, an increase in the overall fatigue response may have occurred, thus delaying recovery. The results, however, were similar to a study by Baker and colleagues (1993). In that study, Baker et al. (1993) found a decrease in twitch force to 16.9% of the initial twitch contraction, 15 minutes into recovery, after exposure to 20 minutes of intermittent maximal contractions. Similarly, an intermittent isometric contraction (10 second cycle time, 40% contraction duty cycle) at 10% MVC with increments of force by 10% every 2 minutes led to a reduction in twitch force post-exercise (Kent-Braun et al., 2002). A reduction of muscle tension may be the result of the loss of K^+ in the transverse muscle membrane folds and accumulation of H^+ ions. This in turn may decrease Ca^{2+} sensitivity of the myofilament. Slow recovery may also be attributed to the slow process in the restoration of K^+ and Ca^{2+} in the sarcoplasmic reticulum (Adamo et al., 2009).

Continuous Measures RMS Results

Continuous measures of MMG and EMG of the lateral and medial heads were continuously monitored for the duration of the exercise protocol (Figure 4.8). MMG RMS values revealed a greater rate of response during sustained isometric contraction (M = 8.514%/min, SD = 8.525) than On/Off (M = 0.979%/min, SD = 1.669), $t(13) = 3.713, p = 0.002, d = 1.383$. Sustained (M = 6.319%/min, SD = 8.905) led to a quicker response than On/Off (M = 0.7625%/min, SD = 2.004) when observing EMG RMS values of the triceps brachii medial head, $t(13) = 2.175, p = 0.025, d = 0.861$. Similar results were found in the lateral head. Sustained isometric condition (M = 10.450%/min, SD = 13.378) led to a quicker rate of response than On/Off (M = 1.592%/min, SD = 3.698), $t(13) = 2.254, p = 0.022, d = 0.903$.



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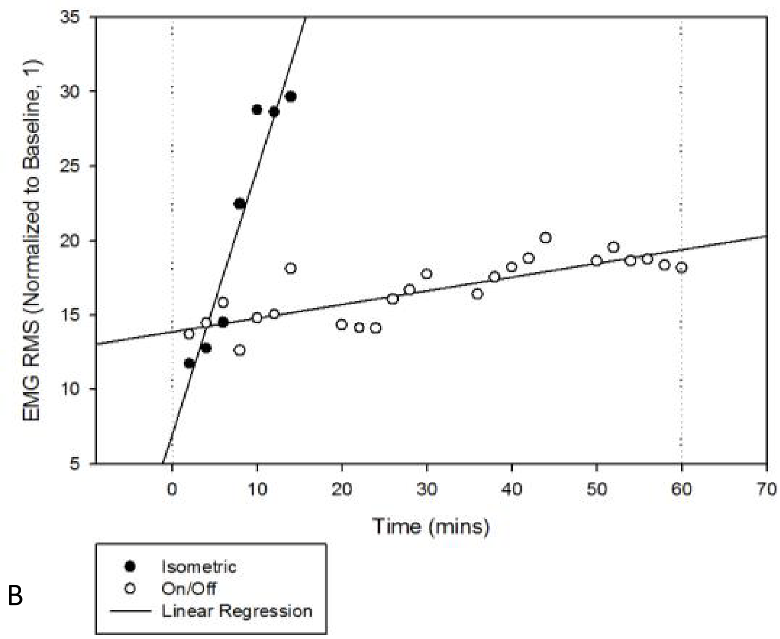


Figure 4.8 Typical Continuous RMS Response During Exercise in MMG (A), EMG (B)

Test Contractions RMS Results

At intervals of 15 minutes, a test contraction at 15% MVF was collected for mechanomyography and electromyography of the lateral and medial heads of the triceps brachii muscle. Test contractions were fitted with non-linear regression lines and compared using pairwise t-tests to test differences between sustained isometric and On/Off conditions. Figure 4.9A is a typical regression using logarithmic fits for MMG during exercise (sustained isometric: mean $r^2 = 0.930$, On/Off: mean $r^2 = 0.293$). EMG RMS test contractions of the medial (sustained isometric: mean $r^2 = 0.927$, On/Off: mean $r^2 = 0.549$) and lateral heads (sustained isometric: mean $r^2 = 0.948$, On/Off: mean $r^2 = 0.393$) were similarly fitted with logarithmic regression lines (Figure 4.9B).

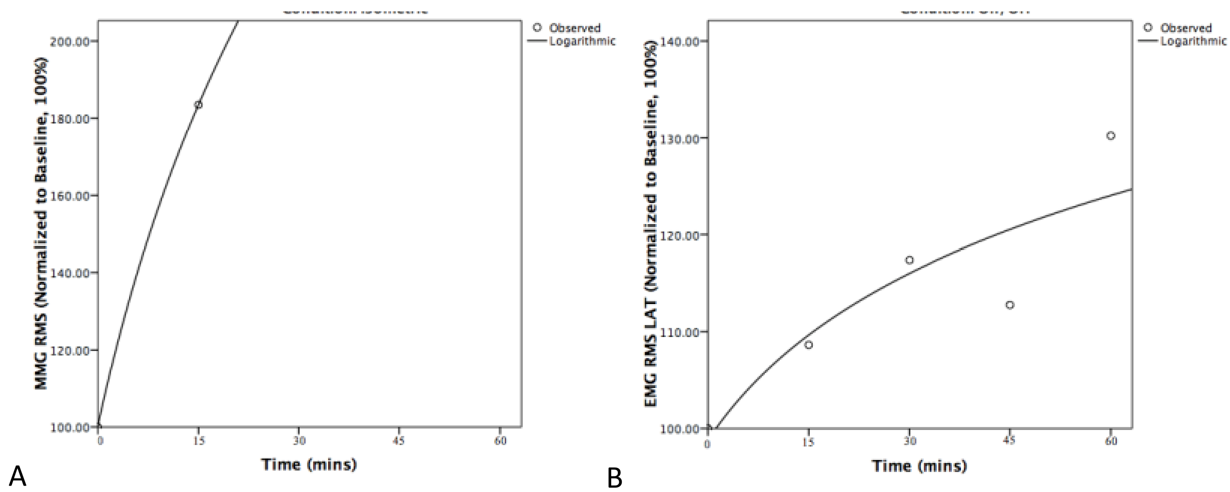


Figure 4.9 Typical Test Contraction MMG (A) and EMG RMS (B) Response During Exercise

A quicker rate of MMG RMS amplitude increase in the sustained isometric (M = 112.028% MVC/Test Battery, SD = 79.186) condition than the On/Off (M = 21.166% MVC/Test Battery, SD = 38.218) condition, *$t(13) = 3.489$, $p = 0.002$, $d = 1.461$,* was observed. A comparison between baseline (M = 15.797% MVC, SD = 10.455) and cessation of exercise (M = 32.276% MVC, SD = 21.208) using MMG indicated large RMS amplitude changes in the sustained isometric condition, $F(0,6) = 7.152$, $p = 0.000$, $\eta_p^2 = 0.373$. MMG RMS values were significantly larger than baseline at 15 minutes (M = 32.507% MVC, SD = 21.208) and 30 minutes (M = 24.325% MVC, SD = 16.043) into recovery. In the On/Off condition, RMS amplitude changes were observed between baseline (M = 16.850% MVC, SD = 8.532) and cessation of exercise (M = 22.568% MVC, SD = 15.730), $F(0,6) = 3.455$, $p = 0.004$, $\eta_p^2 = 0.224$. MMG RMS values after On/Off were not significantly different than baseline at 15 minutes into recovery (M = 20.863% MVC, SD = 9.354). Figure 4.10 presents the comparison between baseline, cessation, and recovery.

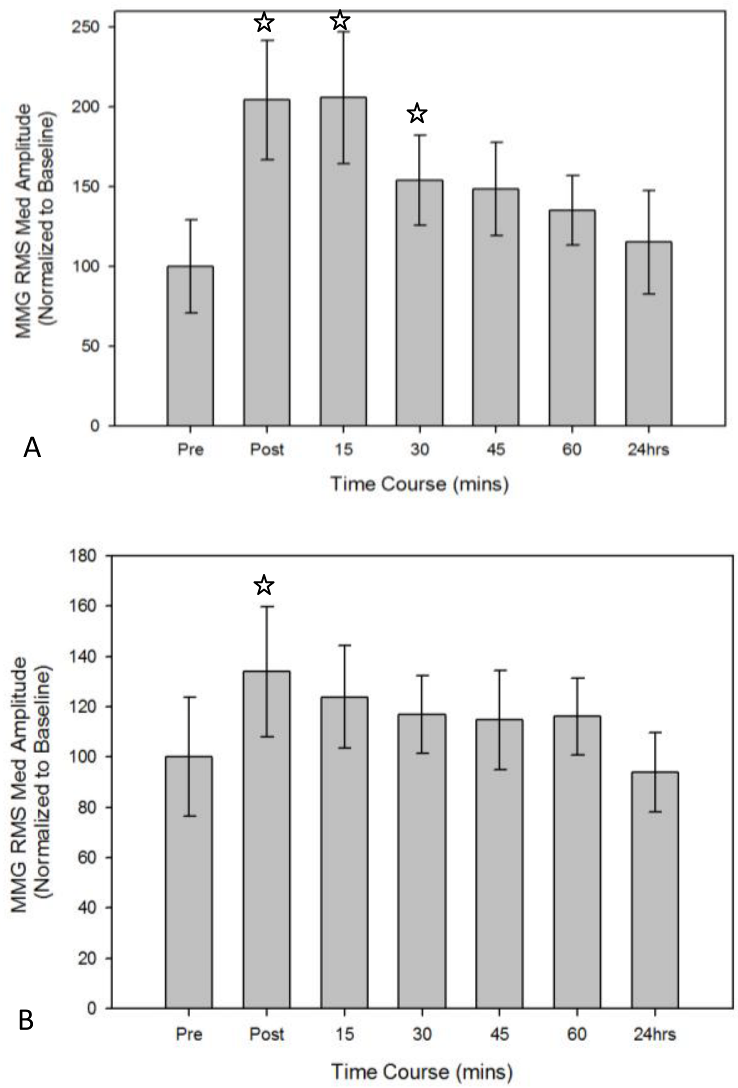


Figure 4.10 MMG RMS Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B)

Sustained isometric (M = 64.757%/Test Battery, SD = 87.953) also led to a quicker rate of amplitude increase in EMG of the medial head than On/Off condition (M = 0.117%/Test Battery, SD = 15.827), *t(13) = 2.920, p = 0.007, d = 1.023*. In the sustained isometric condition, there was a significant difference between baseline (M = 18.516% MVC, SD = 6.444), cessation (M = 26.714% MVC, SD = 10.590), and at 15 minutes recovery (M = 24.950% MVC, SD = 10.934), F(0,6) = 2.610, p = 0.083, $\eta_p^2 = 0.279$. EMG RMS amplitudes of the medial head at cessation (M = 18.070% MVC, SD = 6.760) were not significantly different than baseline (M = 16.825% MVC, SD = 5.823) in the On/Off condition, $F(0,6) = 1.150$, $p = 0.341$, $\eta_p^2 = 0.087$ (Figure 4.11).

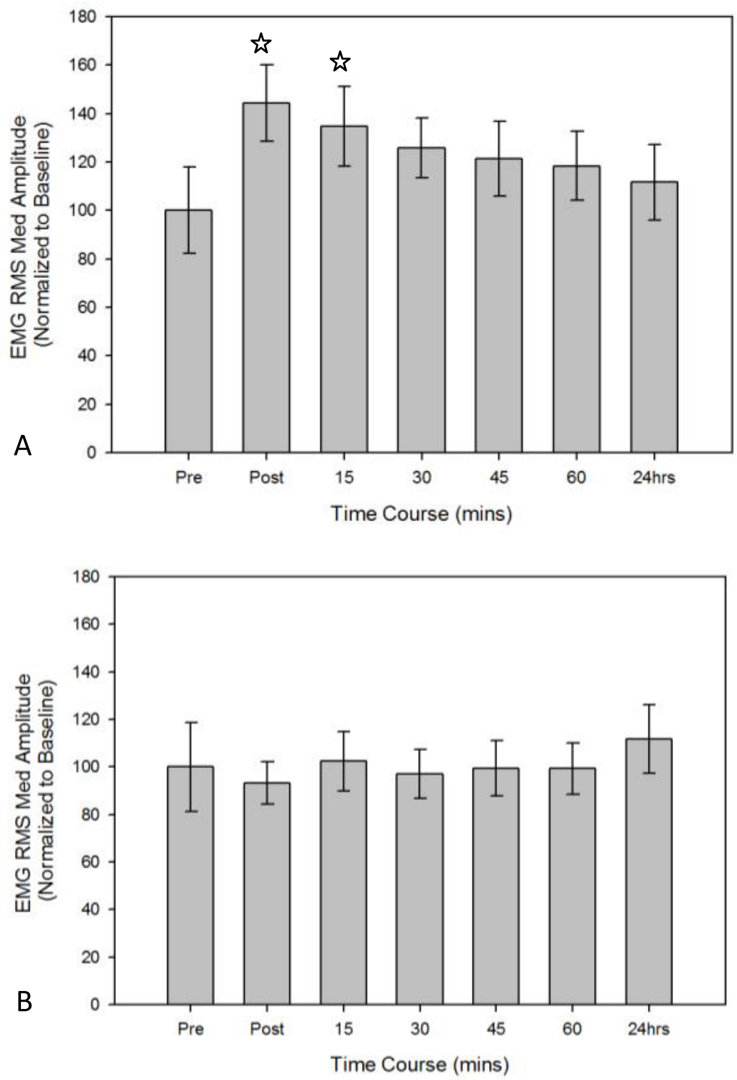


Figure 4.11 EMG RMS Medial Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B)

Muscle activity measured by EMG was also collected on the lateral head (Figure 4.12). Similar to the amplitude response of the medial head, there was a statistical difference between sustained isometric (M = 49.244%/Test Battery, SD = 55.992) and On/Off (M = 15.312%/Test Battery, SD = 38.844) conditions, $t(13) = 1.865, p = 0.043, d = 0.704$. An analysis at the cessation of exercise revealed that after a sustained isometric contraction, the RMS EMG amplitude of the lateral head was greater at the end of exercise (M =

24.877% MVC, SD = 12.523) than baseline (M = 16.066% MVC, SD = 7.126), t(13) = 2.667, p = 0.020, d = 0.865. It was not until 60 minutes of recovery (M = 20.661% MVC, SD = 8.670) when EMG RMS test contractions were not significantly different than baseline. The On/Off condition had recovery values that were not significantly different than baseline (M = 17.870% MVC, SD = 7.434).

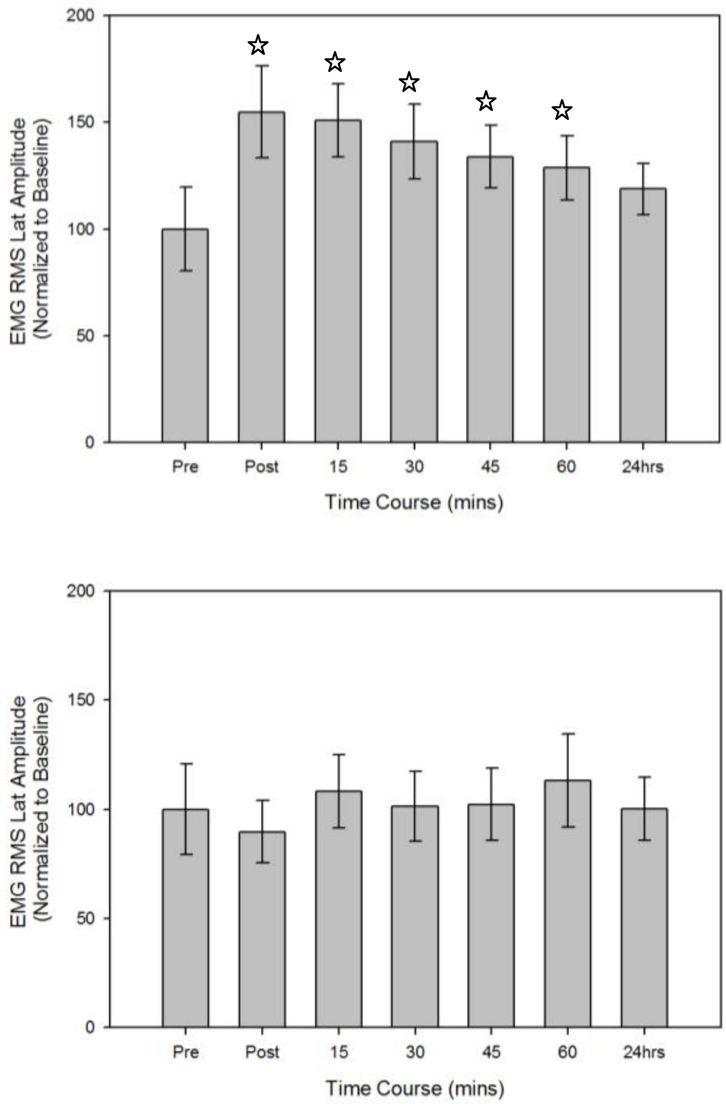


Figure 4.12 EMG RMS Lateral Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B)

RMS Preliminary Discussion

MMG RMS results for the sustained isometric condition were similar to those found by Søgaard and colleagues (2003) who observed increases in MMG amplitude in the time domain using submaximal test contractions at 5% MVC. Similar findings by Blangsted and colleagues (2005) revealed similar effects during 10 minutes of a 10% MVC sustained isometric contraction and during 5% MVC test contractions performed pre-test and during 10 and 30 minutes recovery. The increase in MMG response may be due to the recruitment of additional motor units to generate the same force output over time (Søgaard et al., 2003), reflective of the motor unit activation strategy (motor unit recruitment and firing rate). Theoretically, the recruitment of fast-twitch motor units may lead to greater muscle fiber oscillations that would reflect as an increase in MMG amplitudes (Perry et al., 2001). It has also been suggested that the MMG signal reflects the underlying cross-bridge cycling mechanisms, the process of attachment and detachment between myofilament actin and myosin during contraction (Shinohara & Søgaard, 2006; Vedsted et al., 2006). MMG RMS results for the intermittent contraction pattern were also similar to previous research. An initial "plateau"/decrease of MMG RMS values during continuous intermittent exercise may reflect the de-recruitment of fatigued motor units and a subsequent increase may reflect changes in local muscle fatigue tremor at late stages of fatigue (Al-Zahrani et al., 2009). This local muscle fatigue tremor may be due to increased peripheral fatigue as intermittent contractions may allow muscles to perform greater amounts of work due to improved blood perfusion. Ebersole and colleagues (2006) suggest that initial MMG amplitude decreases are attributed to a decrease in muscular compliance, namely an increase in intramuscular pressure fluid. This fluid pressure increase may be due to the combined increases of muscle thickness, fluid content, and intramuscular pressure (Ebersole et al., 2006). This increase in fluid pressure may restrict lateral muscle fiber oscillations, thereby decreasing the MMG amplitude. However, there is debate as to whether intramuscular fluid pressure has an influence in MMG amplitude (Søgaard et al., 2006).

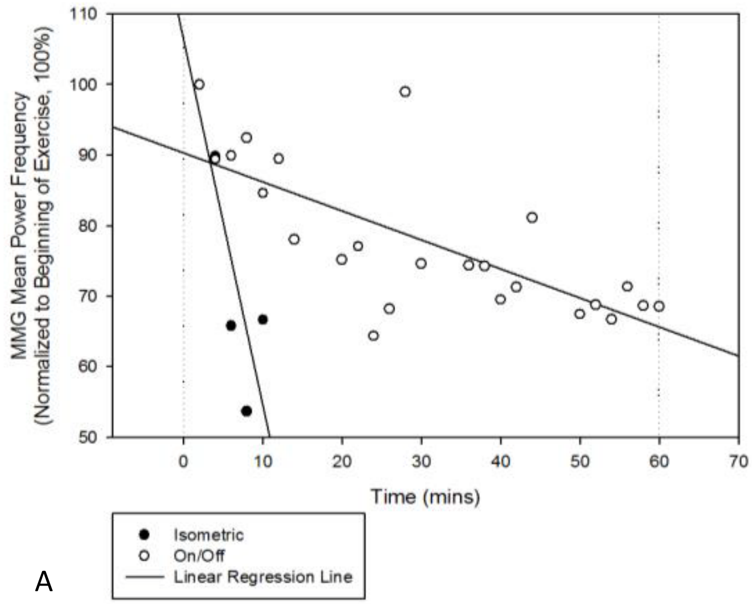
The findings of this study suggest that MMG RMS values recovered quickly after a sustained isometric test but were not significantly different from baseline after 45 minutes of recovery. The quick recovery in the sustained isometric condition may be due to a better recovery in muscle tissue oxygenation (Vedsted et al., 2006). The classical intermittent contraction pattern led to a significantly larger MMG RMS value at cessation of exercise but no significant differences between baseline and recovery intervals. Unlike previous studies that stipulate prolonged recovery and greater peripheral fatigue after an intermittent contraction, possibly due to changes in potassium homeostasis resulting from a longer exposure time (Al-Zahrani et al., 2009), comparison of recovery values against baseline in the On/Off condition suggest otherwise. Quite possibly the extent of fatigue in the On/Off condition was not as 'far reaching' as the sustained isometric protocol (as exercise was terminated at 60 minutes) and is reflected by the rate of fatigue response during exercise based on both continuous and test contraction data. Both rates of response during continuous measurement and test contractions at 15% MVF revealed quicker rates of fatigue in the sustained isometric condition. The On/Off condition led to a higher cessation value but with a slower rate of response, was still within range of the baseline RMS.

Past literature have shown that an increase in EMG RMS amplitude reflect the recruitment of additional motor units to perform the same amount of force. However, a decreased firing rate has been previously shown to reduce EMG amplitude, potentially cancelling out the increase due to recruitment (Søgaard et al., 2003). This mechanism may be due to "muscle wisdom" which serves to maintain force by protecting against conduction failure and by optimizing the input to motor units as contractile properties change. However, this theory has been contested (Fuglevand & Keen, 2003). The results from this study revealed increases in continuous EMG RMS amplitudes in both sustained isometric and On/Off conditions for medial and lateral heads of the triceps brachii. A rise in the EMG after intermittent contractions may be due to a slight rise in excitation rate (Bigland-Ritchie et al., 1986). Although these results are similar to previous findings in both submaximal sustained isometric and intermittent contractions (Søgaard et al., 2003; Vøllestad et al., 1997), there remain inconsistencies of this response, suggesting that EMG amplitude parameters may not be a reliable indicator of muscle fatigue (Bajaj et al., 2002; Vøllestad et al., 1997). Basmajian and DeLuca (1985), suggest that EMG may be a good index of muscle activation only during controlled sustained isometric conditions. As a result, test contractions were performed at a standardized level of sustained isometric force. The EMG RMS response of both heads using test contractions also revealed a quicker rate of response during a sustained isometric contraction than On/Off.

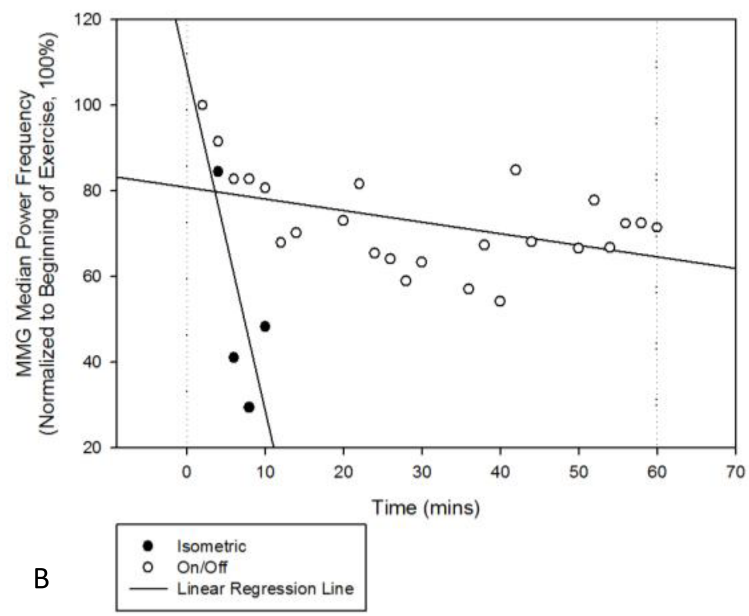
A change in the rate of force may alter the relationship between EMG and activation (Vøllestad et al., 1997), and thus the change (or lack thereof) at cessation was not surprising. In both the medial and lateral heads, cessation RMS values decreased, contrary to the widespread belief of increased EMG RMS amplitude. This study also revealed long-term decreases in EMG RMS amplitudes after a sustained isometric contraction, with a prolonged effect of up to 60 minutes of recovery. Previous research has shown a disparity between EMG and MMG recordings of signals during exercise and post-exercise recovery. Søgaard and colleagues (2003) found a less pronounced EMG response compared to MMG. This was observed during the exercise contraction, where an MMG increase in both conditions was accentuated when compared to the EMG response. Additionally, Søgaard et al. (2003) found further increases in MMG RMS after 10 minutes recovery whereas EMG RMS began to decline. This was also identified in this study where MMG RMS responses increased at 15 minutes recovery after sustained isometric contraction and remained significantly different from baseline until 30 minutes recovery. EMG RMS, on the other hand, showed higher values but did not increase further in recovery. The On/Off condition revealed statistically significant higher RMS values at cessation using MMG but not with EMG. The rate of recovery is misleading. Although sustained isometric appears to lead to a quicker rate of recovery, the magnitude of the RMS value at cessation was lower during On/Off. This may imply that given a pre-defined workload based on time, and not based on exhaustion of all conditions, the recovery rate after a sustained isometric condition was quicker but at the expense of shortened exercise duration. Consequently, both RMS analysis in MMG and EMG showed higher rate of fatigue during exercise and possible prolonged effects during recovery after a sustained isometric submaximal contraction.

Continuous Measures MnPF and MdPF Results

Mean and median power frequencies were analyzed from the continuous MMG and EMG measures. Frequency data were fitted with linear regression and had r^2 values of 0.46, 0.48, 0.67, 0.54, 0.78, and 0.77 for MMG MnPF, MdPF, EMG Med MnPF, MdPF, EMG Lat MnPF, MdPF, respectively. Typical regression fits are shown in Figure 4.13 (MMG) and Figure 4.14 (EMG). Sustained isometric (M = -4.513%/min, SD = 12.106) and On/Off (M = -1.896%/min, SD = 5.579) conditions showed decreased MMG mean frequency response but were not significantly different from one another, $t(13) = 0.869, p = 0.201, d = 0.303$. MMG MdPF response was similar, with sustained isometric (M = -3.051%/min, SD = 22.746) and On/Off (M = -1.981%/min, SD = 8.935) conditions showing a shift towards lower frequencies but were not statistically different, $t(13) = 0.164, p = 0.437, d = 0.062$. EMG response of the medial head revealed quicker decline in frequency during a sustained isometric contraction (M = -8.405%/min, SD = 7.173) than an intermittent contraction between 0% and 30% (M = -1.987%/min, SD = 7.946), $t(13) = 2.793, p = 0.008, d = 1.056$. A similar response was found using median power frequency where sustained isometric (M = -6.420%/min, SD = 7.946) led to a quicker rate of fatigue than On/Off (M = 0.138%/min, SD = 5.755), $t(13) = 2.149, p = 0.014, d = 0.945$. Mean power frequencies of the lateral triceps head revealed shifts to lower frequencies in both sustained isometric (M = -8.444%/min, SD = 10.762) and On/Off (M = -3.100%/min, SD = 8.596). There were no statistical differences, however, between conditions, $t(13) = 1.670, p = 0.121, d = 0.549$. No differences were also found in median power frequencies between sustained isometric (M = -4.068%/min, SD = 12.925) and On/Off (M = 1.695%/min, SD = 18.058), $t(13) = 0.930, p = 0.371, d = 0.367$.

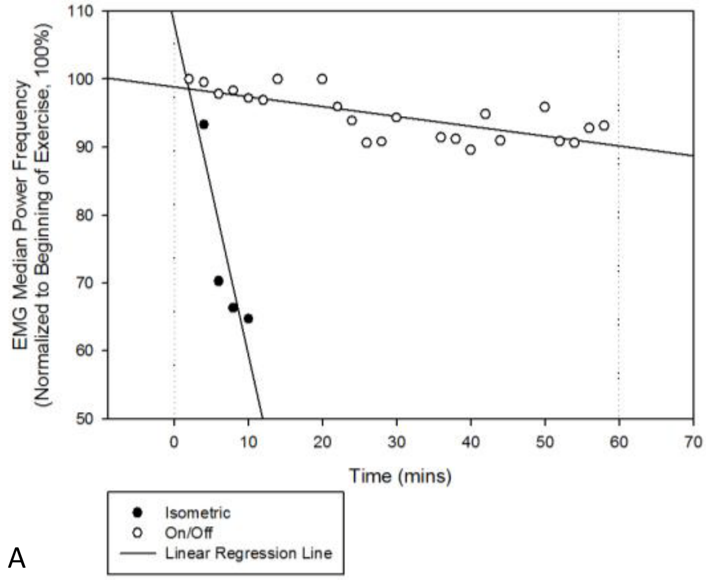


A

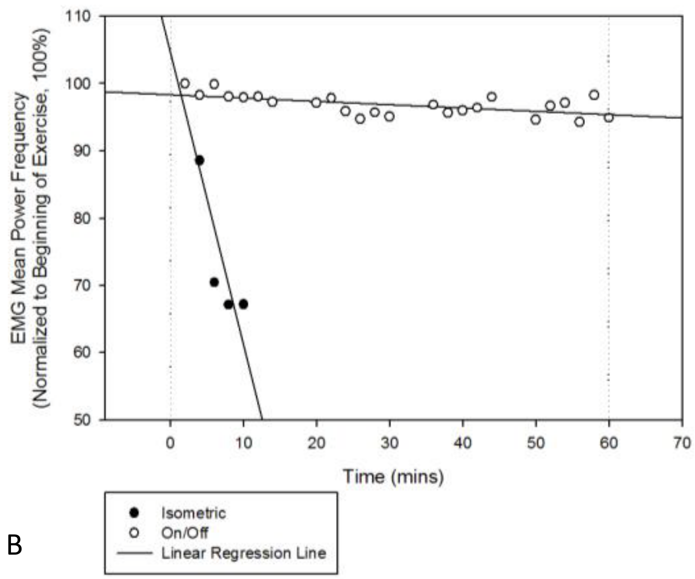


B

Figure 4.13 Typical Continuous MMG MnPF and MdPF Response for Sustained (A) and On/Off (B)



A



B

Figure 4.14 Typical Continuous EMG MnPF and MdPF Response for Medial (A) and Lateral (B) Heads of Triceps Brachii During Exercise

Test Contractions MnPF and MdPF Results

MMG and EMG measures were analyzed in its frequency domain where mean and median power frequency values were calculated during exercise and recovery. Data was fit with logarithmic functions with a mean goodness of fit (r^2) of 0.72 (sustained isometric MMG MnPF), 0.40 (On/Off MMG MnPF), 0.75 (sustained isometric MMG MdPF), 0.39 (On/Off MMG MdPF), 0.87 (sustained isometric EMG Med MnPF), 0.58 (On/Off EMG Med MnPF), 0.81 (sustained isometric EMG Med MdPF), 0.55 (On/Off EMG Med MdPF), 0.87 (sustained isometric EMG Lat MnPF), 0.55 (On/Off EMG Lat MnPF), 0.77 (sustained isometric EMG Lat MdPF), and 0.54 (On/Off EMG Lat MdPF). Figure 4.15 shows typical regression fits for exercise test contractions.

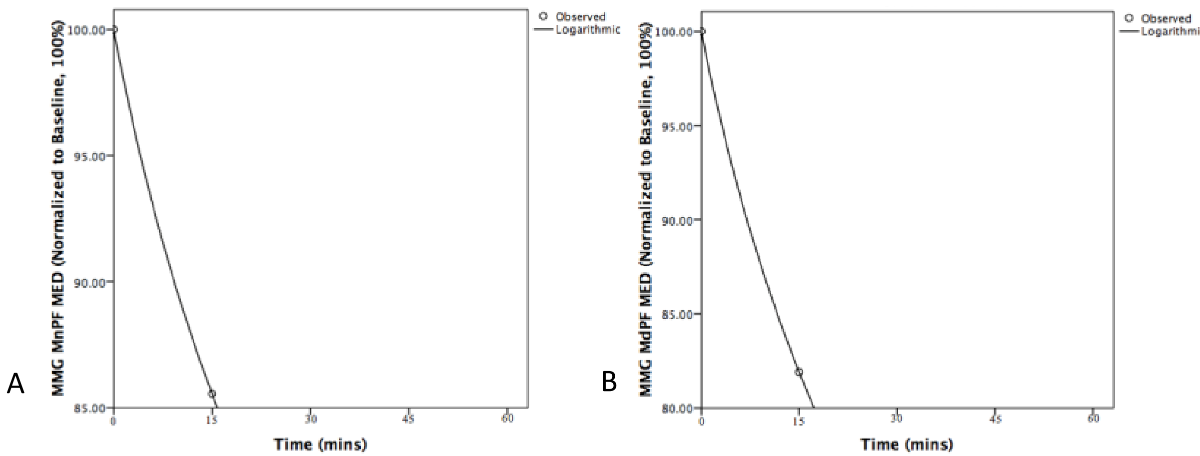


Figure 4.15 Typical Test Contraction MMG and EMG MnPF (A) and MdPF (B) Response for Medial Head

During exercise, the sustained isometric condition (M = -20.664%/Test Battery, SD = 23.653) led to a quicker fatigue response than On/Off (M = -6.699%/Test Battery, SD = 11.370) when considering MMG mean power frequencies. Cessation values (M = 22.610 Hz, SD = 5.542) after a sustained isometric contraction at 15% MVF reveals significant lower MMG MnPF from baseline (M = 26.065 Hz, SD = 6.084), $F(0,6) = 1.665$, $p = 0.201$, $\eta_p^2 = 0.122$ (Figure 4.16). Recovery MMG MnPF values returned towards baseline and were not statistically different from pre-measures. The classical intermittent contraction pattern, on the other hand, led to MMG MnPF shifts to lower frequencies at 15 minutes recovery (M = 22.292 Hz, SD = 5.858) and was statistically lower than baseline (M = 25.077 Hz, SD = 5.242) up to 60 minutes recovery. In contrast to MMG MnPF values, there were no differences in rate of fatigue using median power between sustained isometric (M = -18.215%/Test Battery, SD = 32.951) and On/Off (M = -9.548%/Test Battery, SD = 20.332), $t(13) = 1.0333$, $p = 0.161$, $d = 0.317$ (Figure 4.17).

Median power frequencies after the sustained isometric condition showed no difference at cessation or recovery, $F(0,6) = 1.350$, $p = 0.276$, $\eta_p^2 = 0.101$. The classical intermittent pattern revealed lower MdPF values but was not significantly different at cessation. However, values were statistically lower at 15 minutes (M = 18.328 Hz, SD = 6.839), 60 minutes (M = 18.219 Hz, SD = 6.114) and 24 hours (M = 18.408 Hz, SD = 6.226) when compared to baseline (M = 22.392 Hz, SD = 6.255), $F(0,6) = 1.486$, $p = 0.195$, $\eta_p^2 = 0.110$.

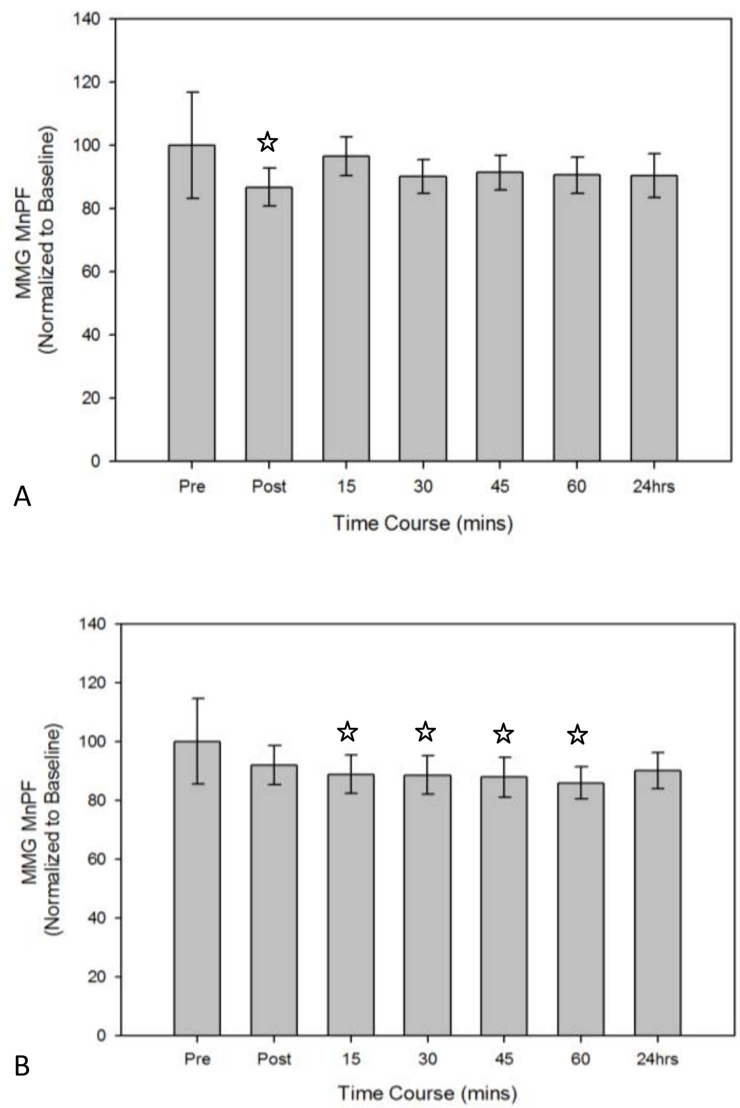


Figure 4.16 MMG MnPF Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B)

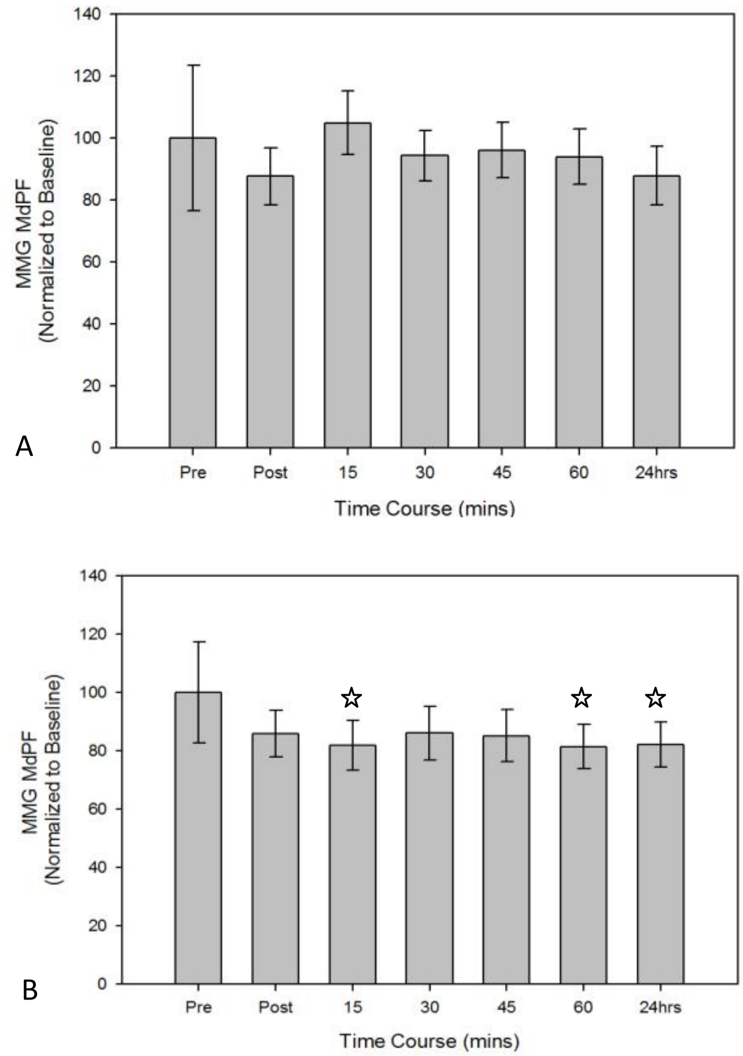


Figure 4.17 MMG MdPF Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B)

Muscle activity from the medial head of the triceps brachii was collected by EMG and analyzed in the power spectrum (Figure 4.18 and Figure 4.19). Both mean and median power frequencies reveal no statistical differences between sustained isometric (MnPF: M = -6.432%/Test Battery, SD = 10.040; MdPF: M = -3.854%/Test Battery, SD = 9.754) and On/Off (MnPF: M = -2.645%/Test Battery, SD = 5.902; MdPF = -2.329%/Test Battery, SD = 4.482), MnPF: $t(13) = 1.142, p = 0.138, d = 0.527$, MdPF: $t(13) = 0.503, p = 0.312, d = 0.201$. Mean power frequency analysis indicate no statistically significant shift to the lower frequency spectra at cessation (M = 97.751 Hz, SD = 18.203) when compared to baseline (M = 103.062 Hz, SD = 17.863) for the medial head after sustained isometric exercise (p = 0.101). Mean power frequency values were reduced but not statistically lower than baseline during cessation and recovery for the On/Off pattern. Median power frequencies demonstrated a similar trend in both sustained isometric and On/Off conditions.

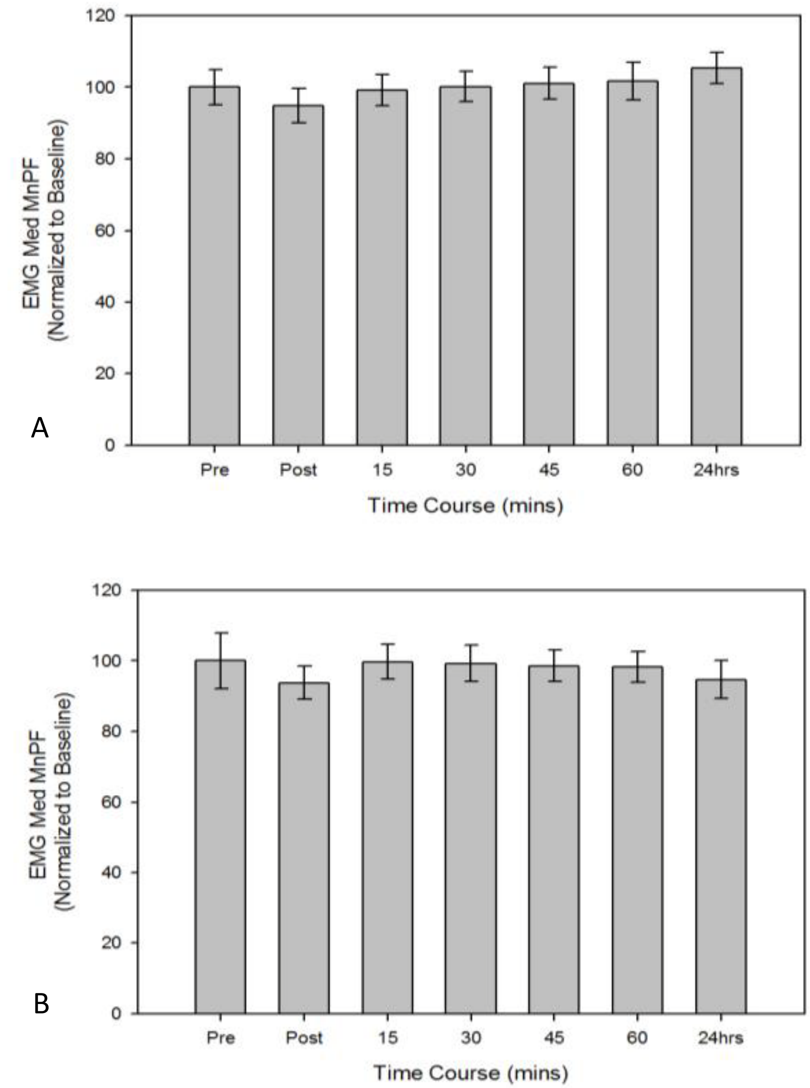


Figure 4.18 EMG MnPF Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B) in Medial Head

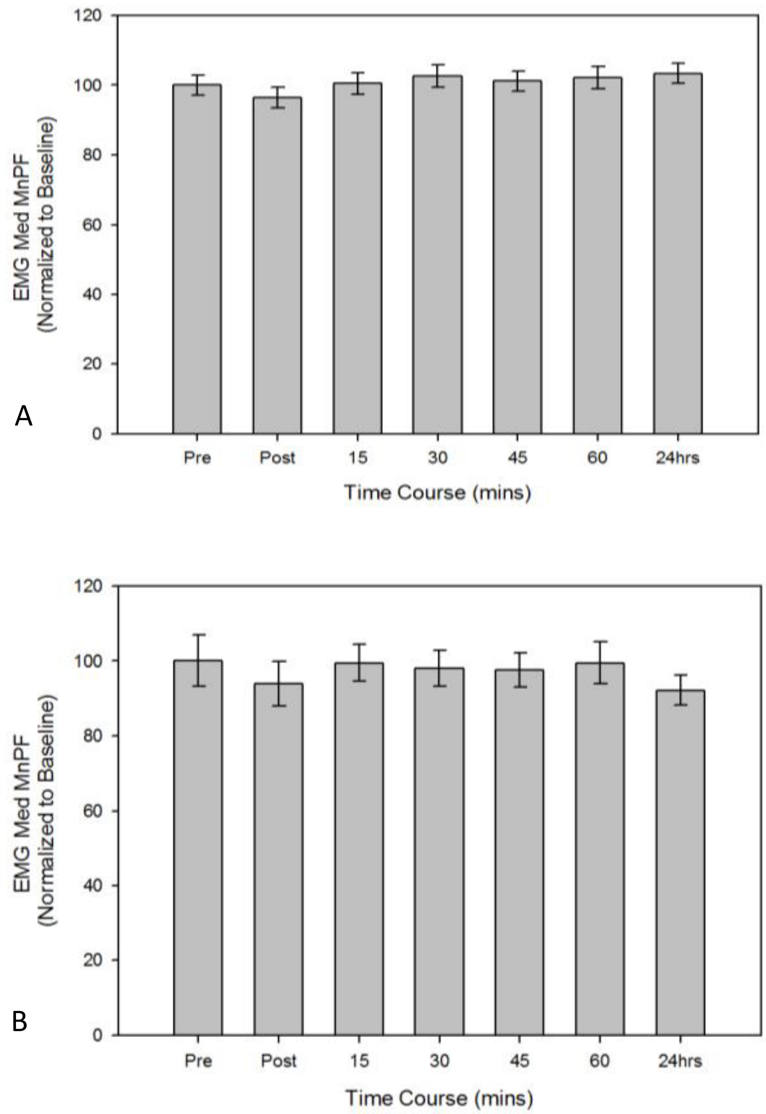


Figure 4.19 EMG MdPF Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B) in Medial Head

The lateral triceps had significant quicker mean power fatigue response in the sustained isometric (M = -14.989%/Test Battery, SD = 12.669) than On/Off (M = -3.347 %/Test Battery, SD = 16.630), *t(13) = 2.335, p = 0.019, d = 0.788*. There were no statistical differences when comparing median power frequency responses between sustained isometric (M = -12.548%/Test Battery, SD = 11.483) and On/Off (M = -5.703%/Test Battery, SD = 12.044), *t(13) = 1.404, p = 0.093, d = 0.582*. When comparing post-exercise measures using mean power frequencies, sustained isometric cessation (M = 115.893 Hz, SD = 19.055) was significantly lower than baseline (M = 131.898 Hz, SD = 27.091), F(0,6) = 4.408, p = 0.003, $\eta_p^2 = 0.269$. On/Off, however, did not lead to statistically significant shifts to lower frequencies at cessation or during recovery, F(0,6) = 0.456, p = 0.676, $\eta_p^2 = 0.037$. Sustained isometric also had depressed values at cessation (M = 89.405 Hz, SD = 8.134) than baseline (M = 98.432 Hz, SD = 12.454) based on mean power frequencies, F(0,6) = 4.264, p = 0.001, $\eta_p^2 = 0.262$. Using median power frequency analysis, there were no significant differences from baseline at cessation and in recovery. Recovery responses are shown in Figure 4.20 and Figure 4.21.

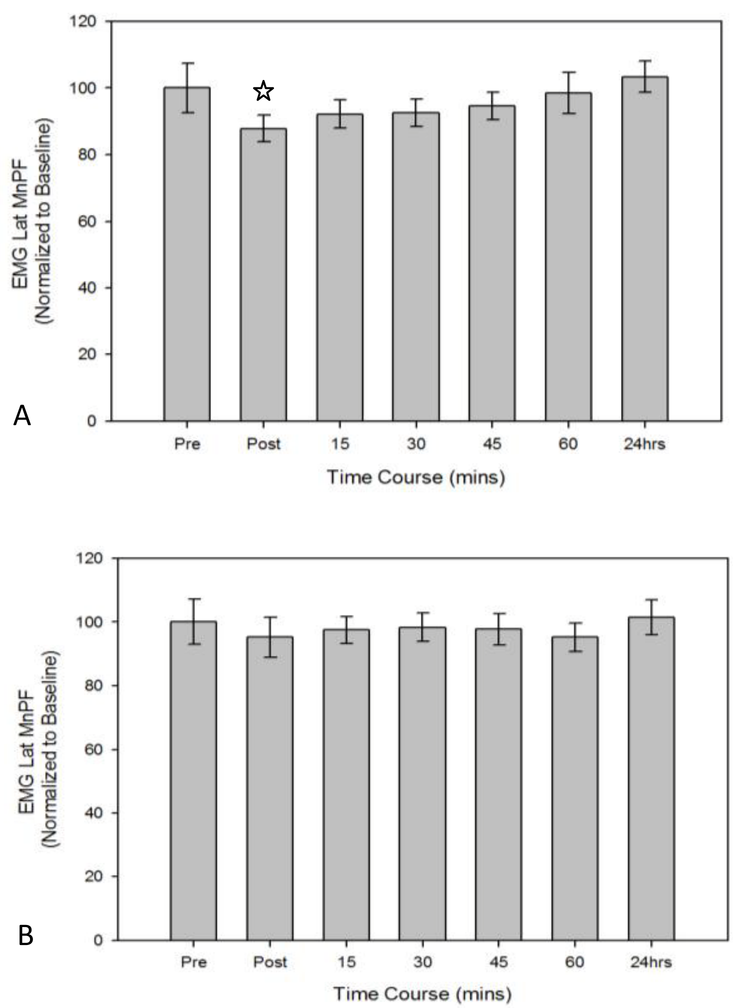


Figure 4.20 EMG MnPF Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B) in Lateral Head

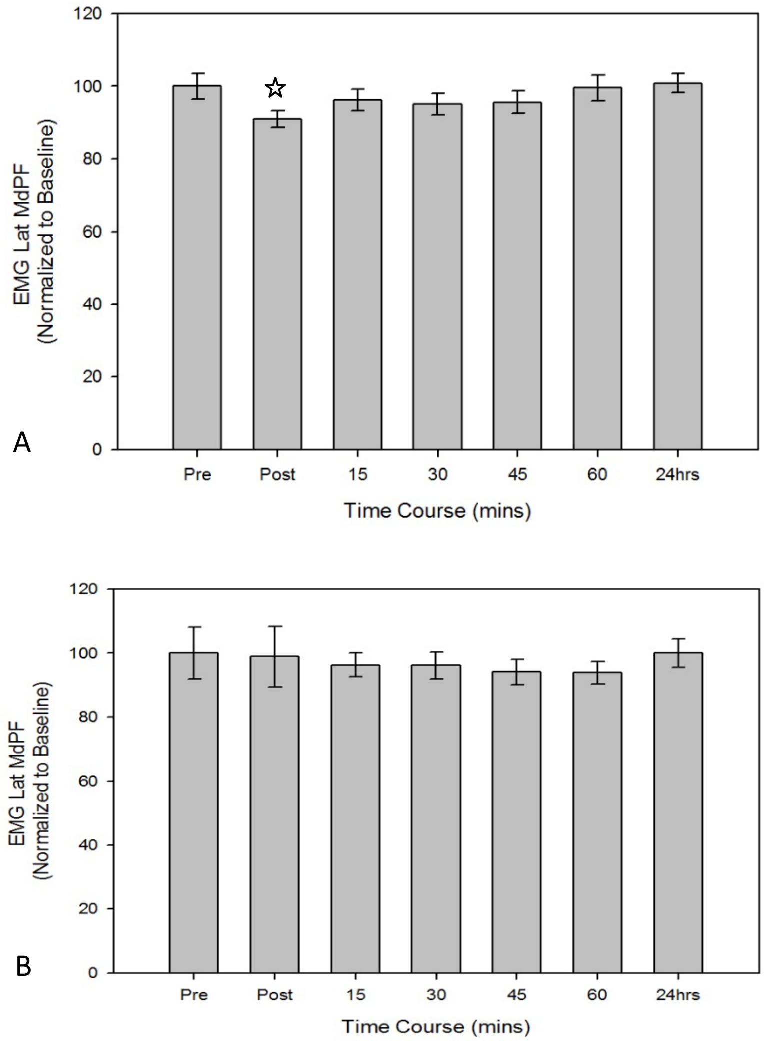


Figure 4.21 EMG MdPF Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B) in Lateral Head

MnPF and MdPF Preliminary Discussion

Analysis of EMG in its power spectrum is often used when studying muscle fatigue. Common statistical measures to characterize the EMG signal in its frequency domain are median power frequency (MdPF) and mean power frequency (MnPF). Median power frequencies are most often calculated (Basmajian & De Luca, 1985) and refer to the frequency of the power spectral density function where half the power lies above and the other half below (Winter, 2005). An alternative measure is mean power frequency (Öberg et al., 1994). During fatigue from either maximum contractions or prolonged exertions, the spectrum shifts towards lower frequencies (Vøllestad, 1997). Previous studies suggest that MMG frequency shifts may represent global firing rate of unfused, activated motor units (Ebersole et al., 2006). EMG frequency shifts, on the other hand, may reflect a decrease in the conduction velocity of active muscle fibres (Vøllestad, 1997; Ebersole et al., 2006) or an increase in motor unit synchronization (Krogh-Lund & Jørgensen, 1992).

Results from this study suggest that both mean and median power frequencies shifted towards lower values when measuring continuous MMG and EMG. The MMG decreases in mean and median power frequencies are consistent with previous research (Weir et al., 2000). The reduction of frequency values may be due to the slowing of the elementary twitch of the motor units summated into MMG (Orizio et al., 2003). It may also be due to "muscle wisdom", as described earlier, with a global reduction of motor unit firing frequency (Weir et al., 2000; Orizio et al., 2003). The sustained isometric effort led to a quicker fatigue response than the intermittent contraction pattern, significantly quicker during test contractions. It has been shown that frequency response is dependent on the intensity of the contraction, as there have been both increases and decreases in frequency over time and its pattern varied by intensity level (Weir et al., 2000). For instance, Orizio and colleagues (2003) found that at 20% MVC sustained isometric effort, the frequency content of MMG did not change. However, at 80% MVC, a transient increase of power followed by a clear shift to lower frequencies was described (Orizio et al., 2003). Since the intermittent contraction pattern led to a lower overall intensity exertion, the decrease in frequency was less marked than the sustained isometric condition. In fact, mean power frequencies at cessation after an On/Off condition was not significantly depressed relative to baseline. However, MnPF recovery values between 15 and 60 minutes were significantly lower than pre-measure. Median power frequencies did not show this relationship. Significant decreased MnPF values were found at 15, 60 minutes and 24 hours post-exercise. As described earlier, MMG may reflect the contractile properties of the muscle, whereas EMG may solely describe the decrease in conduction velocity and increase in synchronization. MMG may thus measure the long-term effects of exercise particularly if greater peripheral fatigue is evident.

EMG-based frequency analysis of the medial head showed that both sustained isometric and On/Off conditions resulted in decreased power spectrum values, consistent with previous research (Hagberg, 1981; Jørgensen et al., 1988; Weir et al., 2000; Ebersole et al., 2006). However the sustained isometric contraction led to a significant shift to lower frequencies at cessation based on mean power frequencies whereas On/Off led to slight decreases in cessation value. Median power analysis revealed decreased, but not significant, frequency at sustained isometric cessation. A quicker rate of fatigue during a sustained isometric contraction was observed when compared to On/Off in both mean and median power frequencies. There were no differences, however, between both conditions based on test contraction MnPF and MdPF. As mentioned earlier, EMG shifts towards lower frequencies may be due to an increase in concentration of extracellular potassium that may subsequently reduce the action potential conduction velocity. Animal experiments have shown decreases in intracellular potassium concentration and contractility of muscle at low-level sustained isometric forces (Jørgensen et al., 1988). On the other hand, intermittent contractions may allow the restoration of contraction-related fluxes of ions during periods of rest. These periods of rest may provide sufficient blood supply for the entire muscle (Jørgensen et al., 1988). An alternative theory is that varying levels of sustained isometric exertions may lead to the recruitment of fast-twitch motor units with increased action potential conduction velocities. This, in turn, will result in increases in both mean and median power frequencies (Perry et al., 2001). Recovery analysis in the sustained isometric condition revealed slight mean and median power frequency shifts that were not significantly different from baseline. These results agree with Zwarts and colleagues (1987) who found that the power spectrum returns close to baseline values within two minutes into recovery. The On/Off condition in both mean and median power frequencies resulted in a slight decrease in frequency value but was not significantly different from baseline. Recovery values, too, were within the baseline frequency spectrum. This result has been shown in past literature. In a study conducted by Moxham and colleagues (1982), after repeated submaximal contractions, the power spectrum remained unchanged during exercise and recovery. Similarly Yassierli and Nussbaum (2008) observed fluctuating linear changes and increasing linear changes during low level efforts. It was suggested that both mean and median power frequencies are oversimplified indicators that are insensitive to fatigue attributed to low-level force. The complexities of low-level force include combinations of decreasing firing rate, motor unit rotation, de-recruitment of motor units, and recruitment of larger motor units (Yassierli & Nussbaum, 2008).

The lateral head showed decreased power spectrum values during exercise. There were no significant differences between sustained isometric and On/Off exercise conditions during continuous measurement of mean and median power frequencies. However test contraction slopes reveal a quicker rate of fatigue response during the sustained isometric condition using mean power frequency analysis. Similar to the medial head, sustained isometric led to lower mean and median power frequency values at cessation and On/Off led to slight, but not significant, shifts to lower frequencies. The similar findings between two heads of the same muscle may provide additional support to the mechanisms described earlier for sustained isometric and intermittent conditions.

A comparison between mean and median power frequencies suggests that mean values may have slightly higher fatigue sensitivity than median. This postulation is supported by Yassierli and Nussbaum (2008) who suggests that fatigue-associated power spectrum density shifts are accompanied by changes in its shape. However, results from this study disputes Yassierli and Nussbaum's (2008) finding that MdPF had higher variability.

EMG Hi-Lo Results

Using the continuous EMG measures, Hi-Lo ratios were analyzed to identify changes in the EMG power spectrum. This Hi-Lo ratio compares the power in the low (20 – 40 Hz) and high (130 – 238 Hz) frequency bands (Bigland-Ritchie et al., 1981; Moxham et al., 1982). Ratios were plotted against time (every 2 minutes) and fitted with logarithmic regression with a goodness of fit (r^2) of 0.61 and 0.45 for sustained isometric and On/Off conditions, respectively for the medial head. Goodness of fit values for the lateral head were $r^2 = 0.70$ for the sustained isometric condition and $r^2 = 0.37$ for On/Off. Typical regression fits for both medial and lateral heads are shown in Figure 4.22.

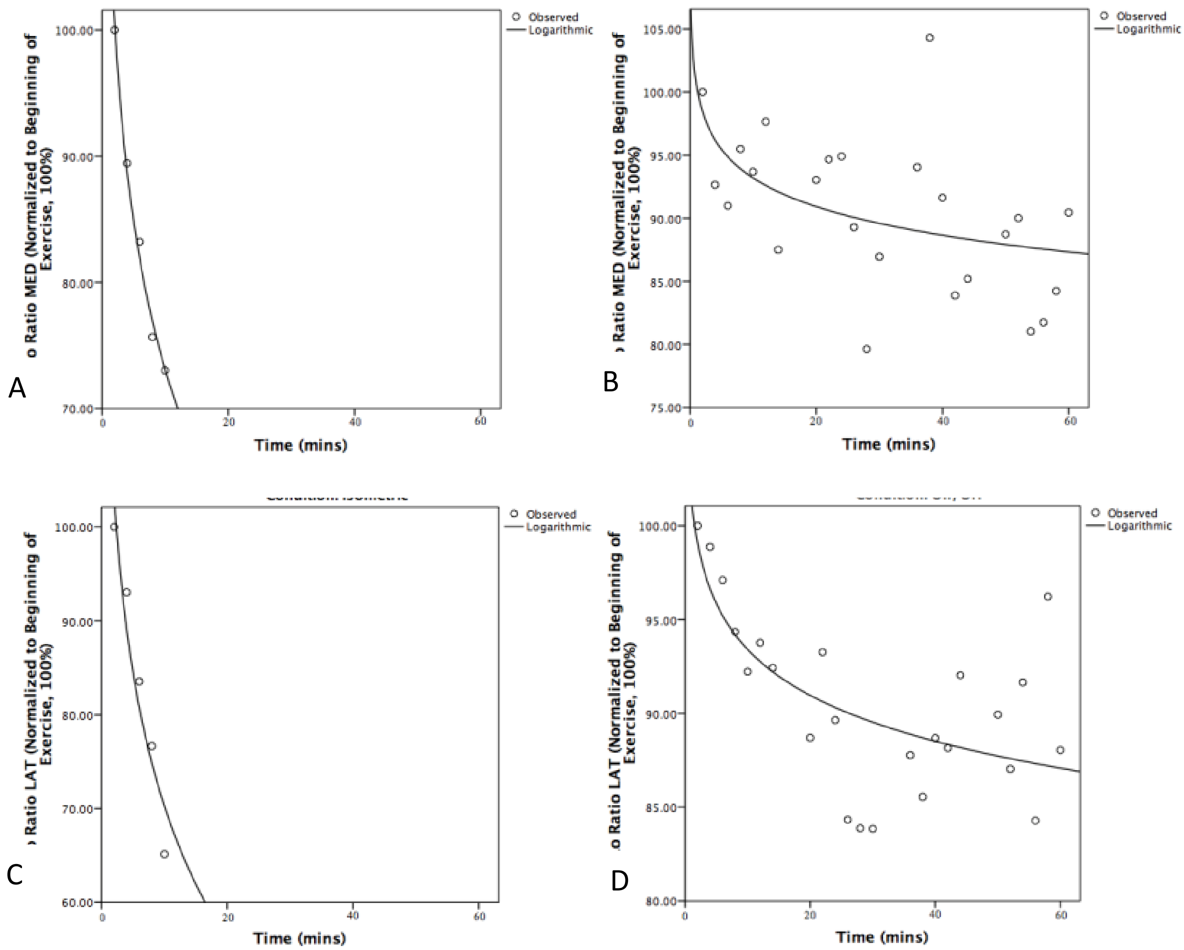


Figure 4.22 Typical Continuous EMG Hi-Lo Ratio Response for Sustained (A) and On/Off (B) in Medial Head and Sustained (C) and On/Off (D) in Lateral Heads of Triceps Brachii

Based on Hi-Lo ratios of the medial head, sustained isometric (M = -15.108%/min, SD = 15.555) effort at 15% MVF led to a quicker response over time than the intermittent pattern (M = -2.400%/min, SD = 10.253), $t(13) = 3.430, p = 0.003, d = 0.965$. The lateral head showed similar trends with a quicker declining rate of response in the sustained isometric (M = -13.067%/min, SD = 18.033) than the On/Off (M = -0.311%/min, SD = 13.793), $t(13) = 2.376, p = 0.018, d = 0.795$.

EMG Hi-Lo Preliminary Discussion

An alternative method to measure and characterize the changes within the EMG power spectrum is with power ratios between different frequency bands. The use of such ratios may be useful when the limitations of Fourier transformation (FFT) result in challenges in interpretation. For instance, a signal that is nonstationary, often seen in non-isometric muscle activities, violates the FFT assumption. Similarly, the movement of electrodes relative to muscles may pose difficulties in using FFT (Allison & Fujiwara, 2002). There have been arguments against the use of the median power frequency (MdPF) as an assessment of central tendency in the power spectrum. It has been suggested that changes in MdPF are associated with high frequency fatigue and less sensitive to low frequency fatigue (Allison & Fujiwara, 2002).

In the present study, both sustained isometric and On/Off conditions led to a decrease in Hi-Lo ratio, results consistent with observations reported by Bigland-Ritchie and colleagues (1981), Moxham and colleagues (1982), and Allison & Fujiwara (2002). Results also suggest a greater decline in Hi-Lo ratio during a sustained isometric contraction in both medial and lateral triceps heads. The lack of ratio reduction in the intermittent condition (medial: M = -2.400%/min, lateral: M = -0.311%/min) may be due to substantial and persistent low-frequency fatigue (Moxham et al., 1982). Moxham and colleagues (1982) observed that after repeated submaximal contractions ratios were either normal or slightly raised. The sustained isometric condition, on the other hand, resulted in progressively decreased EMG Hi-Lo ratios in both medial and lateral heads. Dolan and colleagues (1995) conducted a banding analysis of EMG to determine shifts in the power spectrum in both high and low frequency components. It was found that the most consistent changes with fatigue at all loads and duration times were within the range of 5 and 30 Hz and was a good predictor of endurance time. The lower frequency band chosen in this study was between 20 and 40 Hz to eliminate all possible movement artifact frequencies (Bigland-Ritchie et al., 1981) but resulted in comparable increase in power within lower frequencies. Beyond a reduction in conduction velocity, the large spectral shift may be attributed to the disturbance of the normal random distribution of motor neuron activity (Bigland-Ritchie et al., 1981). The synchronization of motor neuron firing may result in large, low frequency EMG oscillations. It is important to note the criticisms of the use of Hi-Lo ratios to characterize power spectrum changes. According to Hägg (1992), Hi-Lo ratios is not related to any specific fatigue phenomenon by any known model, demonstrates a poor relationship with action potential velocity decrease, and its limits for high and low frequency bands are arbitrarily set without standardization.

Low-Frequency Fatigue Results

Muscle response at stimulations of 20 Hz and 100 Hz were measured at 15-minute intervals during exercise and 15-minute intervals during recovery. A post-exercise response was also measured 24 hours later. Response during exercise was fitted with linear regression and had best-fit values of $r^2 = 0.85$ and $r^2 = 0.61$ for sustained isometric and intermittent contractions, respectively. Typical low frequency fatigue (LFF) ratio curves are shown in Figure 4.23.

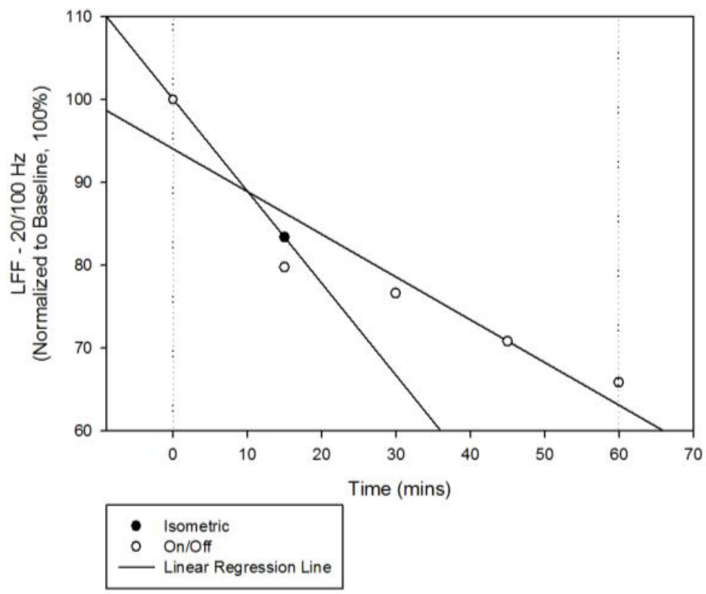
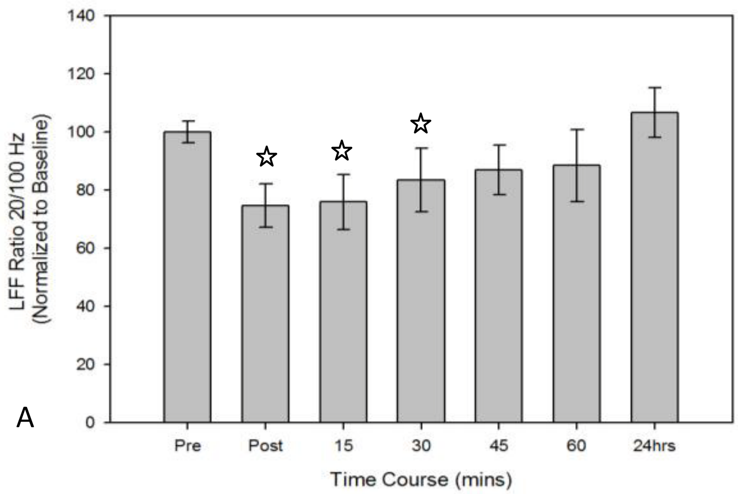


Figure 4.23 Typical Low-Frequency Fatigue Ratio Response During Exercise

Based on LFF ratios during exercise, sustained isometric (M = -18.274%/Test Battery, SD = 15.885) resulted in a quicker decreasing rate of response than the intermittent contraction (M = -4.556%/Test Battery, SD = 10.250), *t(13) = 3.945, p = 0.001, d = 1.026.*

Recovery intervals at 15, 30, 45, 60 minutes and 24 hours post-exercise were compared to baseline (Figure 4.24). After a sustained isometric effort at 15% MVF, LFF ratios remained significantly lower than baseline values (M = 0.626, SD = 0.136) until 45 minutes (M = 0.545, SD = 0.192) into recovery (p=0.085), F(0,6) = 7.213, p = 0.001, $\eta_p^2 = 0.375$. The intermittent contraction, on the other hand, did not lead to a significantly lower LFF ratio (p = 0.523) at cessation (M = 0.569, SD = 0.150) when compared to baseline

(M = 0.625, SD = 0.161), F(0,6) = 2.993, $p = 0.011$, $\eta_p^2 = 0.200$. LFF ratio values, however, were significantly lower than baseline until 30 minutes into recovery (M = 0.488, SD = 0.169), $p = 0.004$.



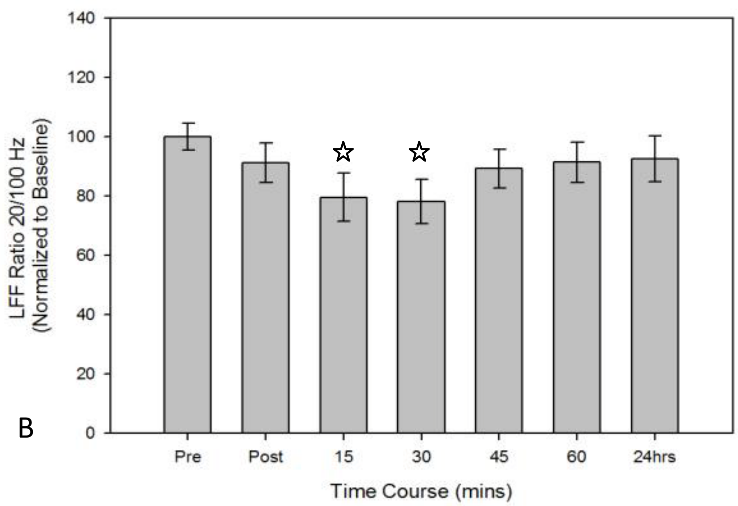


Figure 4.24 LFF Ratio Comparisons Between Baseline & Recovery for Sustained (A) and On/Off (B)

Low-Frequency Fatigue Preliminary Discussion

A decrease in low-frequency fatigue ratio was observed in both sustained isometric and On/Off conditions, in agreement with findings from Bystrom and Fransson-Hall (1994), Vøllestad et al. (1997), and Griffin and Anderson (2008). The rate of LFF ratio decrement was quicker during a sustained isometric effort and led to a significantly lower LFF value at cessation. It has been theorized that LFF can be divided into at least two phases: the impairment of cross-bridge function from metabolic accumulation and the impairment of calcium handling (Griffin & Anderson, 2008). This second phase may be a long-lasting response, which is not dependent on metabolite levels (Binder-Macleod & Russ, 1999). The sustained isometric condition may have reached the second phase (calcium handling impairment). Because the disruption of calcium handling was reflected in the steepest portion of the calcium-force curve (Chin & Allen, 1996), forces in the low frequency range were lower. In contrast, the intermittent contraction may have only involved the impairment of cross-bridge function and did not affect the calcium-force curve during exercise. As a result, significant reduced force at low frequencies was not evident at the conclusion of the intermittent contraction. This contradicts the theory proposed by Binder-Macleod and Russ (1999) who suggested that intermittent contractions might result in an increase in internal work, which in turn would produce more metabolites to impair cross-bridge function.

LFF values remained significantly depressed after sustained isometric effort, relative to baseline, up to 45 minutes into recovery. The intermittent contraction, in turn, led to decreased LFF values up to 30 minutes into recovery. These findings are similar to those observed by Bystom and Fransson-Hall (1994) who found a 15% - 20% decrement in force response at 20 Hz after one hour into recovery after sustained isometric handgrip exercises. Interestingly, LFF recovery values after intermittent contractions were significantly lower during the first 30 minutes of recovery. These trends are similar to findings observed by Saugen and colleagues (1997) who found the lowest value after 10 minutes recovery and a subsequent gradual increase for the next 20 minutes. A similar phenomenon has been reported after electrically evoked fatigue with lowest LFF values between 2 and 13 minutes of recovery (Binder-Macleod & Russ, 1999). Binder-Macleod and Russ (1999) suggest that the impaired calcium-handling phase may have a long onset time and may explain significant decreases of LFF during initial recovery. The prolonged decreases in LFF value during recovery may reflect the sustained effect of the Ca^{2+} dependent phase even though the metabolic phase may be fully recovered (Binder-Macleod & Russ, 1999). The results from this study revealed full recovery of LFF 24 hours post-exercise, contradicting previous trends (Bystrom & Fransson-Hall, 1994). In that study, LFF was present 24 hours after sustained exercise at 25% MVC and after an intermittent protocol at 40% MVC (5 + 6.8 seconds). Most likely the intensity of both the sustained and intermittent contractions in the Bystrom and Fransson-Hall (1994) study led to the prolonged LFF response that was not observed in this study.

Blood Flow Velocity Results

Triceps blood flow was measured directly from mean blood velocity passing through the brachial artery. As described earlier, diameter measurements of the brachial artery were consistent over time for every individual and in each condition. Mean blood flow was measured every 2 minutes for a 30 second window. A typical blood flow velocity response is shown in Figure 4.25. The mean of the measures from the beginning of exercise (at the 2-minute interval) was calculated for each condition and compared (Figure 4.26). The sustained isometric (M = 87.551%, SD = 82.812) condition led to a significantly lower mean blood flow velocity than On/Off (M = 189.87%, SD = 155.440), t(13) = *8.440, p = 0.000, d = 0.380.*

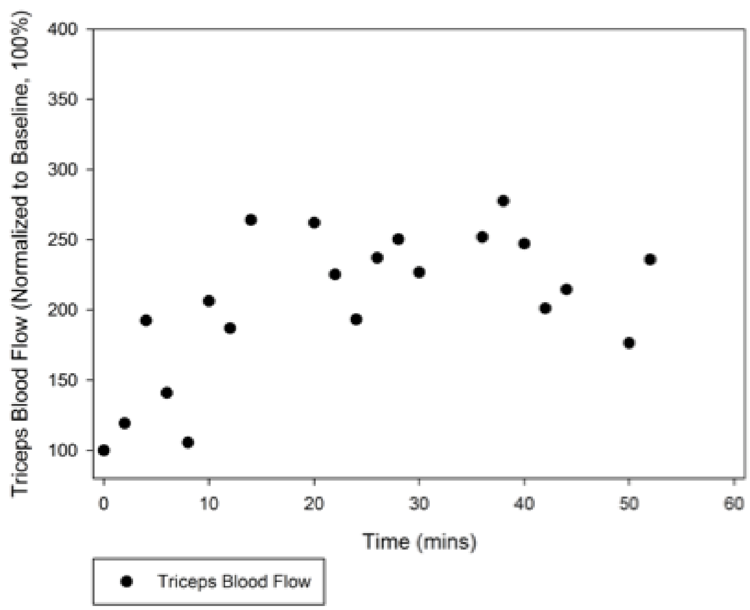


Figure 4.25 Typical Continuous Blood Flow Velocity Response During An Intermittent Contraction

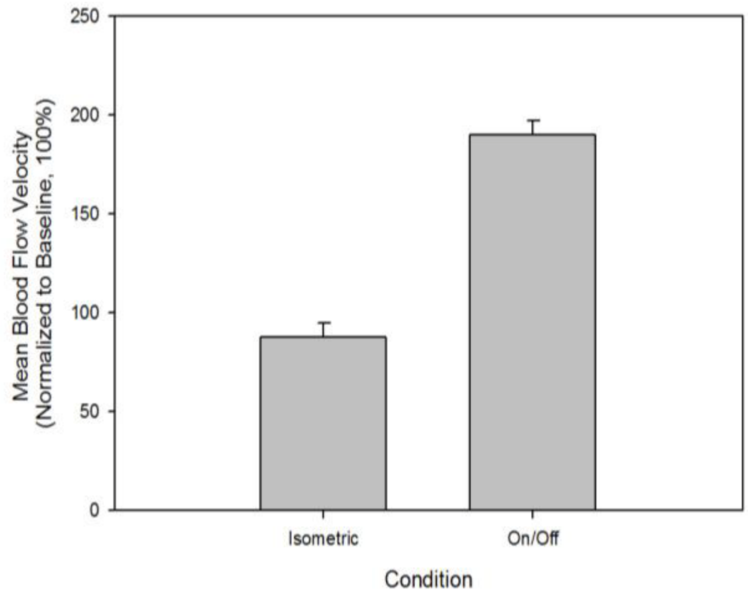


Figure 4.26 Mean Triceps Blood Flow Velocity for Sustained Isometric and On/Off Conditions

Blood Flow Velocity Preliminary Discussion

According to Bystrom and Fransson-Hall (1994), blood flow is a critical factor, even at low-level forces, that contributes to the development of muscle fatigue. It has been suggested that the restriction in blood circulation may disturb muscle homeostasis, leading to the development to musculoskeletal disorders (Galen et al., 2002; Visser & van Dieen, 2006). In this study, during sustained isometric exercise, blood velocity initially increased and quickly decreased in the last few minutes before exhaustion. The intermittent contraction, on the other hand, led to steady increases in blood flow during the protocol, with a plateau towards the end of exercise (typically 60 minutes). The intermittent condition, as a result, led to a quicker rate of blood flow while sustained isometric led to occlusion and a lack of blood supply. Sustained isometric led to a significantly lower mean blood flow velocity compared to the On/Off condition.

During the sustained isometric effort at 15% MVC, muscle contractions may have compressed the vessels supplying the muscle. It has also been suggested that an increase in intramuscular pressure may impede microcirculation, reducing the blood supply to the targeted muscle (Yoshitake et al., 2001; Visser & van Dieen, 2006). Impaired microcirulation may lead to muscle fibres with signs of mitochondrial disturbance (i.e. "ragged red fibres") (Larsson et al., 1999). Previous studies have also shown an accumulation of muscle metabolites and glycogen depletion (Krogh-Lund & Jørgensen, 1992). According to Hultman and Söderlund (1988), muscle glycogen concentration decreased by 17% after 10 minutes of a sustained isometric static contraction at 15% MVC. During an intermittent contraction at 15% MVC, glycogen depletion was assumed to occur within 2 hours.

Conversely previous research has implied a greater intramuscular pressure during intermittent contractions resulting in a larger reduction in blood flow and muscle tissue oxygenation (Vedsted et al., 2006). This study, however, demonstrated greater blood flow during intermittent contractions. Zhang et al., (2004) found that non-static contractions increased blood flow by rhythmically emptying the veins and facilitating the perfusion of the muscle. These intervening periods may facilitate blood flow removal of contraction-inhibiting metabolites (Hagberg, 1981). It has been speculated that rest periods as short as 2 seconds may enhance endurance time (Hagberg, 1981). Local regulation may also play a role in increased blood flow. For instance, due to increased metabolic activity of the exercising triceps muscle, a dilation of the supplying artery was observed. Another mechanism may be an increase in cardiac output as a response to exercise. In the intermittent contraction, it is possible that during the relaxation phase, intramuscular pressure would have dropped so that hyperaemia can be elicited for a brief period of time.

Ratings of Perceived Exertion Results

RPE measures were collected at 2-minute intervals during exercise (Figure 4.27). Sustained isometric (M = 18.237%/min, SD = 12.873) condition led to a quicker rate of psychophysical response than the On/Off (M = 5.287%/min, SD = 6.195) condition, $t(13) = 4.731$, $p = 0.000$, $d = 1.282$. The sustained isometric condition had a mean r^2 value of 0.80 while the On/Off condition was fitted with a mean r^2 of 0.70.

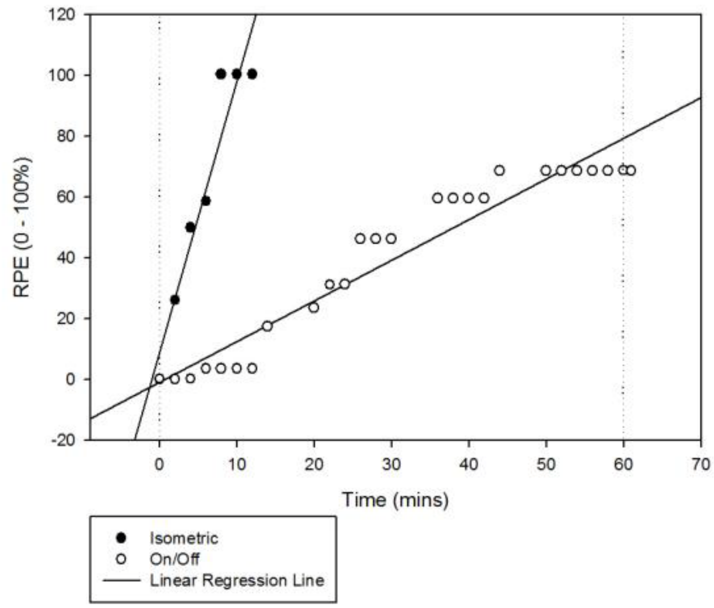


Figure 4.27 Typical Continuous RPE Response During Exercise

Ratings of Perceived Exertion Preliminary Discussion

In this study, participants reported a greater perceived exertion over a shorter period of time while exerting sustained isometric forces than during an intermittent contraction protocol. According to Jones and Killian (2000), perceived exertion is a significant factor that limits exercise performance. This may imply that an elevated perceived exertion might relate to an impaired endurance capacity. Certainly, this was observed in this study, where a quicker rate of perceived exertion was associated with a shorter endurance time. It has been speculated that an increase in RPE are due to increased metabolic demand sensed by the brain via feedback and is related to heart rate, oxygen consumption, blood lactate concentration (Fontes et al., 2010) and other measures of fatigue (Iridiastadi & Nussbaum, 2006). For instance, RPE has also been related to EMG amplitude (Iridiastadi & Nussbaum, 2006; Fontes et al., 2010), mean power frequency at lower level force (Hummel et al., 2005; Iridiastadi & Nussbaum, 2006), and median power frequency (Iridiastadi & Nussbaum, 2006).

Previous studies have shown an increase in perceived exertion as a power function of duration. Other studies have found a linear relationship during a sustained isometric contraction (Krogh-Lund & Jørgensen, 1992; Enoka & Stuart, 1992). Recent studies have applied a third-order polynomial fit to reflect the perceived rating growth (Pincivero et al., 2004) and are recommended for repeated single-joint exercise (Springer & Pincivero, 2010). The results from this study suggest a linear relationship between perceived exertion and time during a sustained isometric condition, similar to earlier studies. It was observed that the intermittent contraction, on the other hand, revealed a sigmoidal trend, which is similar to modeling studies. It is possible that the neuronal networks, between the excitatory drive to a motor neuron pool and the perceived effort, contribute to this nonlinear transformation relationship (Enoka & Stuart, 1992). Interestingly, both sustained isometric and intermittent contraction patterns approached a plateau towards the end of exercise, a phenomenon observed by Springer & Pincivero (2010). A given stimulus towards the end of exercise (higher perceived exerted force) may be less easy to detect and the slope declines (Springer & Pincivero, 2010). Despite the observed trends, to maximize the goodness of fit for both sustained isometric and On/Off conditions, linear regression methods were used to model both responses.

Heart Rate Results

As with EMG, MMG, and blood velocity, heart rate measures were collected continuously for the entire duration of the exercise protocol and analyzed at 2-minute intervals. Typical logarithmic regression fit is shown in Figure 4.28. The goodness of fit (r^2) was 0.57 and 0.48 for sustained isometric and On/Off, respectively. There was a significant statistical difference between sustained isometric (M = 19.698%/min, SD = 14.791) and On/Off (M = 9.457%/min, SD = 8.497) conditions, where the latter had a slower heart rate response, $t(13) = 4.206, p = 0.001, d = 0.849$.

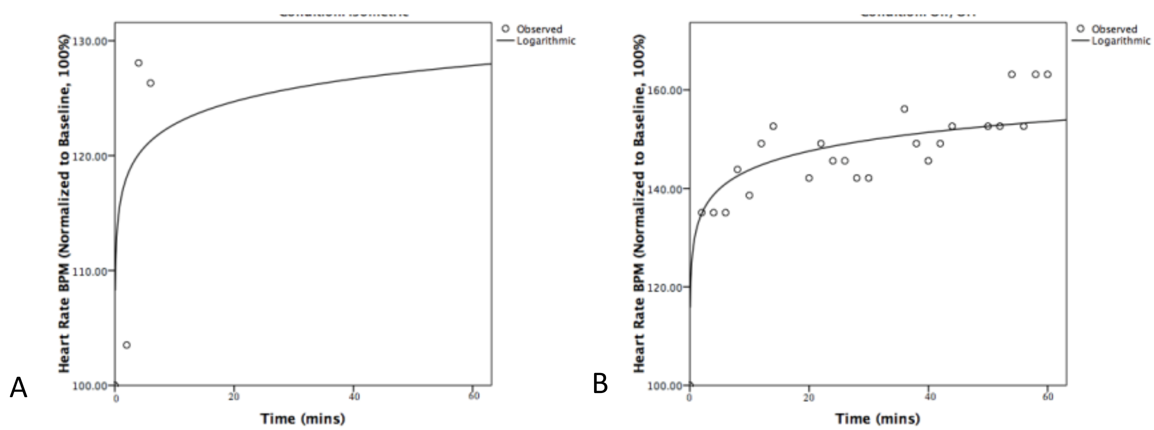


Figure 4.28 Typical Continuous Heart Rate Response During Exercise for Sustained (A) and On/Off (B)

Heart Rate Preliminary Discussion

Heart rate is a reliable measure for predicting maximum work capacity (MacKinnon, 1999). It has been suggested that heart rate, along other central factors such as oxygen uptake and ventilation, may be "amplifiers" that potentiate local factors. These local factors include muscle force and rate of force production (MacKinnon, 1999). A quicker rate of heart rate response was observed during a sustained isometric test when compared to the On/Off task. This result is similar to that found by Bystrom and Fransson-Hall (1994) who found significant increases in heart rate after an isometric effort at 25% MVC and intermittent contraction at 25% and 40%. In a study conducted by Fallentin and colleagues (1985), an isometric 7% MVC elbow extension was sustained for 1 hour. A very modest increase in heart rate was observed (63 +/- 6 to 66 +/- 6 beats per minute). Heart rate response also increased during an intermittent protocol consisting of 50% MVC, 10-second cycle time, 60%-40% duty cycle (Hunter et al., 2004). Helene Garde and colleagues (2003), on the other hand, did not find differences for heart rate during a repetitive intermittent task (10% MVC, 10 second cycle time, 50% duty cycle). The increased heart rate in this study may be attributed to the magnitude of the mean MVF (15%), the type II fibre dominancy, the duty cycle and cycle time.

Even during low-level efforts, there is modest activity from the central command. It is the 'irradiation' of this central command that elicits a heart rate response during exercise (Fallentin et al., 1985). If heart rate is modestly correlated to RPE (MacKinnon, 1999) and was initially used to validate RPE (Borg and

Linderholm, 1970), an increase in RPE observed in this study should correspond to an increase in heart rate. If EMG amplitude reflects the "central command" (Fallentin et al., 1985), an increase in EMG RMS should also correspond to an increase in heart rate. Therefore it is not surprising that heart rate increased significantly during a sustained isometric contraction and marginally increased during an intermittent regime.

EMG Gaps and Mechanical Force Results

The total number of EMG gaps (gaps/minute) of the triceps brachii lateral head were measured for both sustained isometric (Median = 0 gaps/min, 25^{th} = 0, 75^{th} = 0) and On/Off (Median = 8.696 gaps/min, 25^{th} = 0.156, 75^{th} = 21.600) conditions. On/Off condition led to a greater total number of EMG gaps per minute, z = 2.824, p = 0.005. Similarly, EMG gaps analysis of the triceps brachii medial head revealed fewer number of gaps in the sustained isometric (Median = 0 gaps/min, 25^{th} = 0, 75^{th} = 0.09) contraction than On/Off (Median = 8.717 gaps/min, 25^{th} = 0.05, 75^{th} = 25.117), z = 2.667, p = 0.008. Of these EMG gaps, both mean and median duration (seconds) were calculated. On/Off (Median = 0.350 seconds, 25^{th} = 0.222, 75^{th} = 0.574) had lateral head EMG gaps of longer mean duration than the sustained isometric contraction (Median = 0 seconds, 25^{th} = 0, 75^{th} = 0), z = 2.746, p = 0.006. Calculation of mean EMG gap duration of the medial head led to similar results. A longer mean EMG gap duration was observed in the On/Off contraction (Median = 0.290 seconds, 25^{th} = 0.218, 75^{th} = 0.461) than the sustained isometric contraction (Median = 0 seconds, 25^{th} = 0, 75^{th} = 0.144), z = 2.845, p = 0.004. Median EMG gap duration of the lateral head for the On/Off contraction (Median = 0.326 seconds, 25^{th} = 0.222, 75^{th} = 0.369) was longer than the sustained isometric contraction (Median = 0 seconds, 25^{th} = 0, 75^{th} = 0), z = 2.667, p = 0.008. An analysis of the median EMG gap duration for the medial triceps head revealed longer duration in the On/Off condition (Median = 0.262 seconds, 25^{th} = 0.218, 75^{th} = 0.360) than the sustained isometric condition (Median = 0 seconds, 25^{th} = 0, 75^{th} = 0.119), z = 2.756, p = 0.006.

Figure 4.29 shows the data analysis windows used to calculate mechanical force output in the intermittent condition. Windows of 2 seconds were used at both "on" (30% MVF) and "off" (0% MVF) levels. Figure

4.29 also presents the normalized EMG used to calculate EMG gaps. The bottom panel in Figure 4.30 is a detailed view of the EMG of both lateral and medial heads at the 1% MVC threshold.

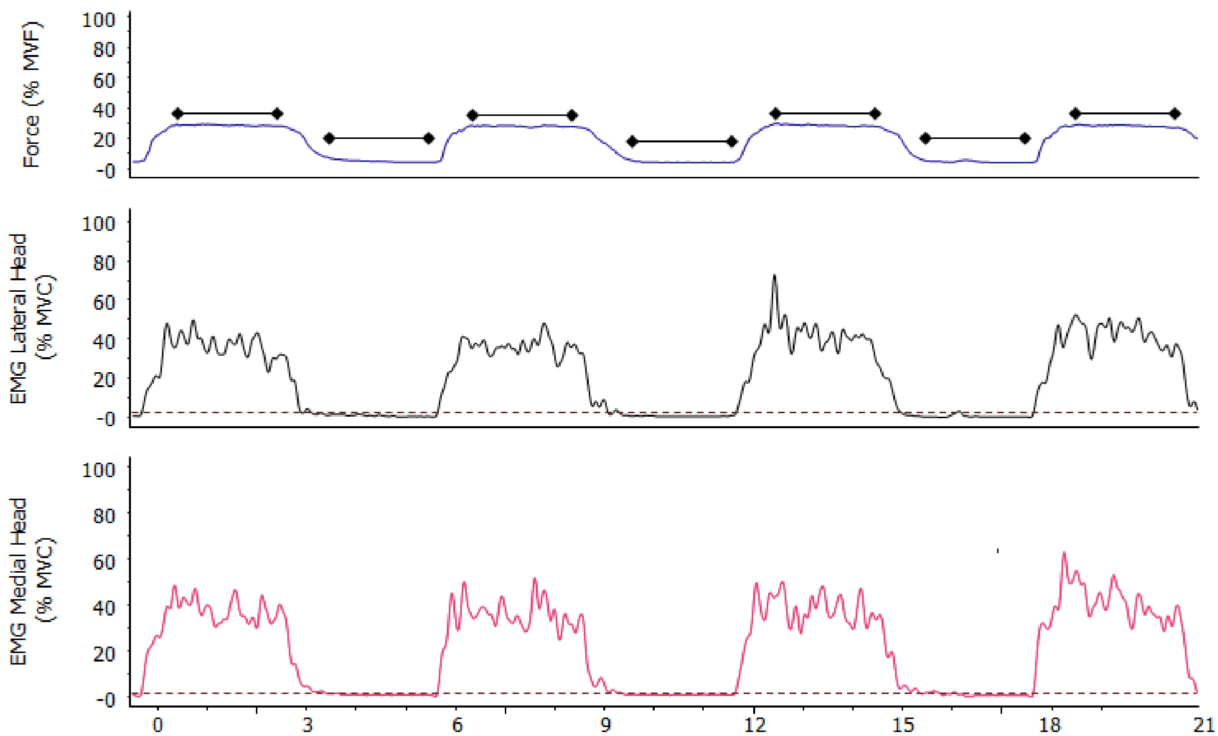


Figure 4.29 Mechanical Force Output and EMG Gaps in Intermittent Contraction On/Off
Mechanical force windows of 2 seconds (double-ended dot bars) at top ("30%") and bottom ("0%") levels of force with normalized EMG of lateral and medial heads of triceps.

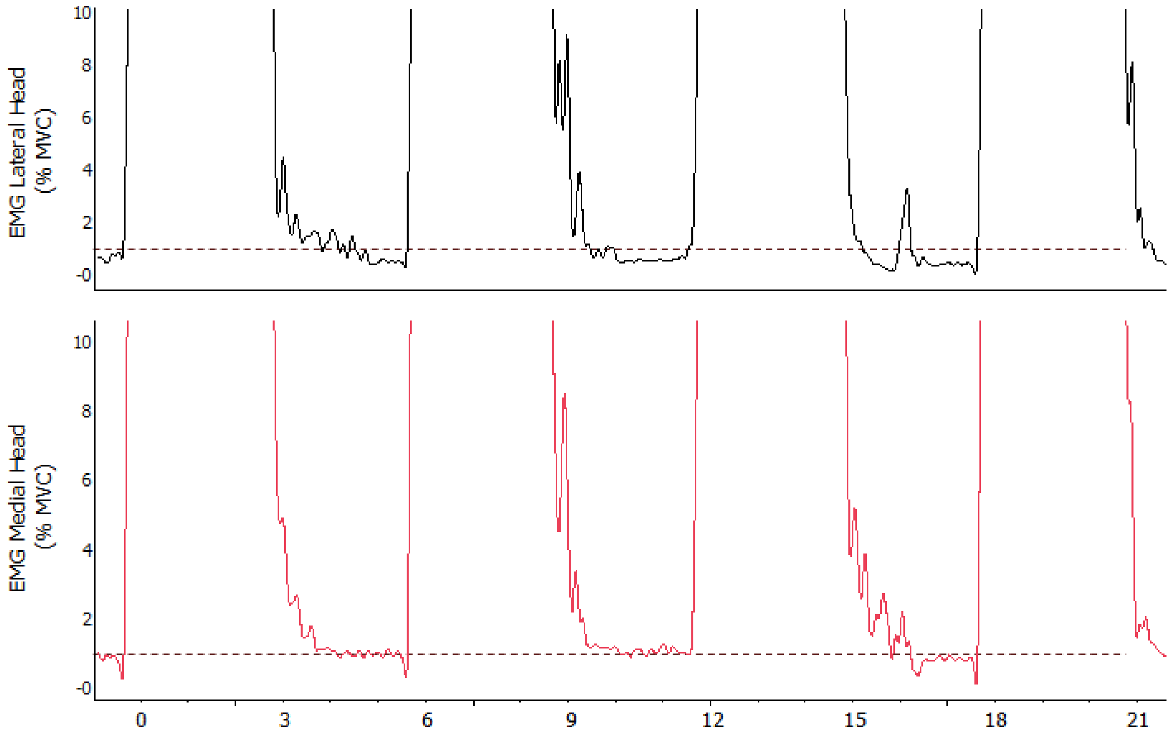
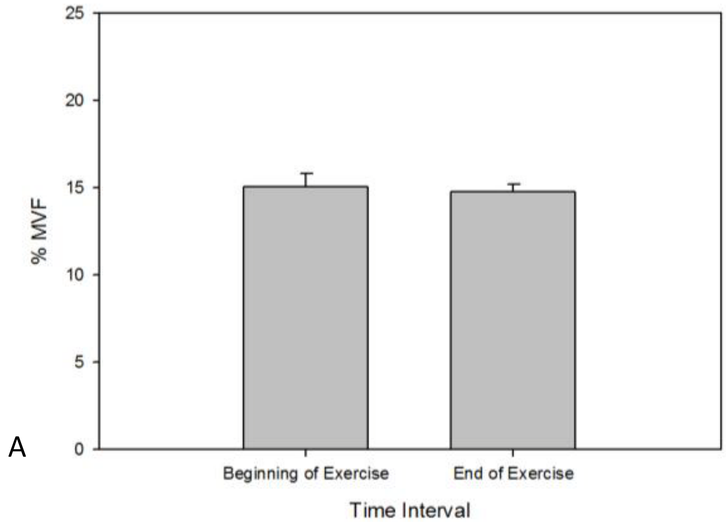


Figure 4.30 Detailed View of EMG Gaps in Intermittent Condition
A detailed view of the EMG gaps analysis (> 1% MVC, > 0.2 seconds) in both lateral and medial heads of the triceps brachii. Dotted line represents the 1% MVC threshold.

Mechanical force outputs were measured for both sustained isometric and On/Off conditions (Figure 4.31). Participants exerted a mean force of 14.64% MVF during the entire exercise duration when a 15% MVF sustained isometric contraction was elicited. There were no differences between the first 30 seconds of sustained isometric exercise (M = 15.034% MVF, SD = 2.838) and the final 30 seconds (M = 14.748% MVF, SD = 1.574), $t(13) = 0.399, p = 0.697, d = 0.125$. The On/Off condition resulted in mean force outputs of 4.31% MVF at the "rest" phase and 28.04% MVF at the "contraction" phase during the entire exercise protocol. There were no differences in either the "rest/off" phase between beginning (M = 4.213% MVF, SD = 1.935) and end (M = 3.670% MVF, SD = 0.816), $t(13) = 0.965, p = 0.353, d = 0.316,$ and the "contraction/on" phase between beginning (M = 29.666% MVF, SD = 1.767) and end (M = 28.894% MVF, SD = 2.300), $t(13) = 1.854, p = 0.088, d = 0.376.$



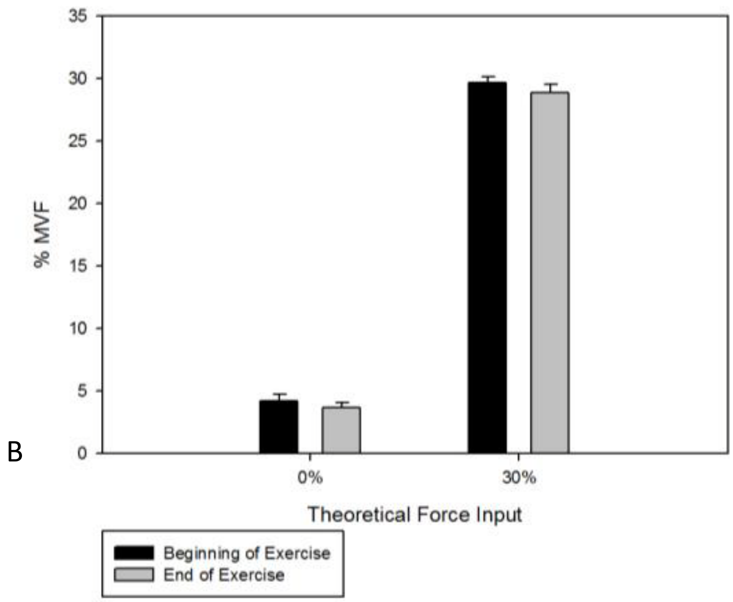


Figure 4.31 Mechanical Force Outputs for Sustained (A) and On/Off (B)

Output forces collected at the beginning (first 10 contractions or 30 seconds in sustained isometric) and end (last 10 contractions or last 30 seconds in sustained isometric) of sustained isometric and intermittent exercise. On/Off characterized with beginning and end force comparisons at two levels of force: rest (off) and contraction (on).

EMG Gaps and Mechanical Force Preliminary Discussion

The occurrence of EMG gaps has been shown to be a good predictor of work-related musculoskeletal disorders (Veiersted et al., 1993) and has been characterized as a method for assessing short muscular breaks. According to Nordander and colleagues (2000), the lack of muscular rest for individual motor units might be a risk factor for developing muscular pain. Gaps analysis was conducted in both medial and lateral EMG to discriminate against muscular rest and muscular activity. As expected, sustained isometric was composed of no gaps, a median of 0 gaps per minute in the lateral head and 0 gaps per minute in the medial head. The number of gaps in the On/Off condition was 8.696 and 8.717 gaps per minute for lateral and medial heads, respectively. Based on the force inputs characterized as 6 second cycle time and 50% duty cycle, theoretically, there should be 10 gaps per minute. Number of gaps may have been less than the idealized quantity if participants were not fully 'relaxed' during the designated 3-second 0% MVF. This was evidenced by the mechanical force output where mean force output during the rest period was 4.31% MVF. Past literature suggest that a high target force level ("contraction/on" phase) and prolonged contraction duration may decrease the muscle contractile relaxation time ($RT_{1/2}$) due to changes in high-energy substrate and metabolite concentrations (Vøllestad et al., 1997). This decrease in relaxation time may result in a higher force output within the 2-second mechanical force collection window. The mean and median duration per gap also supports the assumption of inhibited relaxation, as the calculated durations were less than 3 seconds. A large number of gaps (ie. 75th percentile values of 21.600 and 25.117 gaps per minute for lateral and medial heads, respectively) and short durations may also imply that brief muscular breaks were achieved but interspersed by random motor unit activity that were beyond the threshold (greater than 1% MVC).

General Results

A summary of the trends between sustained isometric and On/Off conditions using continuous exercise measurements and test battery responses are shown in Figure 4.32 and Figure 4.33. Table 4.1 summarizes the statistically significant parameter responses during exercise (rate of response).

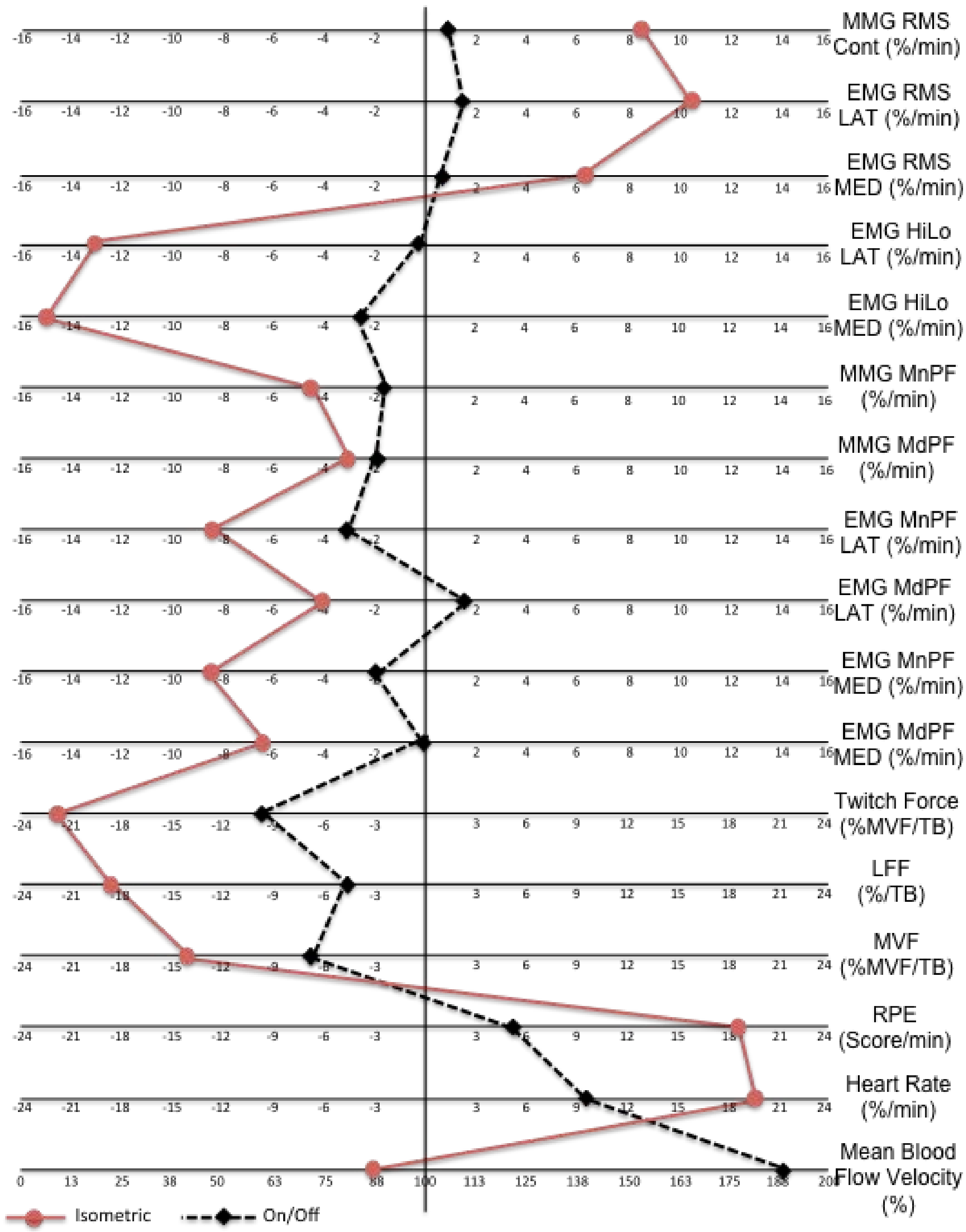


Figure 4.32 Summary of Mean Values for Continuous Responses Between Sustained and On/Off.
Refer to preliminary analysis to the direction that leads to a greater "fatigue response". In general, increasing amplitude parameters (RMS), ratings of perceived exertion, and heart rate, indicate an "exhaustive" response. Remaining parameters indicate a "fatiguing" response by larger decreasing values. Smaller mean blood velocity may indicate lower blood perfusion to working muscle while larger mean blood velocity may be indicative of greater metabolic demand.

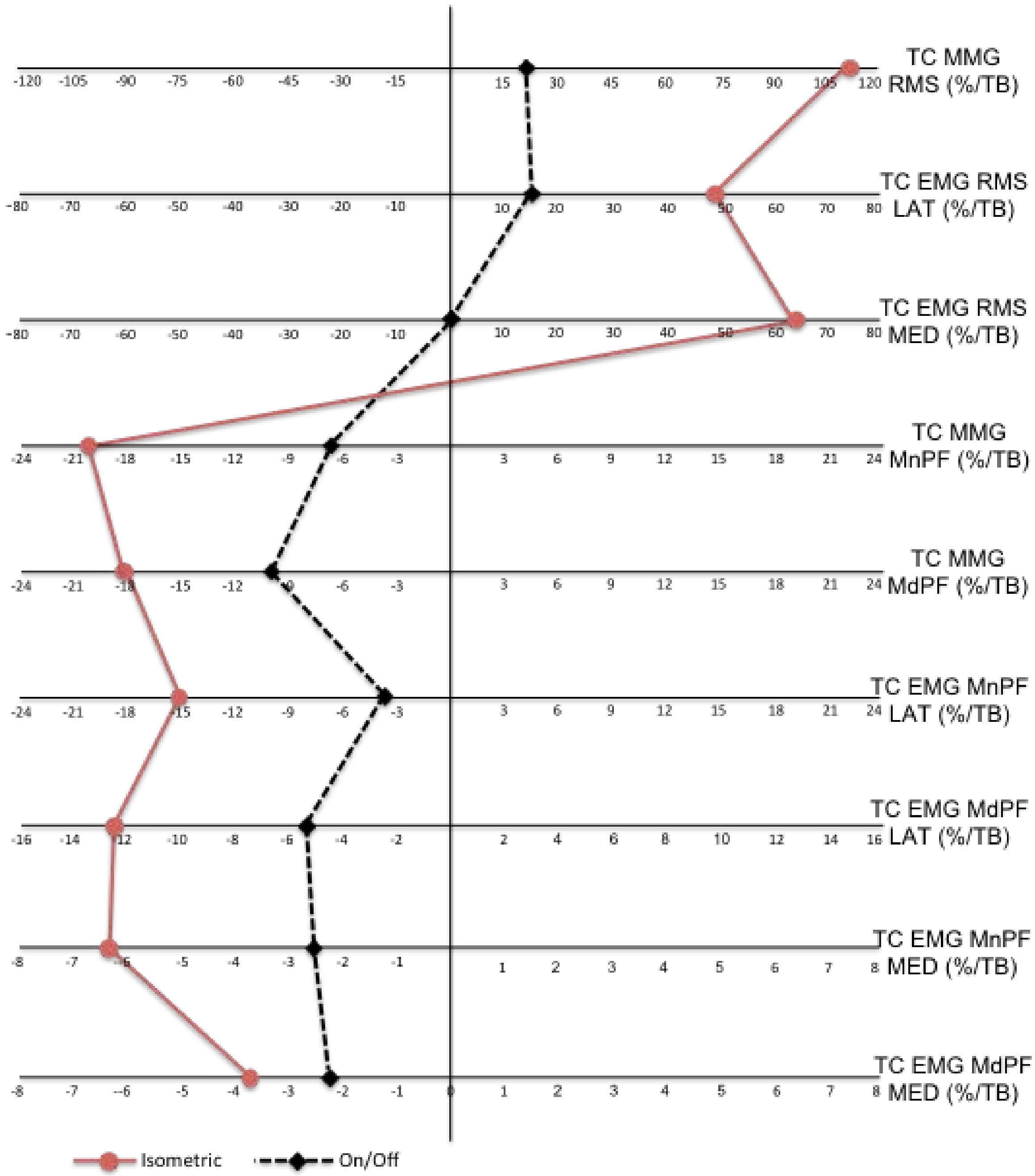


Figure 4.33 Summary of Mean Values for Test Contraction Responses Between Sustained and On/Off.
Refer to preliminary discussion for directions leading to greater "fatigue" response. As with Figure 4.32, increasing amplitude (RMS) values are indicative of larger fatigue response. Decreasing power domain variables (MnPF and MdPF) are indicative of greater fatigue response.

Table 4.1 Summary of Significant Responses During Exercise Between Sustained and On/Off Conditions

Measurement Parameter	Isometric Vs On/Off	P Value	Cohen's D
1. MMG RMS Medial Head Continuous	✓	P = 0.002	D = 1.383
2. EMG RMS Lateral Head Continuous	✓	P = 0.022	D = 0.903
3. EMG RMS Medial Head Continuous	✓	P = 0.025	D = 0.861
4. EMG HiLo Ratio Lateral Head Continuous	✓	P = 0.018	D = 0.795
5. EMG HiLo Ratio Medial Head Continuous	✓	P = 0.003	D = 0.965
6. MMG MnPF Medial Head Continuous		P = 0.201	D = 0.303
7. MMG MdPF Medial Head Continuous		P = 0.436	D = 0.062
8. EMG MnPF Lateral Head Continuous		P = 0.060	D = 0.549
9. EMG MdPF Lateral Head Continuous		P = 0.185	D = 0.367
10. EMG MnPF Medial Head Continuous	✓	P = 0.008	D = 1.056
11. EMG MdPF Medial Head Continuous	✓	P = 0.027	D = 0.945
12. Ratings of Perceived Exertion Continuous	✓	P = 0.000	D = 1.282
13. Heart Rate Continuous	✓	P = 0.000	D = 0.849
14. Triceps Blood Flow Velocity Continuous	✓	P = 0.000	D = 0.380
15. Maximum Voluntary Force Test Battery	✓	P = 0.010	D = 0.647
16. Twitch Force Test Battery	✓	P = 0.023	D = 0.598
17. Low Frequency Fatigue Test Battery	✓	P = 0.001	D = 1.026
18. MMG RMS Medial Head Test Contraction	✓	P = 0.002	D = 1.461
19. EMG RMS Lateral Head Test Contraction	✓	P = 0.043	D = 0.704
20. EMG RMS Medial Head Test Contraction	✓	P = 0.007	D = 1.023
21. MMG MnPF Medial Head Test Contraction	✓	P = 0.033	D = 0.753
22. MMG MdPF Medial Head Test Contraction		P = 0.161	D = 0.317
23. EMG MnPF Lateral Head Test Contraction	✓	P = 0.019	D = 0.788
24. EMG MdPF Lateral Head Test Contraction		P = 0.093	D = 0.582
25. EMG MnPF Medial Head Test Contraction		P = 0.138	D = 0.527
26. EMG MdPF Medial Head Test Contraction		P = 0.312	D = 0.201
27. Endurance Time*	✓	P = 0.002	-
28. EMG Gaps Lateral Head – #/min*	✓	P = 0.003	-
29. EMG Gaps Medial Head – #/min*	✓	P = 0.004	-
29. EMG Gaps Lateral Head – Mean Duration Per Gap*	✓	P = 0.003	-
30. EMG Gaps Lateral Head – Median Duration Per Gap*	✓	P = 0.004	-
31. EMG Gaps Medial Head – Mean Duration Per Gap*	✓	P = 0.002	-
32. EMG Gaps Medial Head – Median Duration Per Gap*	✓	P = 0.003	-
Number of Significant Responses (Isometric Greater Rate of Fatigue Response)		25	

✓ = Significantly different from sustained isometric ($\alpha = 0.05$)

* = Non-parametric test (Wilcoxon Signed-Rank Test)

General Discussion

A major finding from this study is that several subjects found the sustained isometric effort at 15% MVF demanding, reflected by physiological responses, performance (endurance time), and psychophysical data (ratings of perceived exertion). Since there is disparity in opinion of the best method to assess fatigue, a variety of measures and the method of its analysis should be used to determine whether fatigue occurred or not. Each measure reflects the change in function at different sites in the physiological processes involved in fatigue development. For instance electromyography (EMG) may be an index of the CNS process and subsequent motor unit excitation that accompany fatigue. On the other hand, mechanomyography (MMG) may reveal changes in the cross-bridge cycling mechanism and motor unit firing. In contrast, low-frequency fatigue using muscle stimulation may provide information in the cross-bridge functioning as well as calcium release and it's binding to troponin. Marked fatigue-related responses were found in the sustained isometric condition, with greater rate of responses during exercise observed in 25 of the 32 measured parameters. Among these measurement parameters are MMG (amplitude and frequency), EMG (amplitude and frequency), ratings of perceived exertion, force output, electrically evoked forces to assess twitch force and low-frequency fatigue, blood flow velocity, and heart rate. Additionally, parameters to describe performance (endurance time) and muscular rest (EMG gaps) suggest that the sustained isometric contraction led to a significantly lower completion time (quick to exhaustion) and there were no muscular breaks.

Although 25 of 32 measured parameters suggest that sustained isometric contractions led to greater and quicker fatigue responses, further interpretation is required to understand the relationship between common responses. One glaring discrepancy is the diverging response of the same measurement parameter during the test contraction and during the continuous analysis. In other words, why was there a statistical difference between exercise conditions in EMG mean and median power frequencies during continuous exercise and a slight difference during test contractions (or vice versa)? Simply, test contractions represent the cumulative effect up to pre-determined time periods (every 15 minutes) or up until exhaustion. Because sustained isometric condition led to a lower endurance time and hence one test contraction immediately at the end of exercise, the amount of work done reflected in a particular test contraction will be inherently different. The difference in the workload reflected by the test contraction may lead to greater variability, hence a larger p-value. The effect size, notwithstanding, was moderate to large despite lower p-values in the test contraction data. As discussed earlier, power spectral analysis may be inappropriate at lower force levels as its interpretation may be affected by its complexities. This inherent challenge with EMG frequency analysis in low-level activity may affect analysis during continuous measurement and test contractions. Frequency analysis during the continuous analysis may be affected by the continuous recruitment and derecruitment of motor units. During the test contraction, a sustained isometric contraction at 15% MVF, a different set of motor units may be recruited and may not reflect the exercise condition. However, the test contraction ensures that the response measure was not due to a changing force output. The continuous measures may better reflect the response over the duration of the exercise protocol rather than the cumulative effect.

Recovery after exercise was analyzed by comparing incremental recovery times to baseline and cessation. A comparison of the recoveries for both sustained isometric and intermittent cannot be unequivocally made, as the workload after each condition was different from one another. For instance, the rate of recovery was sought using 15-minute increments. It was observed that sustained isometric led to quicker recovery in most measurement parameters. However, what is not known is whether this recovery response is inherent of the condition itself or due to the different workload at the beginning of recovery (exercise cessation). This is particularly true if the intermittent condition did not lead into a large relative change in its response at the end of 60 minutes. For instance, previous literature has speculated a long-term peripheral fatigue effect, of up to 24 hours, after intermittent contraction exercise (Bystrom and Fransson-Hall, 1994). This effect was not observed in this study. However, there was a large depression in LFF ratio (tetanic stimulations at 100 Hz and 20 Hz) up to 30 minutes into recovery, implying the possibility of prolonged fatigue effects if intermittent exercise was completed to exhaustion. Recovery can be compared if the workload was identical based on time or based on level of fatigue (i.e., exhaustion). Instead, this study indeed assessed performance based on a 60-minute protocol. The recovery for each condition can therefore be evaluated to characterize the effect of the mechanical loading scheme.

Interestingly, twitch force and LFF report contradictory results in the sustained isometric condition. As expected, twitch force was depressed 24 hours post-sustained isometric exercise. Low frequency fatigue, on the contrary, recovered after 45 minutes despite previous research that considered otherwise. It is possible that the loading scheme did not result in long-term effects as the reliability of using twitch forces to assess fatigue have come into question (Edwards et al., 1977). On the other hand, another possibility is the influence of the preceding tetanic stimulation before the single twitch stimulus. Although ample rest time was given between the electrically induced contractions, previous research has shown an increase in twitch tension immediately after tetanic stimulation (Takamori et al., 1971), a phenomenon called potentiated twitch. Two opposing processes may co-exist to complicate the assessment of muscle fatigue: one that enhances twitch amplitude (potentiation) and another that decreases it (fatigue) (Kufel et al., 2002). However, Alway and colleagues (1987) found similar twitch properties after a fatiguing exercise and an electrically induced contraction. Kufel and colleagues (2002) found a decline in potentiated twitch after a fatigue protocol and hypothesized that it may be influenced to a greater degree by low-frequency fatigue than mechanisms generating an unpotentiated twitch. As a result, there may be a greater reduction in potentiated twitch (or improved sensitivity to fatigue) if low-frequency fatigue inhibits myosin phosphorylation, a proposed mechanism of potentiation (Kufel et al., 2002). In this scenario, twitch force may be more sensitive than the low-frequency tetani stimulations in detecting muscle fatigue (Vøllestad, 1997), possibly explaining the results found in this study.

Conclusion

Physical variation has been described as a remedy against musculoskeletal disorders in jobs consisting of sustained and low-level loads or repetitive work operations (Mathiassen, 2006). This study adds to the current database by comparing an isometric sustained exertion at 15% MVF and an intermittent contraction between 0% and 30%. In a controlled lab setting, each condition was performed for 60 minutes or until exhaustion at a mean force level of 15% MVF, 10-second cycle time, and 50% duty cycle, task parameters that are relevant to occupation. Proprioceptive feedback was used to elicit these forces.

Measuring the response to a mechanical exposure pattern based on a sustained isometric force or an intermittent pattern may require the collective agreement over a number of measurement parameters. Each measurement parameter may provide information on a few limited mechanisms that are associated with fatigue development. Thus there is no one "gold standard". This study used 1 performance measure (endurance time) and 8 commonly used fatigue-related measurement tools, which were then analyzed based on 14 parameters. Measures were taken from both medial and lateral heads of the triceps brachii muscle.

Results from this study show a collective agreement between 25 of 32 individual parameters, suggesting an overwhelming increase in fatigue response when exposed to a sustained isometric contraction. The intermittent contraction, in contrast, led to a longer endurance time and lower fatigue response. Recovery was analyzed to characterize each exercise condition. In general, twitch force and low-frequency fatigue values suggest long-lasting effects in the sustained isometric condition. The intermittent condition also revealed recovery values that were significantly different from baseline even though the On/Off exercise was not completed until exhaustion.

Chapter V

The Effects of Mechanical Exposure Diversity in Physiological and Psychophysical Responses

Introduction

Current trends in industry are leaning towards specialized production systems that have led to reduced cycle times, standardized work tasks, and ultimately a reduction in restorative work breaks and pauses (Wells et al., 2007). The lack of exposure variation has since become an important issue in contemporary work as the reduction of peak loads and extreme postures may not be applicable in low-level, less varying work tasks (Mathiassen, 2006; Wells et al., 2007). Exposure variation has been argued as a potential "remedy" against work-related musculoskeletal disorders (Mathiassen, 2006). However, mechanical variation is not solely an occupational-related phenomenon. Evidenced in the sports biomechanics literature, mechanical variability may facilitate adaptation to changing environments, improve coordination, and reduce injury risk in contrast to "optimal" movement patterns (Bartlett et al., 2007).

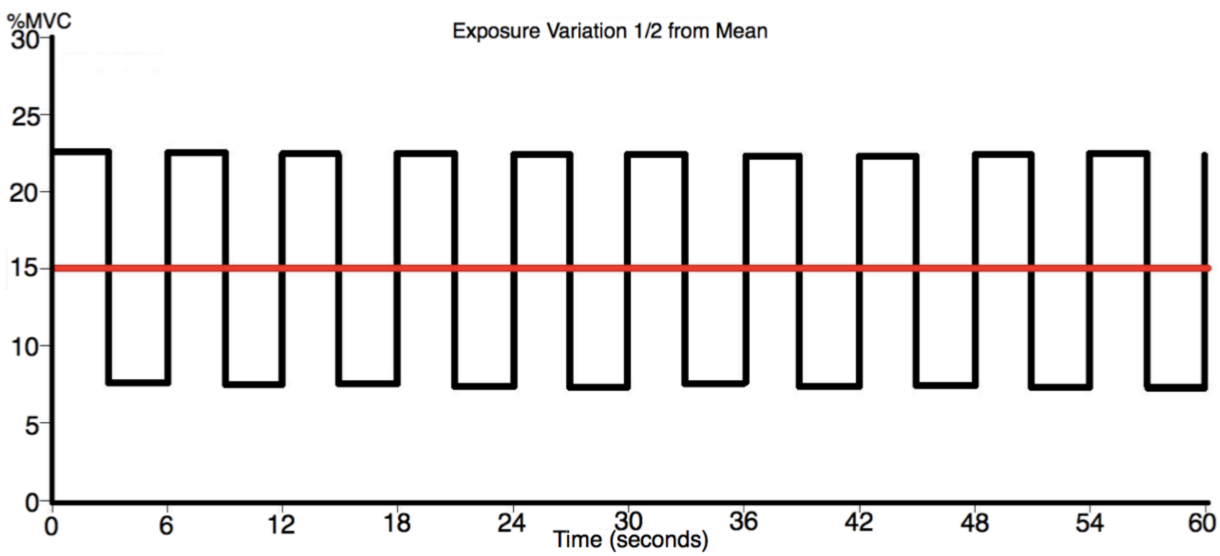
Exposure diversity is closely linked to variation, as it is a descriptor of similarities and discrepancies between different exposure entities (Mathiassen, 2006). Diversity can be assessed based on amplitude, frequency, and duration of biomechanical exposure (Winkel and Westgaard, 1992). Unfortunately there is no consensus of appropriate and standardized metrics for variation and diversity (Wells et al., 2007). If metrics for diversity are established, potential research can be focused towards guidelines and tools for ergonomic practitioners when faced with WMSD in sedentary work.

The aim of this chapter is to understand the physiological and psychophysical effects of exposure diversity, based on changes in amplitude, to further understand its relationship to musculoskeletal outcomes. Realizing the lack of metrics to define diversity, traditional statistical expressions of dispersion, such as amplitude deviations from the mean force, were used to dictate the change in amplitude. The secondary purpose is to understand the effects of these diverse exposure patterns in relation to a sustained isometric contraction and classical intermittent protocol.

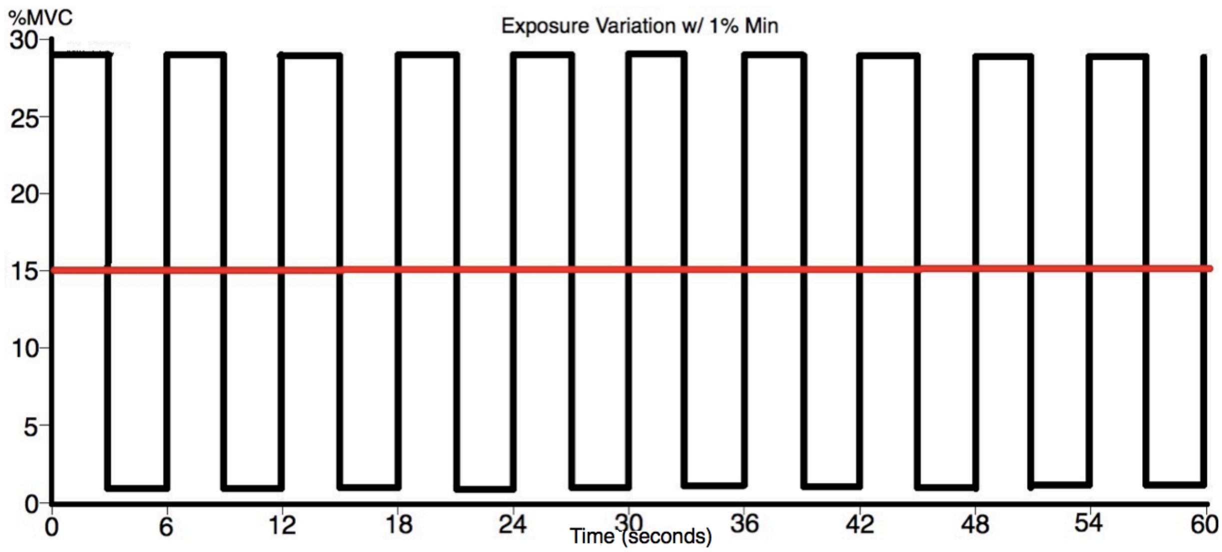
Methods

Exercise Conditions

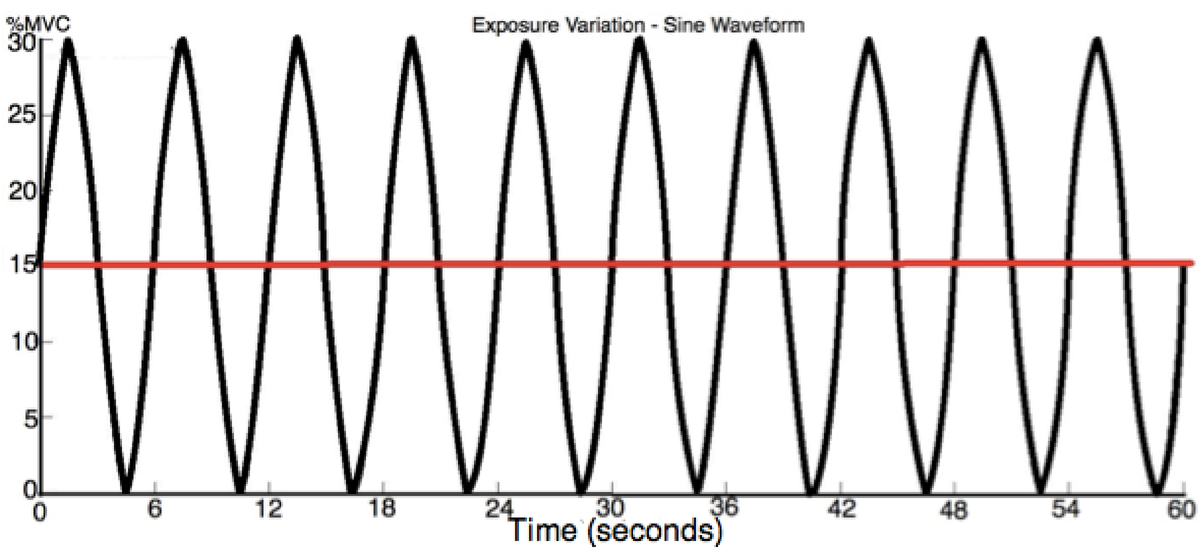
Participants performed three intermittent conditions: an intermittent elbow extension contraction between two levels of force defined as +/- ½ amplitude from the 15% MVF mean (7.5% - 22.5% MVF), an intermittent elbow extension between 1% MVF and 29% MVF, and an intermittent contraction composed of a Sinusoidal wave pattern that peak at 0% and 30% MVF (Figure 5.1). Conditions were randomized and performed on separate days with at least 7 days between sessions. Participants were given sufficient practice time to reduce learning effects.



A



B



C

Figure 5.1 MinMax (A), 1 Percent (B), Sinusoidal (C) Exercise Conditions

All conditions had a mean effort of 15% MVF and were composed of a duty cycle of 50% and cycle time of 6 seconds. The 15% MVF was expressed as a proportion of MVF activity from the largest magnitude of the three pre-experiment MVF contractions.

All conditions were completed using proprioceptive feedback by resisting against a motorized arm apparatus. Detailed explanation of the arm apparatus is found in Chapter III. The intermittent contraction between 7.5% and 22.5% MVF (MinMax) was programmed to transition between the two levels of force with seven steps totaling 700-milliseconds. Visual feedback via a light switch was used to ensure that the participant exerted extension forces to place the arm apparatus at a desired vertical position. Similarly, the 1% and 29% condition (1 Percent) consisted of 700 milliseconds between forces, providing 2.3 seconds at each force. Finally, the 0% to 30% MVF condition composed of sinusoidal waves (Sinusoidal), with 12 points peak to peak. The peak-to-peak duration was 3 seconds.

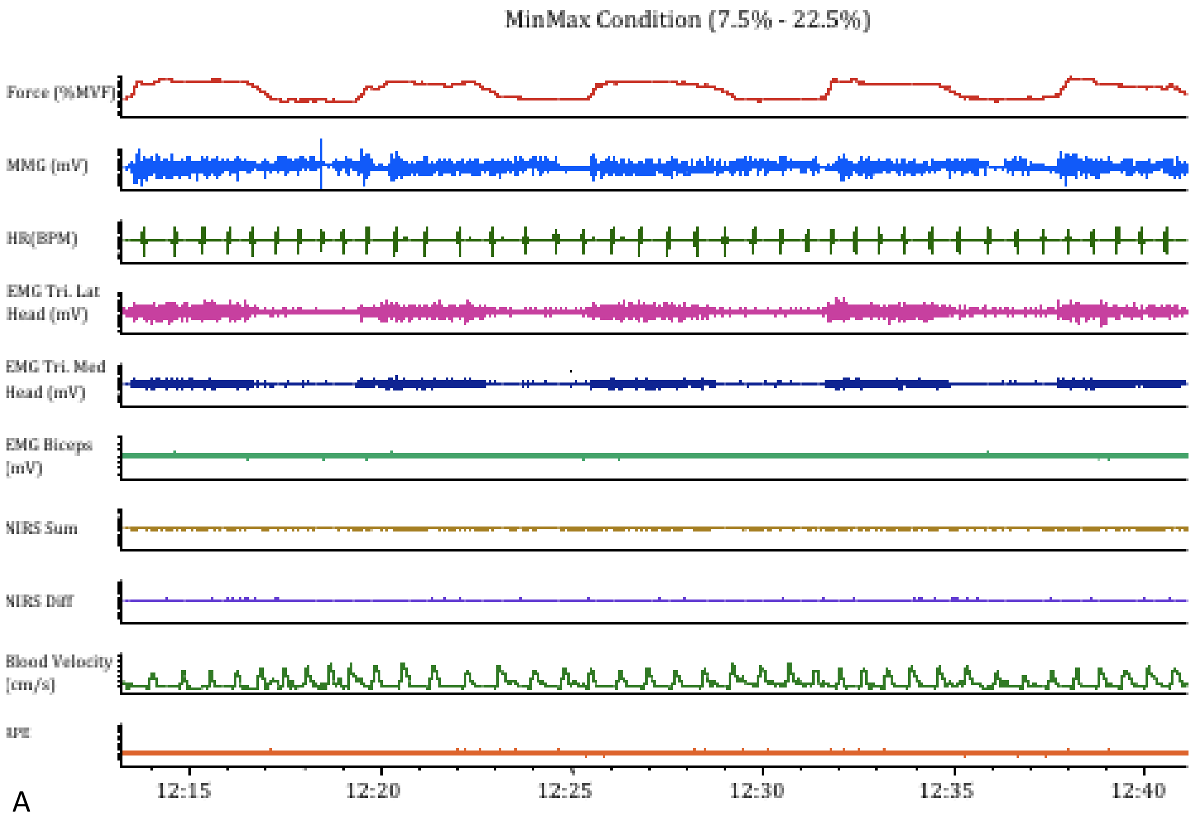
Procedure

Chapter III outlined the general procedure for this study. However, to highlight some key aspects of this procedure:

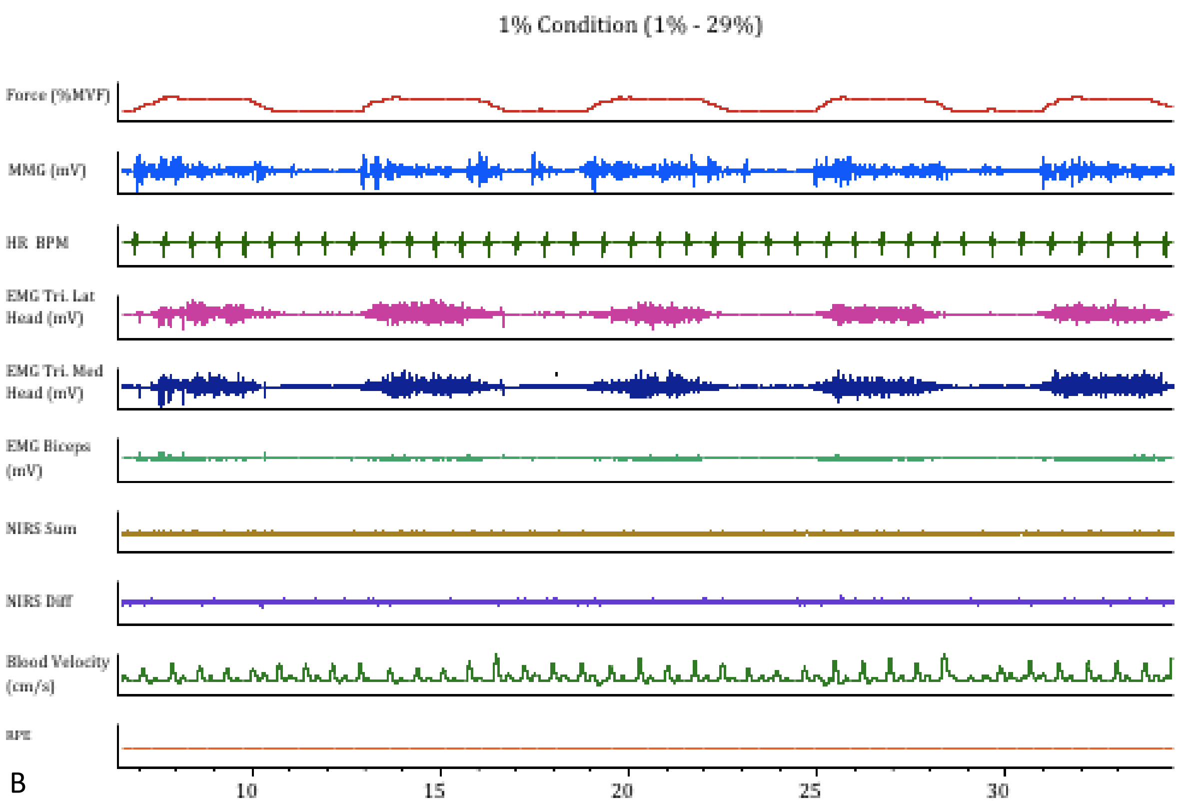
A 10-minute baseline trial was collected prior to every exercise condition. Midway through the baseline trial, a test battery of electrical muscle stimulations, test contraction, and maximum voluntary force was measured.

Every condition was completed up to 60 minutes or until exhaustion. Measures of EMG, MMG, blood velocity, force, and heart rate were continuously recorded and subsequently analyzed every 2 minutes for a 30-second window. Test batteries were collected every 15 minutes. Figure 5B are example data collection profiles for all measures that were monitored continuously.

At the end of exercise, a test battery was immediately collected. Subsequent test batteries were collected in 15-minute intervals during 60 minutes recovery and 24 hours post-exercise.


MinMax Condition (7.5% - 22.5%)
Force (%MVF)
MMG (mV)
HR(BPM)
EMG Tri. Lat Head (mV)
EMG Tri. Med Head (mV)
EMG Biceps (mV)
NIRS Sum
NIRS Diff
Blood Velocity (cm/s)
RPE
12:15
12:20
12:25
12:30
12:35
12:40

A


1% Condition (1% - 29%)
Force (%MVF)
MMG (mV)
HR BPM
EMG Tri. Lat Head (mV)
EMG Tri. Med Head (mV)
EMG Biceps (mV)
NIRS Sum
NIRS Diff
Blood Velocity (cm/s)
RPE
10
15
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B

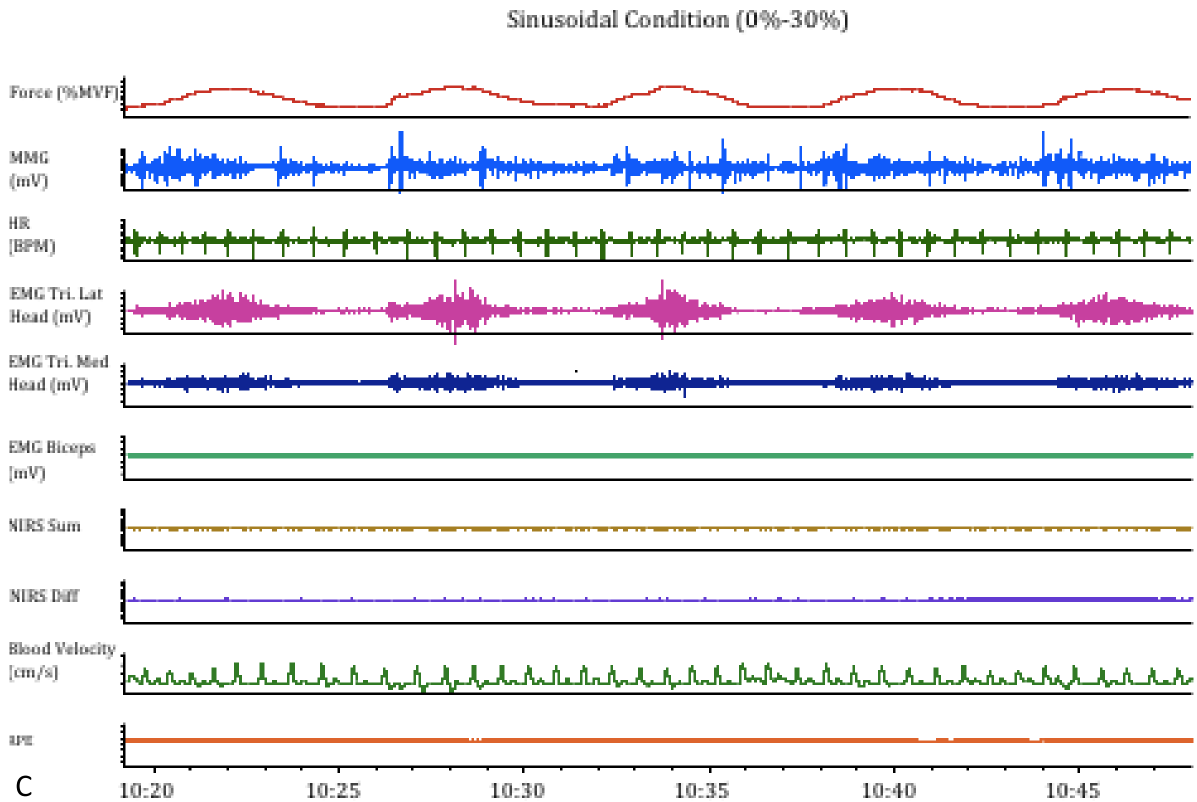


Figure 5.2 Screenshot Data Collection Profiles

Continuous measurements of force (% MVF), MMG (mV), HR (BPM), EMG Lateral (mV), EMG Medial (mV), EMG Biceps (mV), NIRS Sum and Difference, Blood Velocity (cm/s), and RPE (mV) for MinMax (A), 1 percent (B) and sinusoidal contraction (C).

Measurement Parameters

Detailed descriptions of the measurement tools and parameters can be found in Chapters III and IV. To summarize:

Electromyography were analyzed in both time and frequency domains. Root mean square (RMS) values are representative of changes in amplitude while mean and median frequencies quantified shifts in the power spectrum. Hi-Lo power ratios were calculated to distinguish changes in low and high frequency data. EMG gaps were measured for the quantity (# gaps/minute) and duration (mean and median durations per gap).

Mechanomyography was also analyzed in both time and frequency domains. Changes in MMG amplitude were quantified by changes in RMS values. Mean and median power frequencies were measured in the power spectrum.

Triceps blood flow was calculated using brachial artery mean blood velocity and brachial artery diameter. Since brachial artery diameter remained consistent during the exercise protocol, mean blood velocity was used as a direct measure of triceps blood flow.

Ratings of perceived exertion (RPE) was collected every two minutes and were expressed as a rating between 0 (no exertion) to 100 (maximum/intense exertion).

Mean mechanical force outputs were measured for each contraction, using a 2 second window at both low and high levels of contraction (i.e. the two theoretical levels of 7.5% and 22.5% in MinMax). A window of 0.5 seconds was used to measure peak mechanical force outputs in the sinusoidal condition.

Test batteries consisted of muscle electrical stimulation (twitch force and tetanic forces at 100 Hz and 20 Hz), a test contraction at 15% MVF, and maximum voluntary force (MVF). Test batteries were collected every 15 minutes during exercise and every 15 minutes during recovery.

Statistical Analysis

A complete description of the statistical analysis is described in Chapter IV. Key points to consider include:

Measures were plotted against time and fitted with a linear or non-linear regression line. Exercise values were normalized to baseline (heart rate, blood velocity, RPE) or to the first 2-minute interval (EMG and MMG). Regression lines, plotted for every participant in every condition, were compared by one-way repeated measures ANOVA. Dunnett's test was used to compare the three conditions against the sustained isometric condition and to compare the intermittent contraction pattern with the MinMax and

Sinusoidal patterns. A one-tailed a priori analysis was conducted as hypothesis driven questions were undertaken. Alpha level was set at 0.05.

Test battery measurements were collected over 15-minute intervals during exercise and recovery. Cessation and recovery times were compared to baseline using one-way repeated measures ANOVA and Dunnett's test. The baseline value was set as the reference condition for the Dunnett's test. The rate of response, using test batteries, in both exercise and recovery, were normalized to baseline (exercise) or cessation (recovery). One-way repeated measures ANOVA determined differences between conditions based on exercise or recovery slopes. Recovery was measured for all test battery measurement parameters; however, for this study, maximum voluntary force, low-frequency fatigue, and twitch force were analyzed.

Endurance time, number of EMG gaps per minute, mean and median duration of each gap was analyzed using Friedman's test and a subsequent Wilcoxon signed ranks test to compare conditions (sustained isometric vs. 1 Percent, MinMax, Sinusoidal and On/Off vs. MinMax, Sinusoidal). Bonferroni corrections were applied to set the alpha level at 0.01.

Two participants were excluded from analyses, as they did not meet the criteria outlined in Chapter III. The two participants were identical to those eliminated in the Sustained Isometric vs. Intermittent Contraction study.

A complete statistical analysis for each parameter during exercise is shown in Appendix B and during recovery in Appendix C. Checkmarks indicate significant differences from Sustained ($p < 0.05$) and crosses indicate significant differences from On/Off ($p < 0.05$). Stars indicate significant differences from baseline during recovery.

Measurement Results and Preliminary Discussion

Endurance Time Results

Participants performed MinMax (7.5% - 22.5%), 1 Percent (1% - 29%), and Sinusoidal (0% - 30%) conditions for up to 60 minutes or until exhaustion. The 1 Percent (Median = 3202 seconds, 25^{th} = 650, 75th = 3600, z = -2.580, p = 0.005) and Sinusoidal (Median = 2205 seconds, 25th = 711, 75th = 3600, z = -2.353, p = 0.009) led to longer endurance times than sustained isometric (Median = 579 seconds, 25th = 408, 75th= 1191.5). When compared to the On/Off condition (Median = 3600 seconds, 25th = 2274, 75th = 3600), both MinMax (Median = 1474 seconds, 25th = 694, 75th = 2901, z = -2.756, p = 0.003) and Sinusoidal (z = -2.521, p = 0.006) led to shorter endurance times (Figure 5.3).

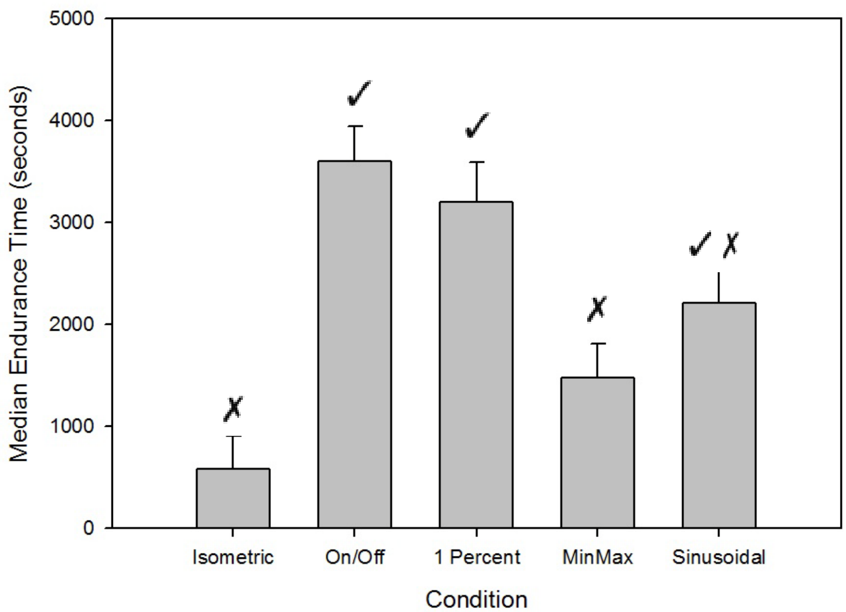


Figure 5.3 Median Endurance Time for 5 Conditions. See Statistical Analysis for Symbol Interpretation

Endurance Time Preliminary Discussion

Endurance time was used as a measure of "performance", as participants were instructed to complete conditions up to 60 minutes or until exhaustion. As discussed in Chapter IV, sustained isometric and On/Off endurance time results agreed with previous literature. Although research has been conducted on intermittent contractions of varying duty cycles, little is known of endurance times related to different intermittent contractions of varying amplitude. Participants performed the classical intermittent contraction (On/Off) for a longer endurance time, followed by 1 Percent, Sinusoidal, MinMax, and sustained isometric. Sinusoidal was statistically different from both sustained isometric and On/Off conditions, implying that its endurance time may be approximately half way between the two conditions.

The 1 Percent condition, on the other hand, had a significantly lower endurance time compared to sustained isometric and had values closer to On/Off. MinMax led to a shorter completion time than the intermittent contraction but was not statistically different to the sustained isometric exertion.

One possible explanation of the longer endurance time (relative to the sustained isometric condition) or shorter endurance time (relative to the intermittent contraction) is the relationship between muscle activation and endurance time. According to Hunter and Enoka (2003), altering the level and pattern of muscle activation may affect the endurance time at submaximal levels of force. In the Hunter and Enoka (2003) study, the endurance time of a 20% MVC sustained isometric contraction increased by delaying the recruitment of more fatigable motor units or by changing the distribution of EMG activity among synergist muscles. It is possible that variation of force amplitude alters muscle activation to prolong endurance time. The On/Off condition may have altered muscle activation to a greater degree and hence resulted in longer endurance times.

Force Results

The rate of force decrement was analyzed using maximum voluntary forces elicited at baseline, every 15 minutes during exercise, and cessation of exercise if the condition was completed in less than 60 minutes. Conditions were fitted with linear regression (MinMax: $r^2 = 0.806$, 1 Percent: $r^2 = 0.618$, Sinusoidal: $r^2 = 0.739$) and compared using repeated measures ANOVA. A typical regression fit for MinMax, 1 Percent, and Sinusoidal contraction patterns are shown in Figure 5.4. Sinusoidal ($M = -6.369$% MVF/Test Battery, $SD = 5.207$, $p = 0.002$) and 1 Percent ($M = -3.739$% MVF/Test Battery, $SD = 3.501$, $p = 0.000$) had greater rate of force decrement than sustained isometric, $F(0,4) = 7.054$, $p = 0.003$, $\eta_p^2 = 0.370$).

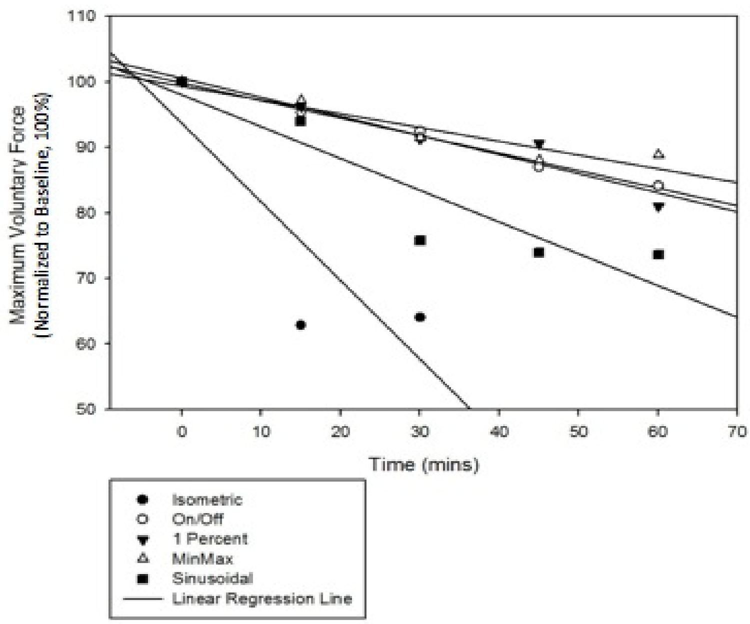
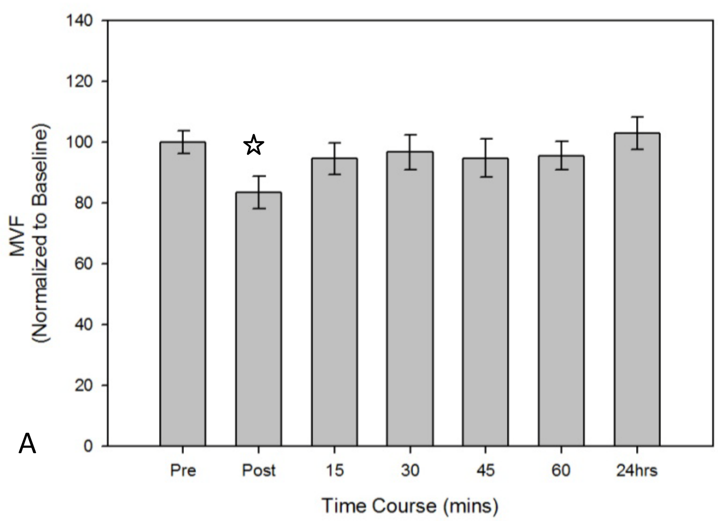
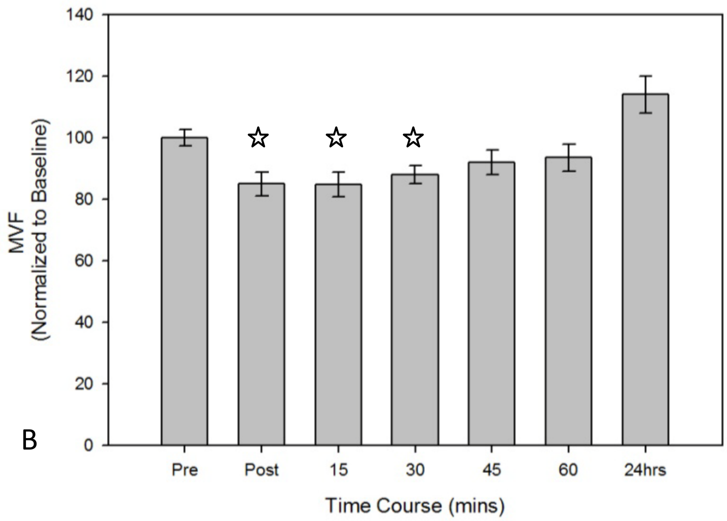


Figure 5.4 Typical Maximum Voluntary Force Response During Exercise for 5 Conditions

Recovery was measured at 15-minute intervals and 24 hours post exercise. Forces at cessation and during recovery were normalized to the condition's baseline value (Figure 5.5)





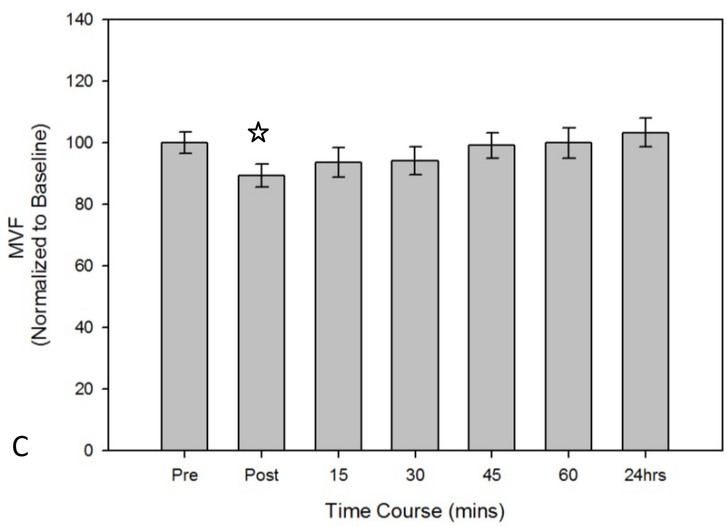


Figure 5.5 Maximum Voluntary Force Comparisons Between Baseline & Recovery for MinMax (A), Sinusoidal (B), and 1 Percent (C). See Statistical Analysis for Symbol Interpretation.

Both MinMax (baseline: M = 76.813% MVF, SD = 13.039) and 1 Percent (baseline: M = 77.101% MVF, SD = 12.333) led to a decrease in maximum voluntary force at cessation (MinMax cessation: 64.149% MVF, SD = 14.905, p = 0.001; 1 Percent cessation: 68.807% MVF, SD = 10.362, p = 0.002). Recovery values in MinMax and 1 Percent were not significantly different than baseline. The Sinusoidal condition, on the other hand, resulted in a decrease in force at cessation (M = 69.310 % MVF, SD = 11.362, p =

0.011), recovery at 15 minutes (M = 69.111% MVF, SD = 11.397, p = 0.009), and recovery at 30 minutes (M = 71.738% MVF, SD = 8.837, p = 0.046) when compared to baseline (M = 81.531% MVF, SD = 10.090).

Force Preliminary Discussion

Similar to sustained isometric and the classical intermittent conditions, there was a decrease in maximum force during exercise in MinMax, 1 Percent, and Sinusoidal exercise. Compared to the sustained isometric condition, both 1 Percent and Sinusoidal had a slower rate of response during exercise. MinMax was not statistically different from the sustained isometric condition but had a mean rate of response that was slightly quicker than On/Off. Chapter IV details possible mechanisms that may explain the decrease in maximal force.

There were no differences in the rates of recovery of maximum force output between conditions. The maximum forces at recovery when compared to baseline, however, may provide insight to the long-term effects of a particular condition. A potential comparison can be made between MinMax and sustained isometric as both had median completion values of 1474 seconds (24 minutes, 34 seconds) and 579 seconds (9 minutes, 39 seconds), respectively. This suggests that both conditions were completed until exhaustion. Interestingly, MinMax condition led to a depressed force output at cessation but quickly recovered henceforth. This is in contrast to sustained isometric, which recovered 30 minutes into recovery. As suggested in Chapter IV, the force rise time may have been sufficient to allow excitation pulses to follow the contraction inputs, as the two levels of force were relatively close to one another (15% difference). It may further emphasize the potential long-term force reduction in the On/Off (0% - 30%) condition, as recovery was significantly depressed over 30 minutes albeit without exhaustion. The Sinusoidal condition also led to significant decreases in force during recovery, up to 30 minutes. Unlike On/Off or 1 Percent, there is continuous loading and unloading of force, leading to higher energy demand. Chasiotis and colleagues (1987) suggest that each contraction during intermittent work results in greater ATP utilization and a pronounced decrease in force production.

Twitch Results

Exercise twitch force (Figure 5.6) was fitted with a linear regression line with a mean goodness of fit of r^2 = 0.618 (1 Percent), r^2 = 0.721 (MinMax), and r^2 = 0.574 (Sinusoidal). Of the three intermittent contractions, Sinusoidal (M = -10.095% MVF/Test Battery, SD = 14.104, p = 0.035) led to a significantly slower rate of twitch force decrement than sustained isometric, $F(0,4) = 2.047$, $p = 0.057$, $\eta_p^2 = 0.270$.

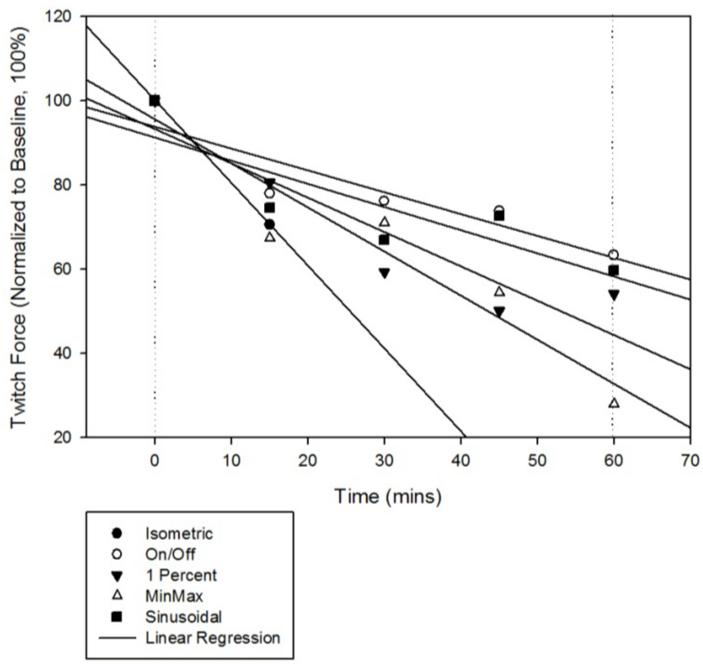


Figure 5.6 Typical Twitch Force Response During Exercise for 5 Conditions

Recovery values were compared to baseline (Figure 5.7). In the 1 Percent condition, recovery twitch force values remained depressed up to 60 minutes (Baseline: M = 5.239% MVF, SD = 2.526; 60 minutes: M = 4.020% MVF, SD = 2.526). MinMax, on the other hand, led to lower twitch values at cessation (M = 3.848% MVF, SD = 2.221) and at 60 minutes recovery (M = 3.796% MVF, SD = 1.513). The Sinusoidal condition had lower twitch force values at cessation (M = 3.755% MVF, SD = 1.435, p = 0.049) and 15 minutes recovery (M = 3.748% MVF, SD = 1.804, p = 0.047) when compared to baseline (M = 4.790% MVF, SD = 1.685).

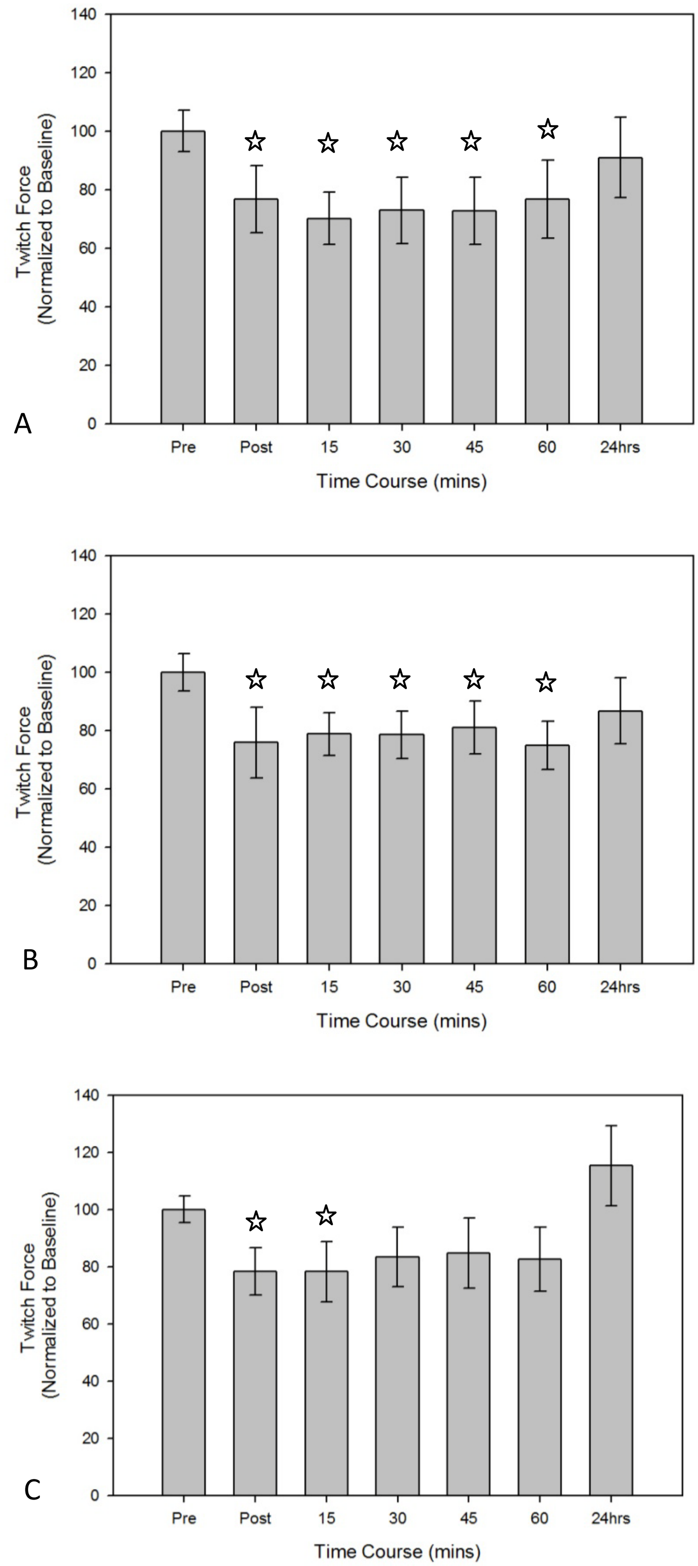


Figure 5.7 Twitch Force Comparisons Between Baseline & Recovery for 1 Percent (A), MinMax (B), and Sinusoidal (C)

Twitch Force Preliminary Discussion

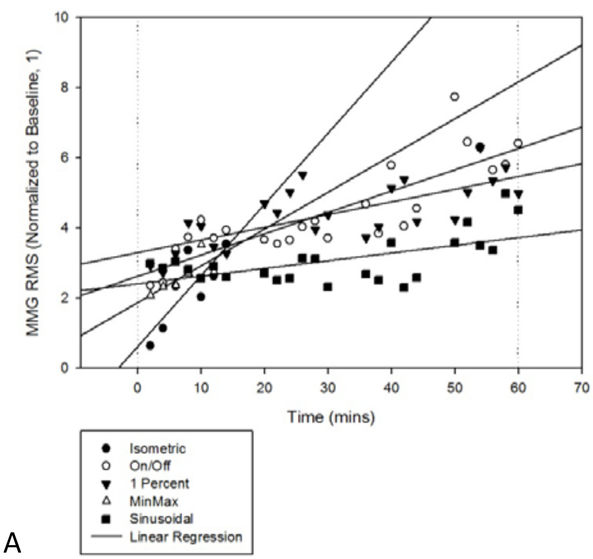
The findings from this study suggest that a slower rate of twitch force decrement in the Sinusoidal contraction pattern when compared to sustained isometric. This is similar to the findings in the previous chapter, where the classical intermittent contraction (On/Off) led to slower rate of response. There were no differences between Sinusoidal and On/Off. MinMax and 1 Percent also led to slower rate of twitch force decrement, relative to sustained isometric, but were not statistically significant. The decrease in twitch force is consistent with previous literature.

Although there were no differences in rate of recovery between conditions, twitch response compared to baseline may provide insight into long-term fatigue for each condition. The 1 Percent led to prolonged twitch force decrement within 60 minutes but recovered 24 hours post exercise. This is in contrast to the sustained isometric effort at 15% MVF, where twitch force remained depressed 24 hours post exercise, and to the On/Off intermittent pattern, where twitch force recovered within the first 15 minutes. It is possible that differences between 1 Percent and On/Off is due to the workload during exercise. As mentioned earlier, 1 Percent resulted in exhaustion more often than the classical intermittent contraction. Described in Chapter IV, the intensity and workload of the exercise protocol may affect the prolongation of twitch force decrement. The MinMax condition led to a fluctuation of twitch force during recovery. A significant decrease in twitch force was identified at cessation and at 60 minutes recovery. The twitch force between cessation and 60 minutes were also depressed, and although not statistically significant, may reveal long-term fatigue at those time intervals. The Sinusoidal condition resulted in significant depressed twitch forces up to 15 minutes recovery. This may contradict the possible notion that Sinusoidal may have had a higher energy demand, and as a result long-term peripheral fatigue, than the intermittent contraction between 0% and 30%.

Continuous Measures RMS Results

MMG and EMG were collected continuously for the duration of the exercise and later analyzed in 30-second windows, every 2 minutes. The mean root mean square values were obtained from these windows.

Data was normalized to the first interval of exercise (2 minutes) and 4 minutes of data were eliminated after every test battery. Linear regression was conducted with resultant best-fit lines of $r^2 = 0.281$ (1 Percent), $r^2 = 0.302$ (MinMax), and $r^2 = 0.273$ (Sinusoidal) for MMG. EMG medial RMS had goodness of fit lines of $r^2 = 0.432$ (1 Percent), $r^2 = 0.439$ (MinMax), and $r^2 = 0.320$ (Sinusoidal). Best-fit lines for EMG lateral RMS were $r^2 = 0.413$ (1 Percent), $r^2 = 0.409$ (MinMax), and $r^2 = 0.463$ (Sinusoidal). Typical responses are shown in Figure 5.8.



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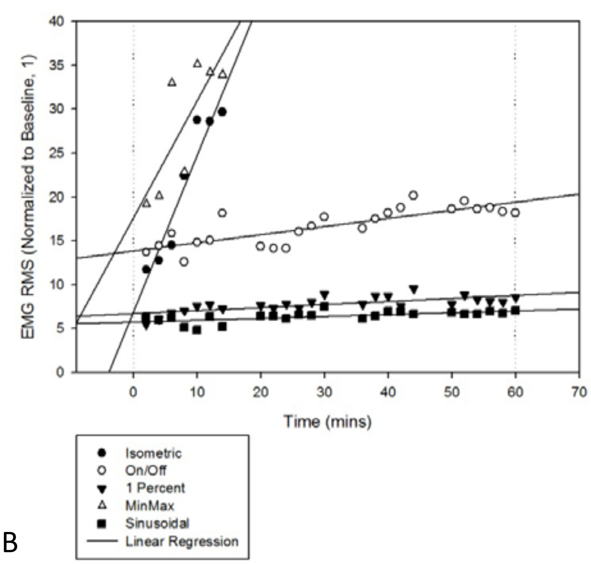


B

Figure 5.8 Typical Continuous MMG (A) and EMG RMS (B) Response During Exercise For Medial and Lateral Heads of Triceps Brachii for 5 Conditions

MMG RMS of the medial head revealed that 1 Percent (M = 0.715%/min, SD = 1.733, $p = 0.000$), MinMax (M = 1.463%/min, SD = 2.236, $p = 0.000$), and Sinusoidal (M = 0.501%/min, SD = 1.864, $p = 0.000$) conditions had significantly slower rates of RMS increase than sustained isometric, $F(0,4) = 11.796$, $p = 0.001$, $\eta_p^2 = 0.496$. There were no differences between On/Off and the Sinusoidal and MinMax conditions.

EMG RMS of the medial head revealed similar results to MMG. 1 Percent (M = 0.709%/min, SD = 1.104, $p = 0.003$), MinMax (M = 1.266%/min, SD = 1.759, $p = 0.006$), and Sinusoidal (M = -0.108%/min, SD = 1.402, $p = 0.001$) led to significantly slower rate of RMS increase than sustained isometric, $F(0,4) = 4.984$, $p = 0.041$, $\eta_p^2 = 0.293$. However, no differences were found in the rate of response during MinMax and Sinusoidal when compared to On/Off.

Lateral head EMG RMS values led to similar trends to MMG and EMG of the medial head. A slower rate of RMS increase was found in 1 Percent (M = 1.038%/min, SD = 1.873, $p = 0.002$), MinMax (M = 3.416%/min, SD = 5.824, $p = 0.019$), and Sinusoidal (M = 0.278%/min, SD = 3.216, $p = 0.001$) when compared to sustained isometric, $F(0,4) = 4.792$, $p = 0.030$, $\eta_p^2 = 0.285$. Again, there were no differences between On/Off and the Sinusoidal and MinMax conditions.

Test Contraction RMS Results

Test batteries were collected every 15 minutes and at cessation of exercise. Each test battery was composed of a test contraction at 15% MVF that was sustained for 12 seconds but analyzed over the middle 10 seconds. Achieving the maximum goodness of fit determined the linear or non-linear regression curve to model the fatigue response (Figure 5.9). Logarithmic curves were applied for MMG (1 Percent: mean $r^2 = 0.507$, MinMax: mean $r^2 = 0.806$, Sinusoidal: mean $r^2 = 0.702$), EMG medial head (1 Percent: mean $r^2 = 0.516$, MinMax: mean $r^2 = 0.727$, Sinusoidal: mean $r^2 = 0.645$), and EMG lateral head (1 Percent: mean $r^2 = 0.584$, MinMax: mean $r^2 = 0.718$, Sinusoidal: mean $r^2 = 0.568$). 1 Percent (M = 16.061% MVC/Test Battery, SD = 24.409, $p = 0.000$) led to a slower rate of MMG RMS increase than sustained isometric. Sinusoidal (M = 57.175% MVC/Test Battery, SD = 54.235, $p = 0.028$) also led to a slower rate of MMG RMS increase when compared to the static sustained isometric condition. There was a significant difference between MinMax (M = 104.201% MVC/Test Battery, SD = 66.780, p = 0.001) and On/Off, as MinMax led to a quicker rate of response. EMG RMS of the medial head revealed a significantly slower rate of response in the 1 Percent (M = -1.848% MVC/Test Battery, SD = 21.729, p = 0.026) than the sustained isometric. The MinMax (M = 76.065% MVC/Test Battery, SD = 122.886, p = 0.008), on the contrary, led to an increased rate of RMS increase when compared to the On/Off condition. Finally the lateral head EMG RMS analysis revealed a significantly quicker rate of response during the MinMax (M = 119.187% MVC/Test Battery, SD = 158.260, p = 0.004) condition when compared to On/Off.

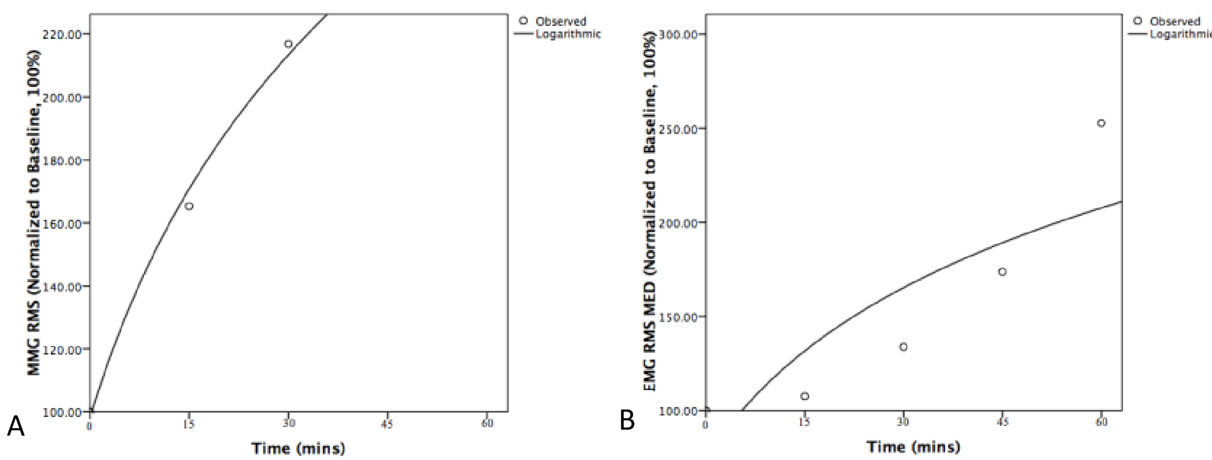


Figure 5.9 Typical Test Contraction MMG (A) and EMG RMS (B) Response During Exercise

RMS Preliminary Discussion

As with the previous chapter, studying the differences between sustained isometric and the classical intermittent contraction, MMG RMS increased in MinMax, 1 Percent, and Sinusoidal conditions. The significantly slower rate of response in the 1 Percent, MinMax, and Sinusoidal conditions when compared to sustained isometric may imply that intermittent contractions, independent on the amplitude diversity, will result in delayed fatigue response. As such, there was no difference between the classical intermittent contraction (On/Off) and the Sinusoidal and MinMax conditions. Based on test contractions, 1 Percent and Sinusoidal conditions were significantly slower in rate of response than sustained isometric. MinMax, however, was not significantly slower than sustained isometric but was significantly quicker than On/Off.

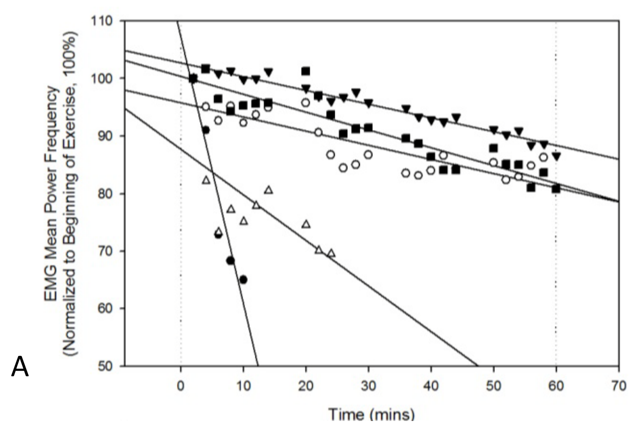
MMG RMS recovery, however, did not show any differences between conditions. Based on rates of responses observed via continuous measurement and test battery, it is clear that Sinusoidal and 1 Percent were clearly slower than sustained isometric. MinMax (+/- ½ amplitude) is less clear as there is a discrepancy between the two time intervals of analysis. This may indicate that MinMax, based on MMG RMS, is "halfway" between On/Off and sustained isometric. Differences between test contraction and continuous measurements were discussed in Chapter IV.

An apparent contradiction between the interpretation of MMG RMS increase and proposed mechanisms of intermittent contractions may exist. According to Shinohara and Søgaard (2006) and Vested and colleagues (2006), MMG may reflect the underlying cross-bridge cycling mechanism. However, past research have suggested an increase in cross-bridge turnover, and hence total energy consumption, is associated with intermittent work (Chasiotis et al., 1987; Vøllestad et al., 1997). The results from this study may show that MMG may not reflect cross-bridge cycling. Alternatively, results may also show, but less convincingly, that intermittent contractions do not lead to increased cross-bridge turnover rate.

Continuous EMG RMS of both medial and lateral heads revealed similar trends to MMG, where there was a slower rate of EMG amplitude increase in 1 Percent, MinMax, and Sinusoidal conditions. The test contractions revealed a significantly higher rate of RMS increase during the MinMax condition when compared to On/Off in the lateral head. EMG RMS response in the medial head showed similar response to the lateral head, where the MinMax condition led to a quicker rate of response. 1 Percent led to a slower rate of response compared to sustained isometric in the medial head. These results may provide further evidence that MinMax may be in the middle of the spectrum between sustained isometric and the classical intermittent pattern as there were significant differences between both polar "opposites". Surprisingly there was a small negative mean slope in the Sinusoidal condition during continuous collection and 1 Percent during test contractions, both in EMG analysis of the triceps medial head. This may be due to "muscle wisdom" as described in chapter IV where a decreased firing rate may reduce EMG amplitude.

Continuous Measures MnPF and MdPF Results

MMG MnPF continuous data was fitted with linear regression with goodness of fit values of $r^2 = 0.235$ (1 Percent), $r^2 = 0.286$ (MinMax), $r^2 = 0.361$ (Sinusoidal). Linear regression best-fit lines for MMG MdPF were $r^2 = 0.208$ (1 Percent), $r^2 = 0.306$ (MinMax), $r^2 = 0.305$ (Sinusoidal). EMG of the medial head had goodness of fit values of $r^2 = 0.529$ (1 Percent MnPF), $r^2 = 0.537$ (MinMax MnPF), $r^2 = 0.511$ (Sinusoidal MnPF), $r^2 = 0.363$ (1 Percent MdPF), $r^2 = 0.394$ (MinMax MdPF), $r^2 = 0.383$ (Sinusoidal MdPF). Finally EMG of the lateral head were fitted with linear regression with goodness of fit of $r^2 = 0.473$ (1 Percent MnPF), $r^2 = 0.496$ (MinMax MnPF), $r^2 = 0.386$ (Sinusoidal MnPF), $r^2 = 0.372$ (1 Percent MdPF), $r^2 = 0.462$ (MinMax MdPF), and $r^2 = 0.297$ (Sinusoidal MdPF). Typical linear regression fits are shown in Figure 5.10.



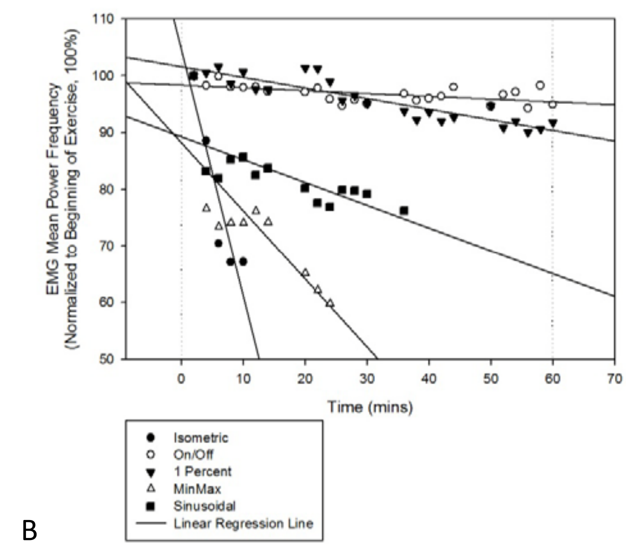


Figure 5.10 Typical MnPG Continuous MMG and EMG Medial (A) and Lateral (B) Heads for 5 Conditions

MMG MnPF analysis revealed no differences between conditions, as there were no statistically significant differences between the three varied intermittent contractions and sustained isometric or On/Off. Similar results were found using MMG median power frequencies. EMG of the medial head revealed a slower rate of response during the Sinusoidal condition (M = -2.539%/min, SD = 10.289, $p = 0.026$) than the sustained isometric contraction. Mean power frequencies of the medial head also had identical trends. A slower rate of MMG shift to lower frequencies was identified during the Sinusoidal condition (M = -0.878%/min, SD = 7.037, $p = 0.023$) when compared to sustained isometric. Sinusoidal condition (M = -2.229%/min, SD = 8.707, $p = 0.044$) also led to slower rate of EMG MnPF decrease, relative to sustained isometric, in the lateral head. There were no differences, however, in lateral head EMG median power frequencies.

Test Contractions MnPF and MdPF Results

Test contractions consisting of a 15% MVF were analyzed in its frequency domain. MMG and EMG MnPF and MdPF were assessed and fitted with logarithmic lines. The goodness of fit was as follows: r^2 = 0.585 (1 Percent MMG MnPF), r^2 = 0.776 (MinMax MMG MnPF), r^2 = 0.598 (Sinusoidal MMG MnPF), r^2 = 0.648 (1 Percent MMG MdPF), r^2 = 0.713 (MinMax MMG MdPF), r^2 = 0.578 (Sinusoidal MMG MdPF), r^2 = 0.645 (1 Percent EMG Med MnPF), r^2 = 0.730 (MinMax EMG Med MnPF), r^2 = 0.707 (Sinusoidal EMG Med MnPF), r^2 = 0.619 (1 Percent EMG Med MdPF), r^2 = 0.759 (MinMax EMG Med MdPF), r^2 = 0.713 (Sinusoidal EMG Med MdPF), r^2 = 0.749 (1 Percent EMG Lat MnPF), r^2 = 0.802 (MinMax EMG Lat MnPF), r^2 = 0.832 (Sinusoidal EMG Lat MnPF), r^2 = 0.671 (1 Percent EMG Lat MdPF), r^2 = 0.751 (MinMax EMG Lat MdPF), r^2 = 0.784 (Sinusoidal EMG Lat MdPF). Typical regression fits are shown in Figure 5.11.

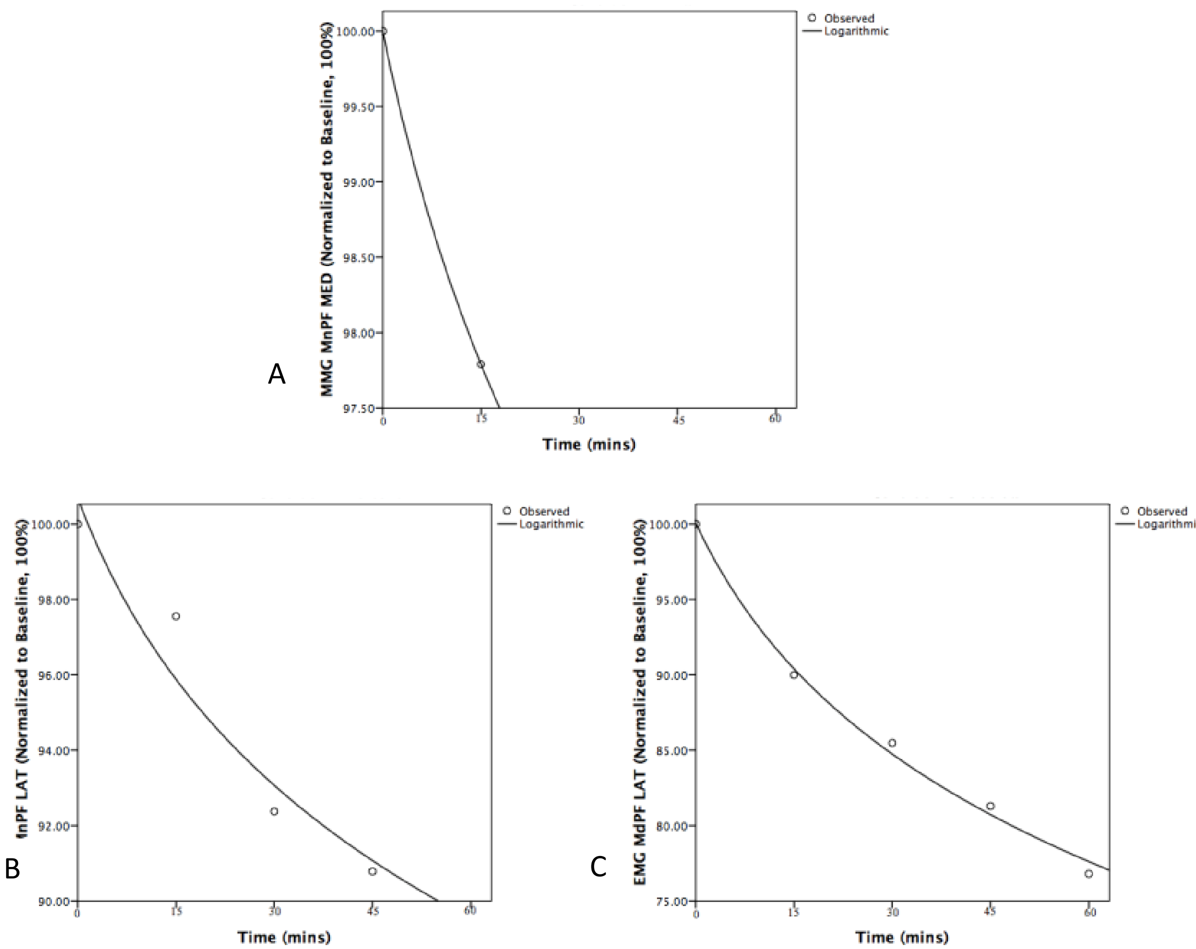


Figure 5.11 Typical Test Contraction MMG (A) and EMG MnPF (B) and MdPF (C) Responses During Exercise for Medial and Lateral Heads of Triceps Brachii – MinMax, 1 Percent, Sinusoidal Conditions

In both mean and median power frequencies of test contraction MMG there were no differences between the three intermittent conditions and the sustained isometric and On/Off efforts. Similar results were observed in EMG frequency analysis of the lateral head. Medial head EMG mean power spectrum analysis, however, found that MinMax (M = -18.090%/Test Battery, SD = 17.223) led to a slower rate of shift to lower frequencies compared to sustained isometric (p = 0.010) and a quicker rate compared to On/Off (p = 0.001). Median power frequencies of the medial head also showed a similar trend. MinMax (M = -12.457%/Test Battery, SD = 10.922) led to a slower rate of decreasing frequency values than sustained isometric (p = 0.028) and a quicker rate compared to On/Off (p = 0.010).

MnPF and MdPF Preliminary Discussion

Continuous MMG power spectrum analysis in both MinMax and Sinusoidal conditions were consistent with previous literature that suggested a shift towards lower frequencies as a consequence to fatigue (Weir et al., 2000). However, the 1 Percent condition revealed positive rate of response (shift towards higher frequencies). Yoshitake and colleagues (2001) demonstrated a nearly constant mean power frequency concurrent with an increase in MMG RMS. It was argued that if MMG represents the firing rate of motor units, the exercise protocol led to motor units that were firing at a constant rate. There were no differences between conditions, which may suggest that the rate of fatigue response did not differ between mechanical exposure amplitude patterns or MMG frequency analysis is not a reliable and valid measure to determine fatigue response.

Test contraction analysis in MMG frequency domain reveals similar trends. There were no differences between conditions but there were increasing rate of frequency response in the MinMax and Sinusoidal condition. This is an apparent contradiction to the continuous measurement responses that found decreasing values in MinMax and Sinusoidal contraction patterns. In addition to the explanation between test contraction and continuous measures discussed in chapter IV, this may be additional support to the stochastic nature of frequency analysis in MMG.

Power spectrum analysis in EMG shared similar responses that were consistent with past literature (Hagberg, 1981; Jørgensen et al., 1988; Weir et al., 2000; Ebersole et al., 2006). The intermittent patterns of varying mechanical force led to smaller shifts towards lower frequencies compared to the sustained isometric condition. Continuous measures of EMG in both medial and lateral heads reveal a significant slower rate of frequency decrease during the Sinusoidal condition in mean power frequency but were only significant in the lateral head when based on median power frequency. Test contractions led to different trends. Lateral head EMG power analysis revealed no differences between conditions. On the other hand, MinMax was significantly different from sustained isometric and On/Off in medial head frequency domain analysis. This may provide further evidence that in the medial head, MinMax rate of fatigue response is between sustained isometric and On/Off conditions. Additionally, there is further evidence that Sinusoidal contractions may delay fatigue response when compared to a sustained submaximal effort.

EMG Hi-Lo Results

EMG Hi-Lo ratios were collected continuously and sampled every 2 minutes. The mean ratio identified in the 30-second window was plotted against time and were fit with logarithmic regression (Figure 5.12). The 1 Percent achieved goodness of fit values of $r^2 = 0.431$ (medial head) and $r^2 = 0.357$ (lateral head). MinMax had best-fit lines of $r^2 = 0.419$ (medial head) and $r^2 = 0.448$ (lateral head). Sinusoidal were fitted with lines $r^2 = 0.411$ (medial head) and $r^2 = 0.355$ (lateral head).

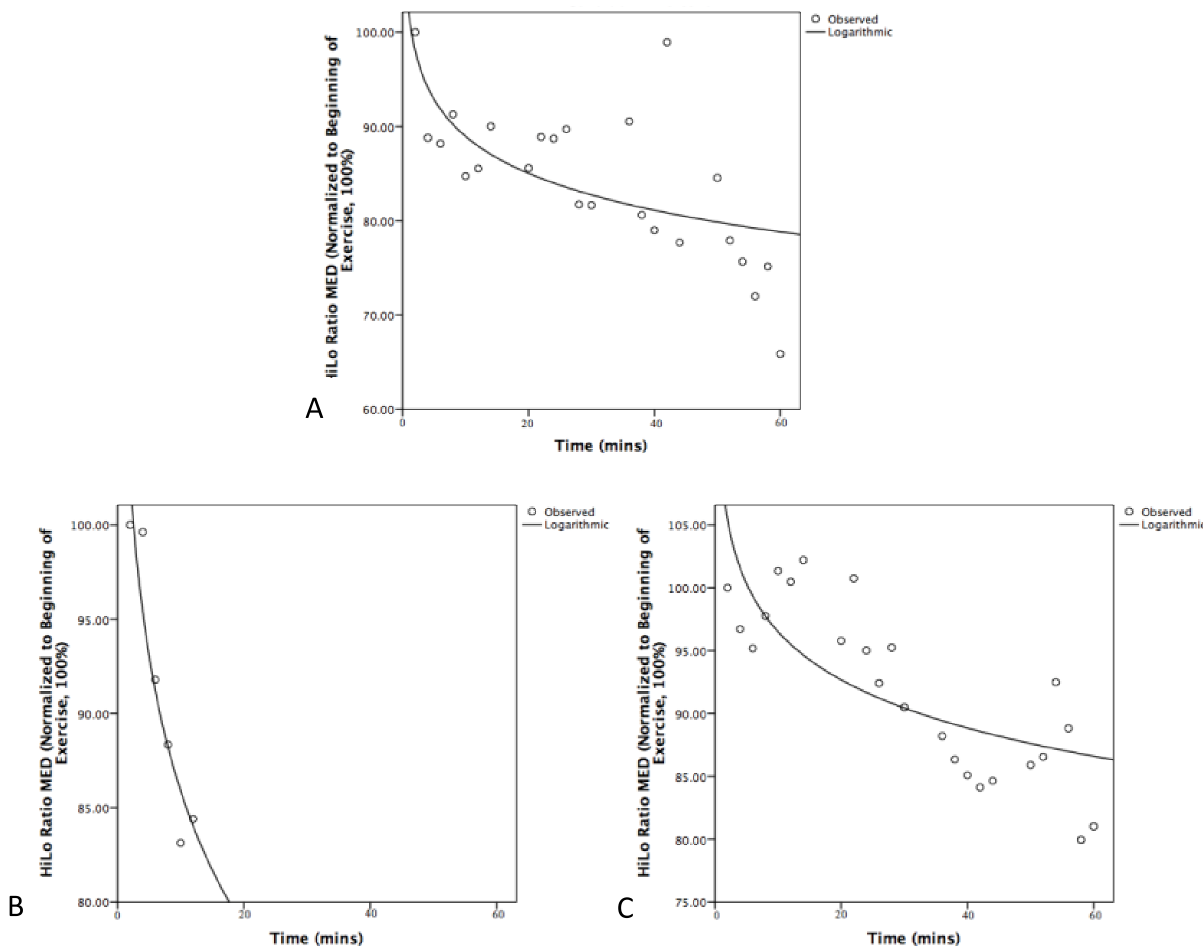


Figure 5.12 Typical Continuous EMG Hi-Lo Response During Exercise for 1 Percent (A), MinMax (B), and Sinusoidal (C) in Medial and Lateral Heads of Triceps Brachii

EMG Hi-Lo ratios of the medial head revealed slower rate of fatigue response in the 1 Percent (M = -3.628%/min, SD = 9.719, p = 0.002), MinMax (M = -7.166%/min, SD = 7.632, p = 0.034), and Sinusoidal (M = -2.304%/min, SD = 9.016, p = 0.001) conditions when compared to the sustained isometric effort. There were no differences between conditions, however, observed in the lateral head.

EMG Hi-Lo Ratio Preliminary Discussion

Hi-Lo ratio responses during 1 Percent, MinMax, and Sinusoidal were consistent with previous observations that reported a decline in values over time (Bigland-Ritchie et al., 1981; Moxham et al., 1982; Allison & Fujiwara, 2002). As suggested in Chapter IV, a lack of ratio reduction may be due to substantial and persistent low-frequency fatigue (Moxham et al., 1982). The 1 Percent, MinMax, and Sinusoidal conditions had moderate decreases in Hi-Lo ratios, when compared to sustained isometric, in both medial and lateral heads. However, only medial head revealed a large discrepancy in rate of response between the intermittent contractions and sustained isometric. Along with mean and median power frequency results, the lateral head may have not led to the same extent of fatigue as the medial head. Unlike mean and median power frequency results, Hi-Lo ratios may be more sensitive to fatigue, particularly at lower levels of force (Allison & Fujiwara, 2002).

Low-Frequency Fatigue Results

Using electrical muscle stimulations at 20 Hz and 100 Hz, low frequency fatigue was assessed during exercise and during recovery (Figures 5.13). Low-frequency fatigue ratio response was fitted with linear regression and had best-fit values of $r^2 = 0.623$ (1 Percent), $r^2 = 0.776$ (MinMax), and $r^2 = 0.665$ (Sinusoidal). Recovery was also fitted with linear regression and had best-fit values of $r^2 = 0.302$ (1 Percent), $r^2 = 0.284$ (MinMax), $r^2 = 0.316$ (Sinusoidal). During exercise, the rate of low-frequency fatigue decline was slower in the Sinusoidal condition (M = 6.556%/Test Battery, SD = 42.021, p = 0.014) when compared to the 15% MVF sustained isometric exertion. There were no differences between conditions during recovery.

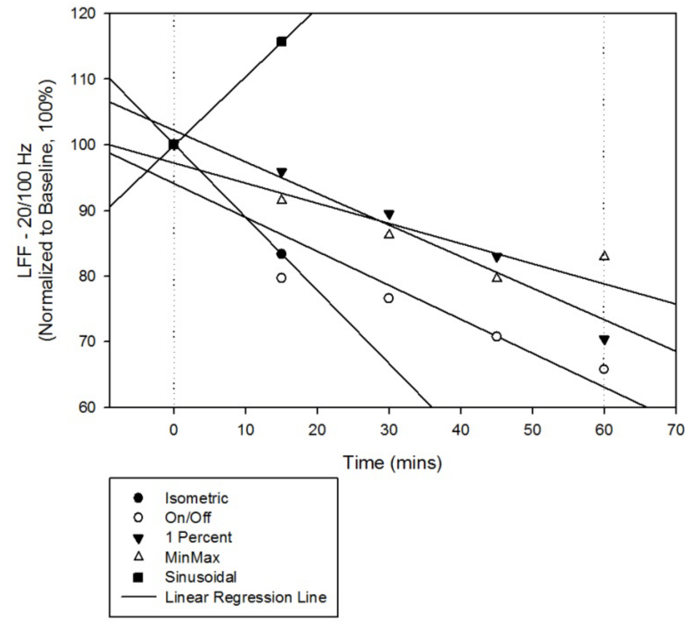


Figure 5.13 Typical Low-Frequency Fatigue Ratio Response During Exercise for 5 Conditions

Cessation and recovery at 15-minute intervals and 24 hours post exercise were compared to baseline low-frequency fatigue values (Figure 5.14). After the 1 Percent exercise condition, LFF values remained depressed up to 60 minutes of activity, $F(0,6) = 5.127$, $p = 0.000$, $\eta_p^2 = 0.299$. The MinMax condition, on the other hand, did not lead to a significant reduction in LFF at cessation nor during recovery, $F(0,6) = 2.339$, $p = 0.054$, $\eta_p^2 = 0.163$. Sinusoidal condition was similar to MinMax. There was no significant decline in low-frequency fatigue ratio at cessation or during recovery, $F(0,6) = 0.926$, $p = 0.481$, $\eta_p^2 = 0.072$.

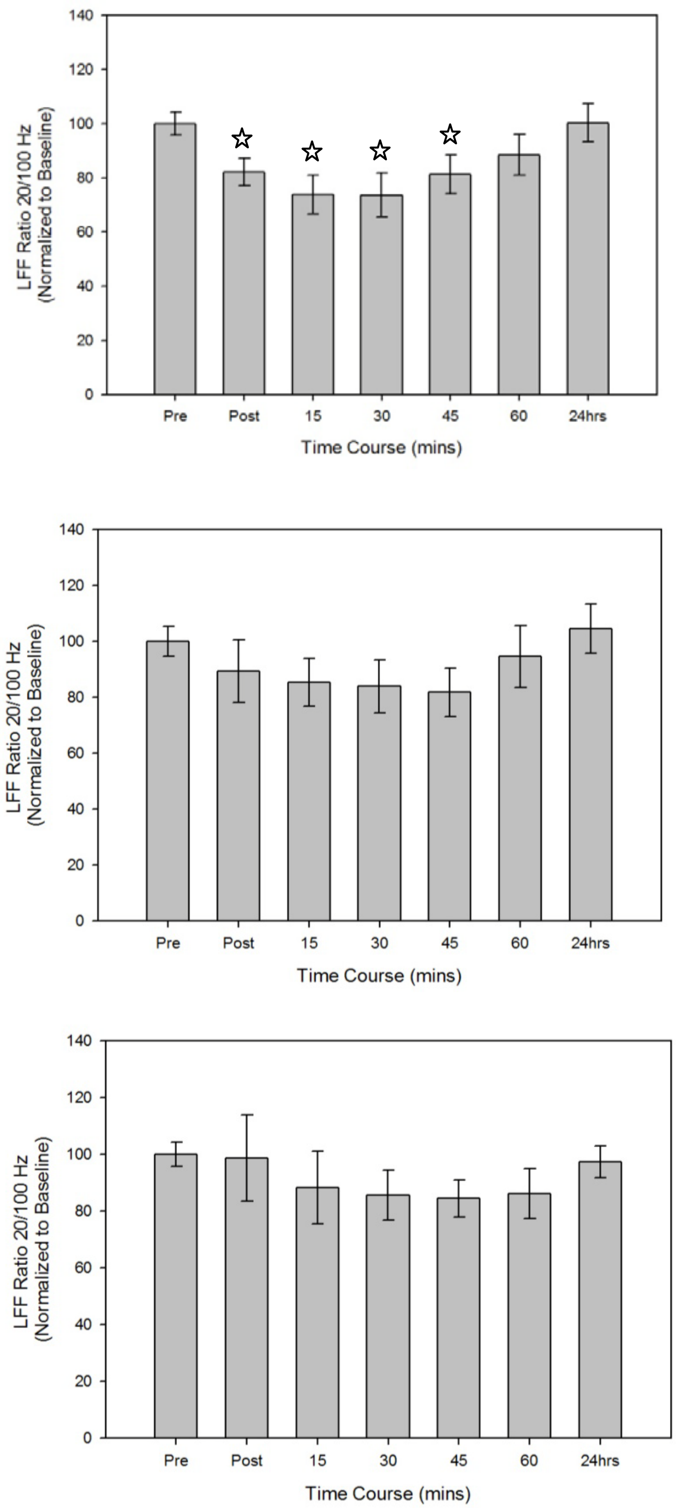


Figure 5.14 Low Frequency Fatigue Comparisons Between Baseline & Recovery for 1 Percent (A), MinMax (B), and Sinusoidal (C).

Low-Frequency Fatigue Preliminary Discussion

A decrease in low-frequency fatigue was observed in the 1 Percent condition but a concomitant increase in Sinusoidal. MinMax resulted in no change at all. These rates of responses are supported by cessation values where 1 Percent led to depressed LFF ratio while Sinusoidal and MinMax led to no differences between pre and post-exercise. The decrease in LFF agreed with previous findings from Bystrom and Fransson-Hall (1994), Vøllestad et al. (1997), and Griffin and Anderson (2008). An increase in LFF may possibly be best explained by the role of postactivation potentiation (PAP) or activity-dependent potentiation. As described by Sale (2004), there may be a disproportionate increase in low frequency tetanic force due to a conditioning activity, such as a series of repetitive dynamic contractions. Postactivation potentiation is the phosphorylation of myosin regulatory light chains that increases Ca^{2+} sensitivity of the myofilaments, resulting in greater myosin cross bridge activity (Rijkelijkhuizen et al., 2005). Sale (2004) argued that when performing a series of submaximal contractions, the contractions themselves might have a cumulative effect in mobilizing PAP mechanisms. Rijkelijkhuizen and colleagues (2005) found an increase in LFF post-exercise, possibly due to either potentiation and/or increased muscle lengths.

Recovery values remained significantly depressed after the 1 Percent condition, upwards to 60 minutes post-exercise. This is similar to previous findings where LFF was present over a prolonged period of recovery. Similar to the On/Off classical intermittent contraction, the lowest LFF ratio value was observed after cessation, within the first few intervals of recovery. The MinMax and Sinusoidal conditions did not result in a decrease in LFF at cessation or recovery, perhaps due to potentiation.

Blood Flow Velocity Results

Mean blood velocity was collected every 2 minutes for a 30-second window and normalized to baseline (100%). The mean of these measures was calculated from the beginning of exercise (2 minute interval). 1 Percent led to a mean blood flow velocity of 121.85% (SD = 46.496), which was significantly higher than sustained isometric ($p = 0.001$). Sinusoidal (M = 121.28%, SD = 93.169) also led to a higher mean blood flow velocity when compared to sustained isometric ($p = 0.002$) and was also significantly lower than On/Off. There was no difference between MinMax ($M = 107.78\%$, $SD = 88.025$) and sustained isometric ($p = 0.072$). However, MinMax had a significantly lower mean blood flow velocity when compared to On/Off ($p = 0.000$). Mean blood flow velocity for all conditions is shown in Figure 5.15

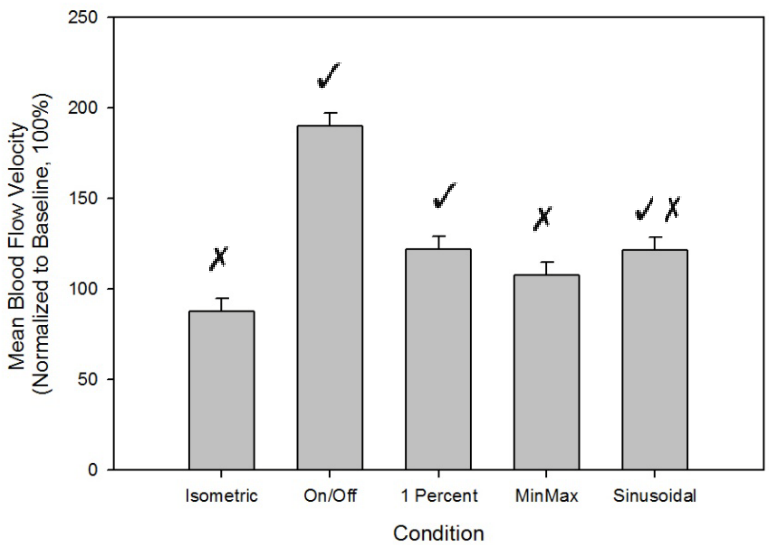


Figure 5.15 Mean Blood Velocity Comparisons Between Baseline & Recovery for 5 Conditions

Blood Flow Velocity Preliminary Discussion

It was observed in this study that MinMax led to a mean blood flow velocity that was similar to sustained isometric. An increase in blood velocity occurred during the initial few minutes of exercise and a decrease towards exhaustion. 1 Percent was significantly higher than sustained isometric, leading to the belief that there was a steady increase in blood flow with a plateau towards the end of exercise, similar to the classical intermittent contraction. Hughson and colleagues (1996) demonstrated a rapid increase in blood flow within the first contraction/relaxation due to the activation of the muscle pump. After the initial increase, Williams and colleagues (1985) reported constant blood flow despite a large increase in mean arterial blood pressure. Sinusoidal had mean blood flow velocities that were higher than sustained isometric yet lower than On/Off. Possibly Sinusoidal, based on blood flow velocity, had a response that was intermediate of On/Off and sustained isometric.

As with the previous chapter, this study showed an increase in blood flow during the various intermittent contractions. During muscular exercise, there is a compromise between blood vessel dilation, compression of the local vessels by contracting muscle, and sympathetic vasoconstrictor tone (Williams et al., 1985). Past research have shown little change in muscle sympathetic nerve activity after both static contractions at 15% MVC (Seals et al, 1988) and was not expected during intermittent contractions (Hughson et al., 1996). However, studies have shown that metabolically induced dilation might adjust blood flow as a result of demand (Hughson et al., 1996). The gradual rise in blood velocity in all intermittent conditions may imply lower metabolic demand compared to sustained isometric but an increase demand over time. It is also possible that intermittent contractions allow greater blood perfusion into the muscle as compression of the artery is reduced.

Ratings of Perceived Exertion Results

Perceived exertion was measured every 2 minutes during exercise and later fitted with linear regression (1 Percent $r^2 = 0.703$, MinMax $r^2 = 0.661$, Sinusoidal $r^2 = 0.690$; Figure 5.16). The three various intermittent contractions [1 Percent ($M = 6.501\%/min$, $SD = 7.756$, $p = 0.000$), MinMax ($M = 10.853\%/min$, $SD = 9.446$, $p = 0.003$), Sinusoidal ($M = 9.484\%/min$, $SD = 11.108$, $p = 0.001$)] led to significantly slower rates of RPE increase than sustained isometric. MinMax ($p = 0.022$) led to a quicker increase in ratings of perceived exertion when compared to On/Off.

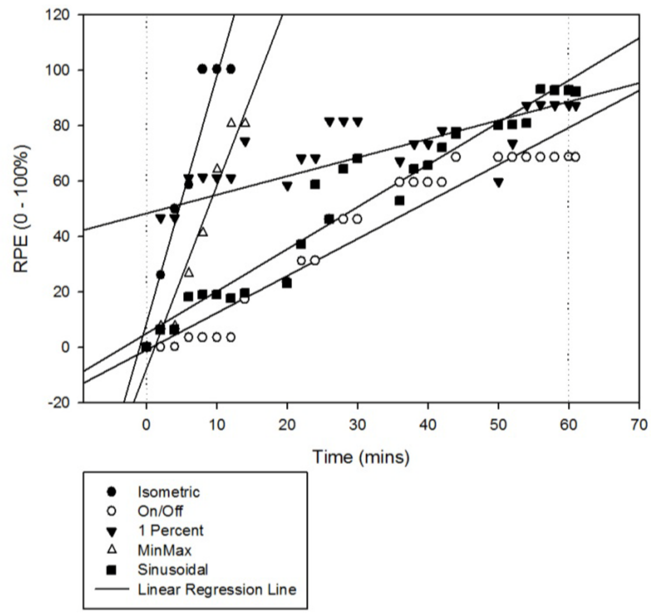


Figure 5.16 Typical Continuous RPE Response During Exercise for 5 Conditions

Ratings of Perceived Exertion Preliminary Discussion

An increase in ratings of perceived exertion was observed for all conditions. A quick increase in RPE rate of increase was found in the MinMax condition when compared to the classical intermittent contraction pattern. MinMax was also significantly slower than the sustained isometric condition, providing further evidence that this intermittent contraction at 7.5% and 22.5% MVF is between On/Off and sustained isometric. Sinusoidal and 1 Percent conditions resulted in significantly slower rates of RPE increase, relative to sustained isometric, and were similar to On/Off. Described in the previous chapter, perceived exertion might relate to impaired endurance capacity (Jones & Killian, 2000). In this study, there was an inverse relationship between rate of RPE increase and endurance time. A quicker rate of RPE led to shorter endurance time.

Heart Rate Results

Heart rate was measured continuously for the entire duration of the exercise protocol and later analyzed at 2-minute intervals to obtain mean heart rate beats per minute. Logarithmic regression fits are shown in Figure 5.17 with goodness of fit values of $r^2 = 0.535$ (1 Percent), $r^2 = 0.526$ (MinMax), $r^2 = 0.389$ (Sinusoidal). There were no differences in the rate of heart rate response among the conditions.

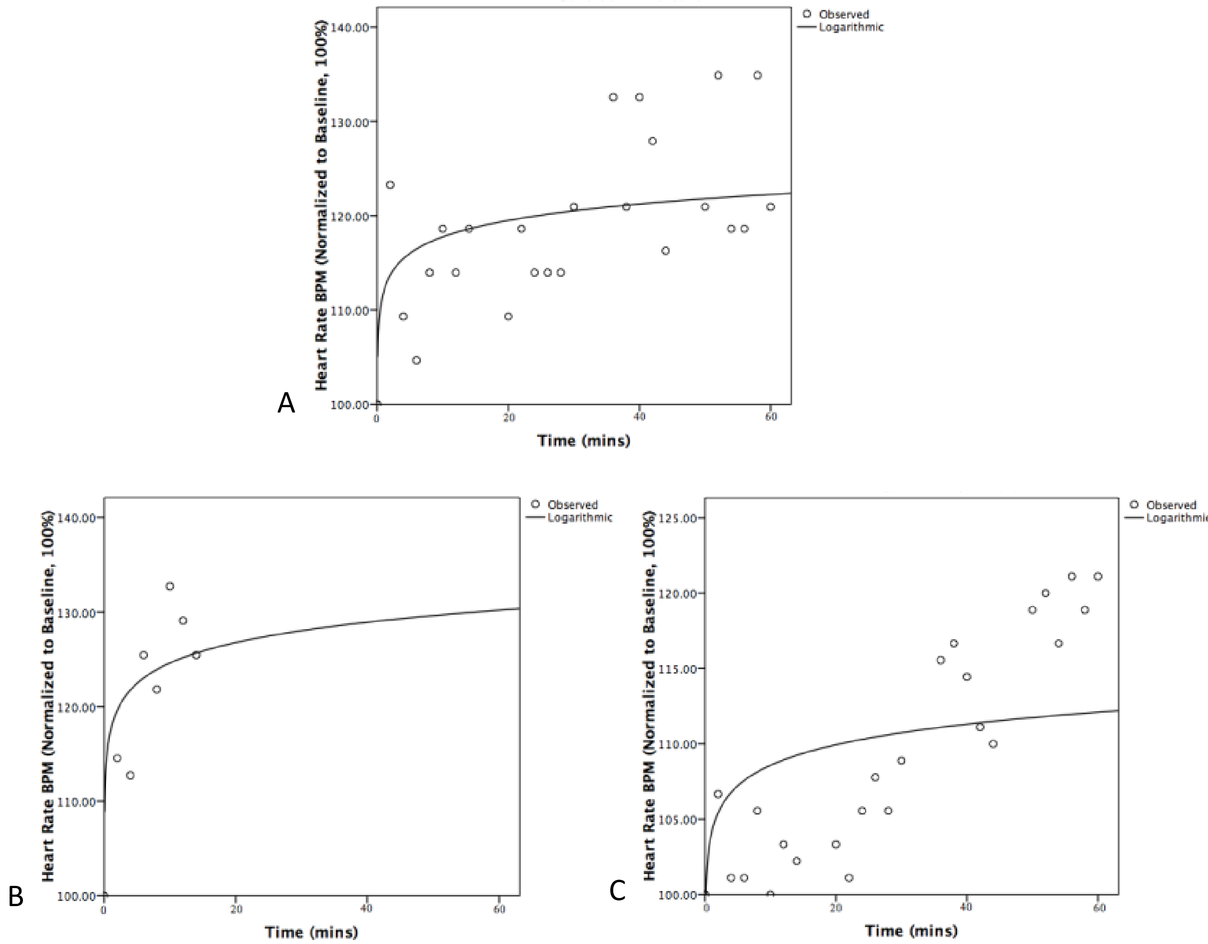


Figure 5.17 Typical Continuous Heart Rate Response During Exercise for 1 Percent (A), MinMax (B) and Sinusoidal (C)

Heart Rate Preliminary Discussion

An increase in heart rate was observed in all conditions, consistent with previous studies by Bystrom and Fransson-Hall (1994), Fallentin and colleagues (1985), and Hunter and colleagues (2004). The increase in heart rate may be due to the magnitude of the force (15% MVF), the type II fibre dominancy of the triceps brachii muscle, and the duty cycle and cycle time of the exercise protocol. There were no differences, however, to distinguish conditions from sustained isometric and the intermittent contraction patterns.

EMG Gaps and Mechanical Force Results

Figure 5.18A (top panel) is an example of the window length (0.5 seconds) used to calculate mechanical force in the sinusoidal condition. Figure 5.18A (bottom panel) is a detailed view of the normalized EMG for lateral and medial heads of the triceps brachii used to calculate EMG gaps. Refer to Figure 5.18B for window selections in mechanical force output and EMG gaps analysis for the On/Off condition, identical to the procedure used in 1 Percent and MinMax.

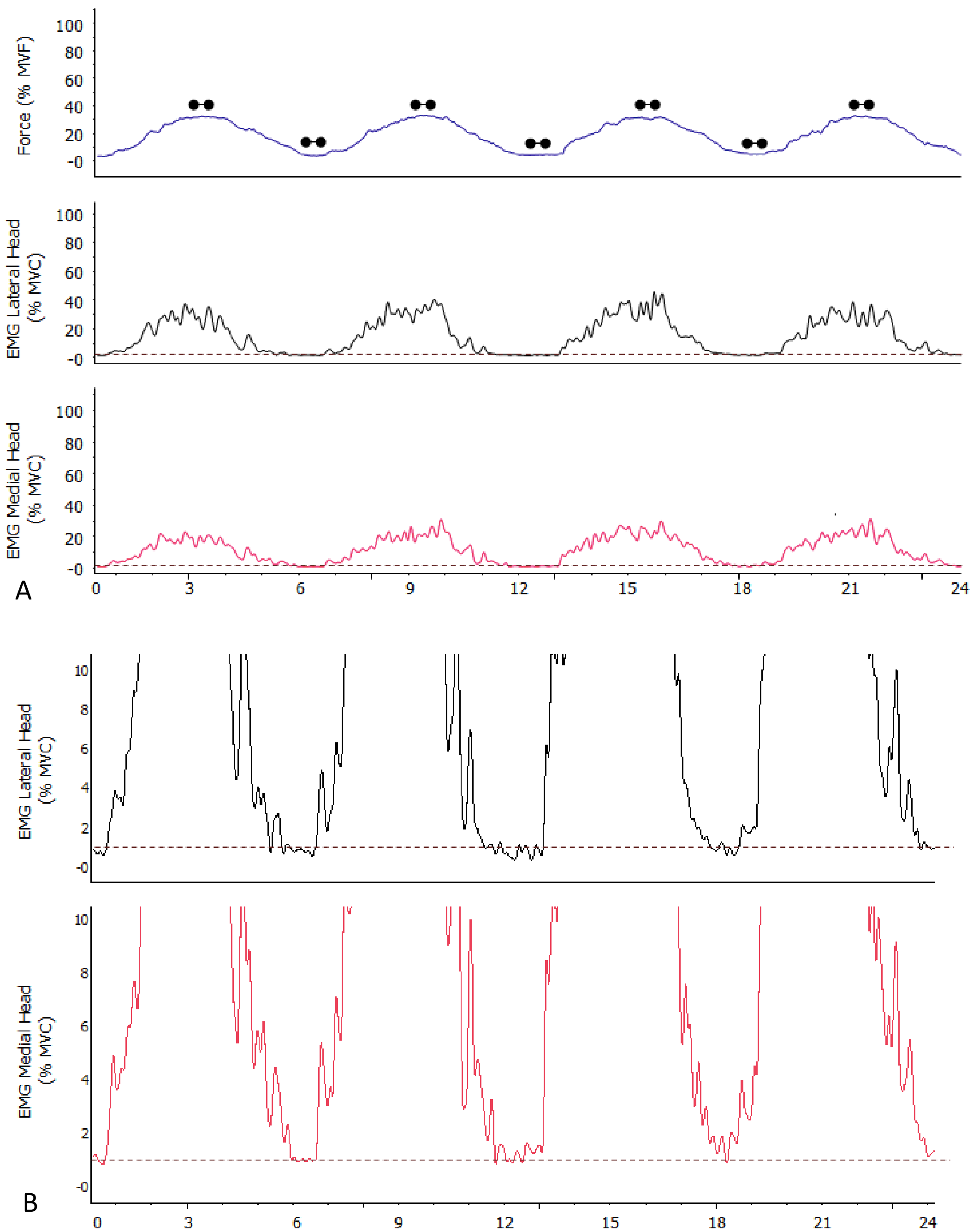
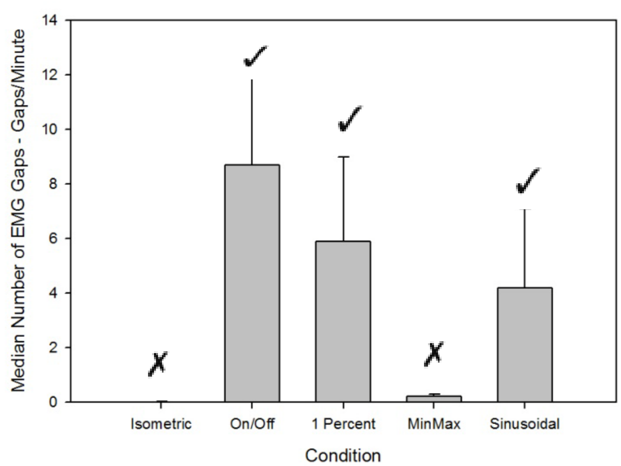
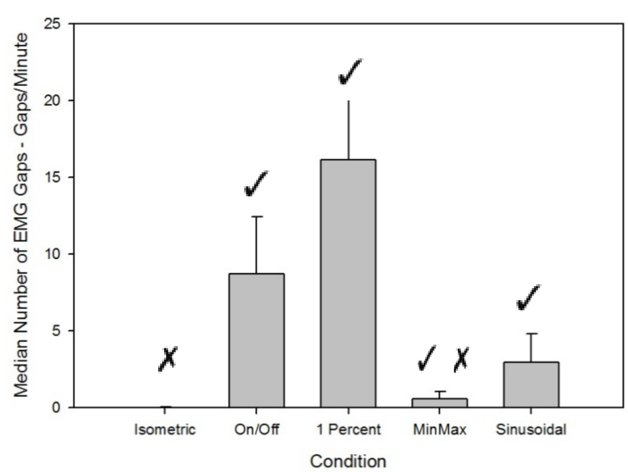


Figure 5.18 Mechanical Force Output (A) and EMG Gaps (A and B) in Sinusoidal Condition.
Windows of 0.5 seconds (double-ended dot bars) were used at low and high peaks to measure mechanical force output (5.18A top panel). A detailed view of normalized EMG lateral and medial heads of triceps brachii (5.18B) used to calculate EMG gaps (> 1% MVC, > 0.2 seconds). Dotted line is the 1% MVC EMG gaps threshold.

The number of EMG gaps (gaps/minute) for both medial and lateral heads of the triceps brachii was calculated over the entire exercise period for each condition (Figure 5.19). In the medial head, 1 Percent (Median = 16.133 gaps/min, 25th = 0.811, 75th = 33.134, p = 0.003), MinMax (Median = 0.550 gaps/min, 25th = 0, 75th = 1.842, p = 0.006), and Sinusoidal (Median = 2.933 gaps/min, 25th = 0.050, 75th = 10.877, p = 0.003) had more gaps per minute than sustained isometric. In comparison to the On/Off condition, MinMax (p = 0.004) had significantly fewer gaps. The lateral head showed similar trends with the exception of MinMax. There was greater number of EMG gaps in the 1 Percent (Median = 5.882 gaps/min, 25th = 0.085, 75th = 22.550, p = 0.002) and Sinusoidal (Median = 4.182 gaps/min, 25th = 0, 75th = 12.546, p = 0.007) than sustained isometric. MinMax (Median = 0.217 gaps/min, 25th = 0, 75th = 0.444, p = 0.002) had fewer EMG gaps than On/Off.



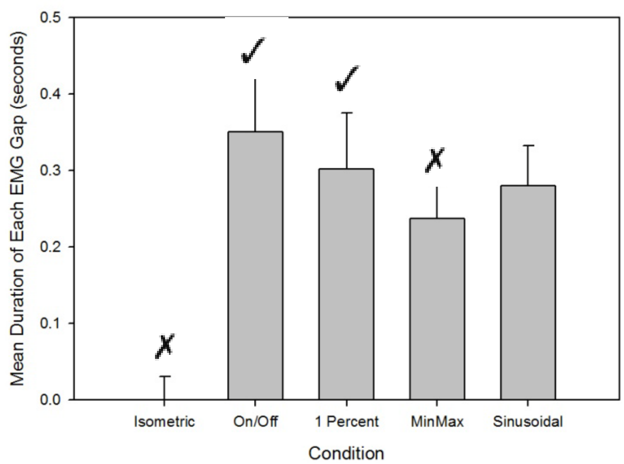
A



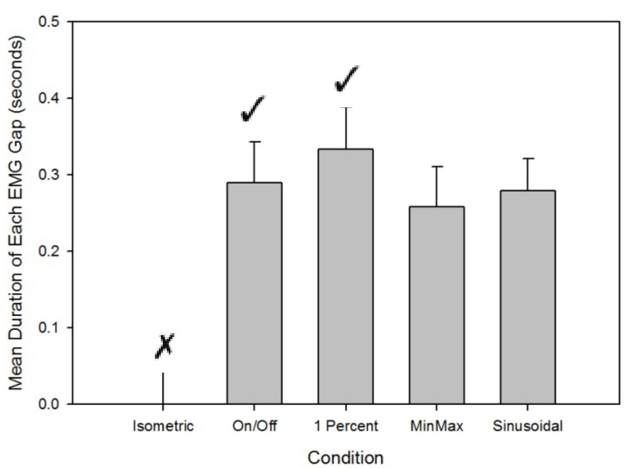
B

Figure 5.19 Median Number of Gaps (Per Minute) for Lateral (A) and Medial (B) Heads of Triceps Brachii

The mean (Figure 5.20) and median (Figure 5.21) duration per gap was also measured for both heads of the triceps brachii. In the medial head, 1 Percent (Median = 0.333 seconds, 25^{th} = 0.140, 75^{th} = 0.443, p = 0.007) had a longer mean duration per gap than sustained isometric. Similarly, 1 Percent (Median = 0.296 seconds, 25^{th} = 0.144, 75^{th} = 0.355, p = 0.007) led to a longer median duration than sustained isometric. The lateral head led to the same conclusion. The 1 Percent condition (Median = 0.302 seconds, 25^{th} = 0.209, 75^{th} = 0.407, p = 0.002) had a longer mean duration per gap than sustained isometric. Likewise, 1 Percent (Median = 0.267 seconds, 25^{th} = 0.209, 75^{th} = 0.334, p = 0.002) led to longer median duration per gap than sustained isometric. Additionally, in the lateral head, the MinMax condition (Median = 0.237 seconds, 25^{th} = 0, 75^{th} = 0.297, p = 0.004) had a shorter mean duration than On/Off. Based on median duration, MinMax (Median = 0.224, 25^{th} = 0, 75^{th} = 0.262, p = 0.002) led to shorter duration than On/Off.



A



B

Figure 5.20 Mean Duration of EMG Gaps (Per Gap) for the Lateral (A) and Medial (B) Heads of Triceps Brachii

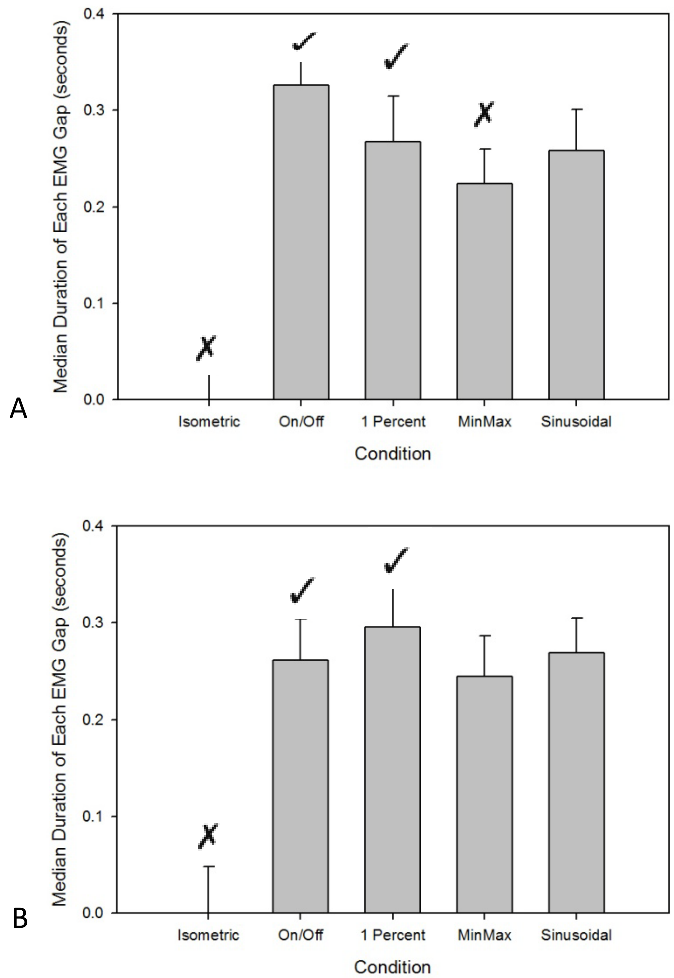
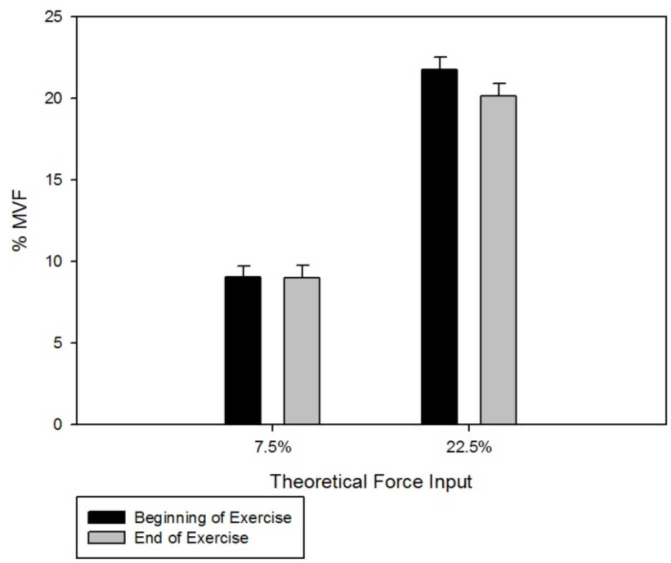


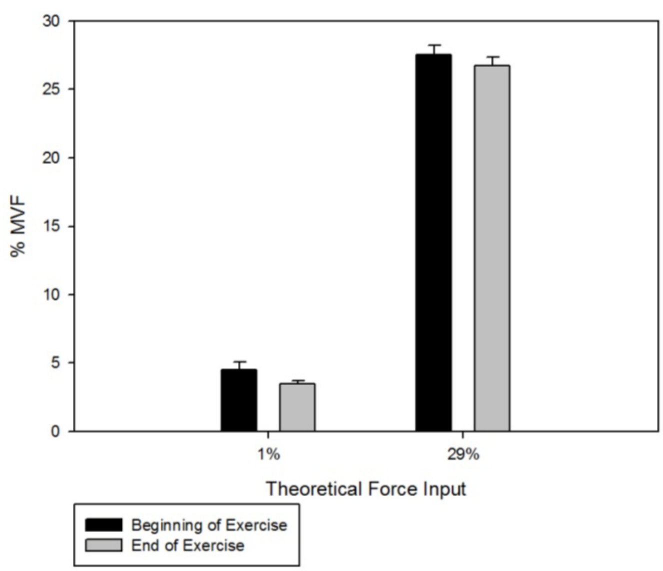
Figure 5.21 Median Duration of EMG Gaps (Per Gap) for Lateral (A) and Medial (B) Heads of Triceps Brachii

Mechanical force outputs were also measured for the entire exercise protocol, during the first 10 contractions at both levels of force at the beginning of exercise, and the last 10 contractions at both levels of force prior to exhaustion (Figure 5.22). Although the force inputs for the 1 Percent condition was 1% and 29% MVF, participants exerted a mean force of 5.82% MVF and 27.33% MVF for the entire duration of exercise. The first 10 contractions had a mean of 4.213% MVF and 27.528% MVF. Towards the end of exercise, participants exerted mean forces of 3.501% MVF and 26.717% MVF. There was a significantly lower force exertion at the theoretical 29% MVF during the end of exercise when compared to the beginning, $t(13) = 2.687, p = 0.020, d = 0.327$. MinMax led to a total mean force output of 9.44% MVF and 20.62% MVF. There was a difference, however, at the 22.5% MVF level as the end of exercise (M =

20.137% MVF, SD = 2.765) was lower than at the beginning of exercise (M = 21.736% MVF, SD = 1.881), $t(13) = 2.543, p = 0.026, d = 0.676$. Finally, Sinusoidal had mean total force output of 5.43% MVF and 28.44% MVF, representing the 0% and 30% levels. There was a significant decrease in force in both the lower (Beginning: M = 6.481% MVF, SD = 1.527; End: M = 4.420% MVF, SD = 1.410; $t(13) = 5.521, p = 0.000, d = 1.402$) and higher (Beginning: M = 29.018% MVF, SD = 2.389; End: M = 27.126% MVF, SD = 3.779; $t(13) = 2.681, p = 0.020, d = 0.598$) levels of force.



A



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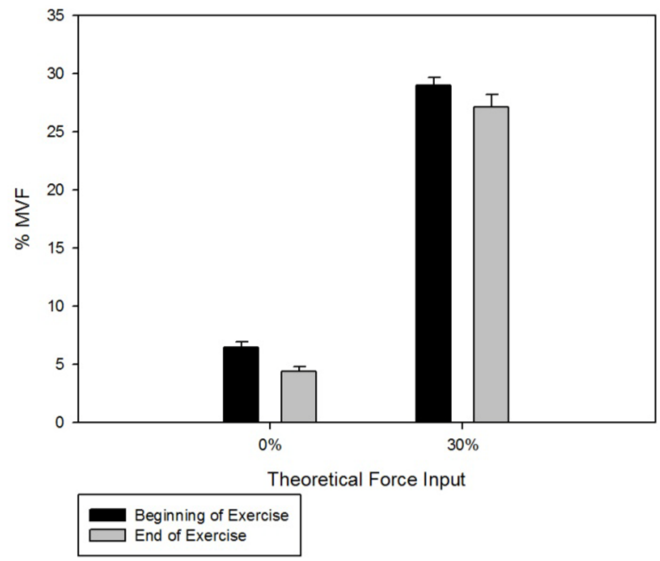

C

Figure 5.22 Mechanical Force Outputs for Varying Conditions Based on Beginning and End of Exercise for MinMax (A), 1 Percent (B), and Sinusoidal (C) Conditions.

EMG Gaps and Mechanical Force Preliminary Discussion

Gaps analysis was conducted in each of the three conditions for both medial and lateral heads of the triceps. The number of gaps per minute and gap duration was consistent with findings from Veiersted and colleagues (1990) who found EMG gaps between 0.7 and 20 per minute and durations between 0.15 – 7.3 seconds per minute. As expected, there were a greater number of gaps in all conditions when compared to the sustained isometric effort at 15% MVF. In both lateral and medial heads, 1 Percent and Sinusoidal led to a significantly greater number of EMG gaps per minute. MinMax had significantly fewer gaps per minute in both medial and lateral heads relative to On/Off. In theory, Sinusoidal should have approximately 10 gaps per minute, MinMax should have zero, and 1 Percent should have no gaps as well. However, this was not the case. In fact, Sinusoidal had 4.182 gaps/minute and 2.933 gaps/minute for lateral and medial heads, respectively. This can be explained by the mechanical force output where the lower force level was 6.481% MVF and 4.420% MVF at the beginning and end of exercise. This may be due to the delayed relaxation of motor units and contraction (and motor unit derecruitment) efficiency as a result of different patterns of motor unit excitation. A high target force level in combination with the duration of the contraction may decrease muscle contractile half relaxation time ($RT_{1/2}$) due to changes in high-energy substrate and metabolite concentrations (Vøllestad et al., 1997). A reduction in $RT_{1/2}$ may prevent the muscle from fully relaxing during the 0.5-second peak, and is reflected by increased low-force levels for a given contraction. This is further evidenced by the mean and median durations of each EMG gap where both medial and lateral heads were less than 0.3 seconds. Consequently the mean of 0.5 seconds, representing sinusoidal peak forces, had mechanical forces greater than 0% MVF and subsequently fewer EMG gaps than expected. The 1 Percent condition, in contrast, led to more EMG gaps than expected. Based on force inputs, it was expected that 1 Percent had no EMG gaps. This may be explained by the ability (or lack thereof) to discriminate against different forces, particularly at levels that require great precision and changes all the time (de Graaf et al., 2004). Gaining precision awareness to reproduce muscular forces is a demanding task (de Graaf et al., 2004). However, the total mean force at the lower level in the 1 Percent condition was approximately 1.5% MVF greater than On/Off. MinMax led to a small number of EMG gaps per minute, as expected. Small gaps may have occurred during brief and sporadic rest periods within the exercise protocol to contribute to the quantity of gaps.

An interesting trend occurred over time with respect to mechanical force and subsequent EMG gaps. 1 Percent and MinMax led to slight decline while Sinusoidal led to a significant decrease in mechanical force output when comparing the beginning and end of exercise. This may be a consequence of fatigue itself. According to Vøllestad and colleagues (1997), repetitive low-force isometric contractions induced a reduction in $RT_{1/2}$. This reduction may be due to increased turnover rate of the SR Ca^{2+}- ATPase or myosin ATPase (Vøllestad et al., 1997). A decrease in relaxation time will increase relaxation rate, and hence lowered measured force, but at the consequence of an increase in force oscillations and energy cost of each contraction. A decrease in force towards the end of exercise was also observed for the high level of force. This was true for all three intermittent contractions where all conditions were significantly lower than the beginning of exercise. This too may be a consequence of fatigue as force capacity diminished over time. Time-dependent EMG gap analysis was not reported in this study but initial trends demonstrate an increase in EMG gaps for every successive 15-minute interval. The increase in EMG gaps may be a consequence of decreased relaxation time, muscle "wisdom", or a compensatory mechanism to reduce motor unit overloading.

General Results

The three intermittent isometric contractions of varying amplitudes (1 Percent, MinMax, Sinusoidal) were plotted against sustained isometric and On/Off conditions for measurements during continuous exercise (Figure 5.23) and test batteries (Figure 5.24). A summary of the statistically significant parameter responses between all conditions is shown in Table 5.1.

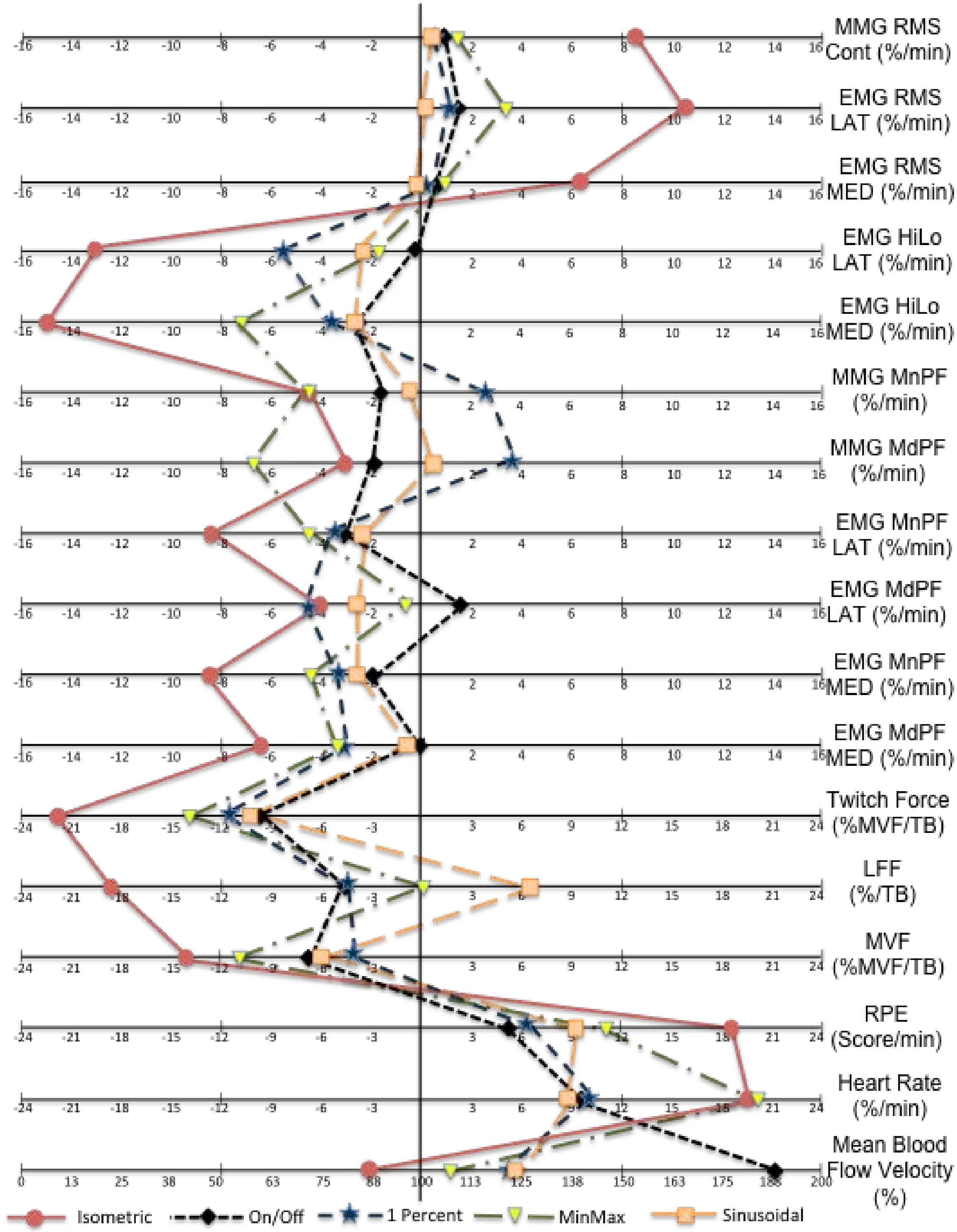

Figure 5.23 Summary of Mean Values for Continuous Responses Between Sustained Isometric, On/Off, 1 Percent, MinMax, and Sinusoidal Conditions.

Refer to preliminary analysis to the direction that leads to a greater "fatigue response". In general, increasing amplitude parameters (RMS), ratings of perceived exertion, and heart rate, indicate an "exhaustive" response. Remaining parameters indicate a "fatiguing" response by larger decreasing values.

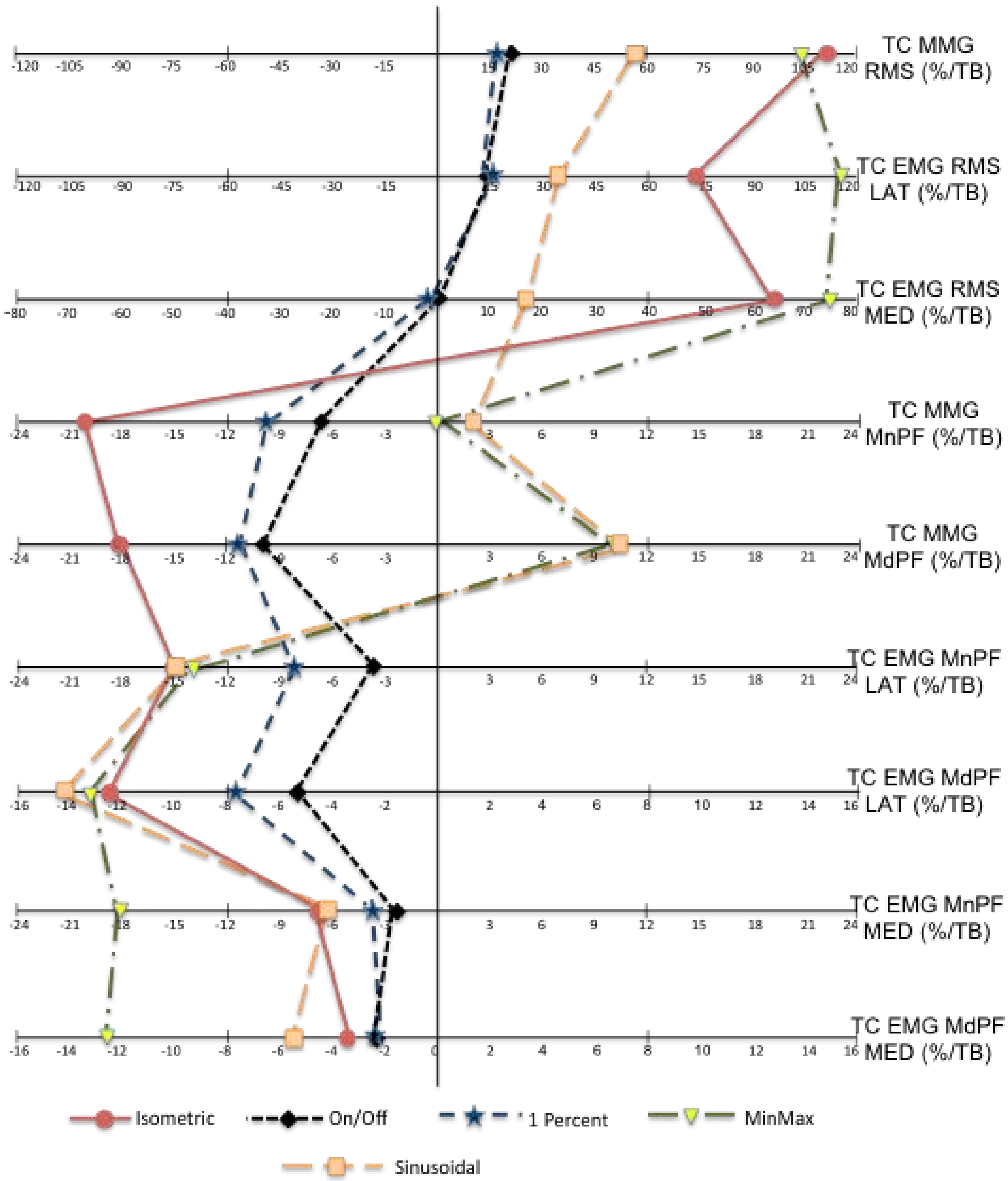

Figure 5.24 Summary of Mean Values for Test Contraction Responses Between Sustained Isometric, On/Off, 1 Percent, MinMax, and Sinusoidal Conditions.

Refer to preliminary discussion for directions leading to greater "fatigue" response. As with Figure 5T, increasing amplitude (RMS) values are indicative of larger fatigue response. Decreasing power domain variables (MnPF and MdPF) are indicative of greater fatigue response.

Table 5.1. Summary of Significant Responses During Exercise Between Sustained Isometric, On/Off, 1 Percent, MinMax, and Sinusoidal Conditions.

Measurement Parameter	Sustained	MinMax	Sinusoidal	1%	On/Off
1. MMG RMS Medial Head Continuous	✗	✓	✓	✓	✓
2. EMG RMS Lateral Head Continuous	✗	✓	✓	✓	✓
3. EMG RMS Medial Head Continuous	✗	✓	✓	✓	✓
4. EMG HiLo Ratio Lateral Head Continuous	✗				✓
5. EMG HiLo Ratio Medial Head Continuous	✗	✓	✓	✓	✓
6. MMG MnPF Medial Head Continuous					
7. MMG MdPF Medial Head Continuous					
8. EMG MnPF Lateral Head Continuous			✓		
9. EMG MdPF Lateral Head Continuous					
10. EMG MnPF Medial Head Continuous	✗		✓		✓
11. EMG MdPF Medial Head Continuous	✗		✓		✓
12. Ratings of Perceived Exertion Continuous	✗	✗✓	✓	✓	✓
13. Heart Rate Continuous	✗				✓
14. Triceps Blood Flow Velocity Continuous	✗	✗	✗✓	✓	✓
15. Maximum Voluntary Force Test Battery	✗		✓	✓	✓
16. Twitch Force Test Battery	✗		✓		✓
17. Low Frequency Fatigue Test Battery	✗		✓		✓
18. MMG RMS Medial Head Test Contraction	✗	✗	✓	✓	✓
19. EMG RMS Lateral Head Test Contraction	✗	✗			✓
20. EMG RMS Medial Head Test Contraction	✗	✗		✓	✓
21. MMG MnPF Medial Head Test Contraction	✗				✓
22. MMG MdPF Medial Head Test Contraction					
23. EMG MnPF Lateral Head Test Contraction	✗				✓
24. EMG MdPF Lateral Head Test Contraction					
25. EMG MnPF Medial Head Test Contraction		✗✓			
26. EMG MdPF Medial Head Test Contraction		✗✓			
27. Endurance Time*	✗	✗	✗✓	✓	✓
28. EMG Gaps Lateral Head – #/min*	✗	✗	✓	✓	✓
29. EMG Gaps Medial Head – #/min*	✗	✗✓	✓	✓	✓
29. EMG Gaps Lateral Head – Mean Duration*	✗	✗		✓	✓
30. EMG Gaps Lateral Head – Median Duration*	✗	✗		✓	✓
31. EMG Gaps Medial Head – Mean Duration*	✗			✓	✓
32. EMG Gaps Medial Head – Median Duration*	✗			✓	✓
Condition vs. Sustained	-	8	16	16	25
Condition vs. On/Off	25	12	2	0	-

✓ = Significantly different from sustained isometric (α = 0.05)
✗ = Significantly different from On/Off (α = 0.05)
* = Non-Parametric Test

General Discussion

One primary finding from this study is that mechanical exposure diversity, for the most part, reduces the rate of fatigue response when compared to the sustained isometric condition. The extent of fatigue for each varying intermittent contraction can be determined based on the trends between multiple measurement parameters. As described in Chapter IV, each measure reflects a change in function at different stages in the physiological processes involved in fatigue development. Based on statistical analysis, the 1 Percent condition, like the On/Off contraction, was significantly slower than sustained isometric in the rate of fatigue response. Sinusoidal also led to significant reduction in the rate of fatigue response when compared to sustained isometric but may be "midway" between sustained isometric and On/Off in blood flow velocity and endurance time. The MinMax condition resulted in 8 conditions that were significantly different than sustained isometric (slower rate of fatigue development) and 12 conditions that were significantly different than On/Off. Of these differences, 4 were significantly different from both On/Off and sustained isometric. This may suggest that MinMax may be precisely between sustained isometric and On/Off, with values trending towards sustained isometric.

Measurement parameters have shown responses where the varying intermittent contractions had slightly larger rate of response magnitudes than sustained isometric or On/Off. For instance, the MinMax condition led to a larger rate of response than sustained isometric in EMG RMS of the lateral head during test contractions. This may suggest that MinMax was more 'problematic' than sustained isometric based on EMG RMS. Meanwhile, Sinusoidal led to a smaller reduction in LFF ratio during exercise when compared to On/Off. This relationship may suggest that Sinusoidal led to delayed LFF response and was more effective than the classical intermittent contraction. Although the mean values of the varying intermittent contractions may have slightly larger magnitudes than sustained isometric or On/Off, there were no statistical differences. Additionally, these aforementioned relationships were parameter specific and were not consistent trends between all measures. In fact, On/Off and sustained isometric were observed to be at polar ends of the response spectrum on a more frequent basis. The larger number of statistically significant differences between sustained isometric and On/Off may further provide evidence that these two conditions belong at opposite ends of the spectrum. As such, the three varying intermittent contractions fit along a continuum between sustained isometric and On/Off, where MinMax is at the middle (closer to sustained isometric) and Sinusoidal and 1 Percent are closer to On/Off.

This continuum reflects the physiological responses during exercise. Recovery, on the other hand, may reveal a different relationship between conditions. As discussed earlier, since all conditions were not completed until exhaustion, a comparison of recoveries cannot be made between exercise protocols. On the contrary, describing the recovery response for each condition can characterize its long-term effects. For example, a decrease in low-frequency fatigue ratio during exercise and recovery may reflect prolonged peripheral fatigue. Sustained isometric, On/Off, and 1 Percent conditions resulted in a decrease in LFF ratio and subsequent recovery 45 to 60 minutes post-exercise, consistent with past literature. MinMax and Sinusoidal resulted in no differences in LFF ratio at cessation and recovery. As discussed in the preliminary discussion, Sinusoidal and MinMax may have led to an adaptation response to prevent the occurrence of low-frequency fatigue. Alternatively, a reduction in LFF ratio may not have occurred if the exercise workloads were different between conditions.

Since LFF measures the capacity of the muscle, contributions from central fatigue are independent to changes in LFF ratio (Vøllestad, 1997). Comparing LFF with changes in maximum voluntary force may provide an estimate of central fatigue (Vøllestad, 1997). All conditions, with the exception of Sinusoidal, led to MVF recovery to within baseline values, 15 to 30 minutes post-exercise. Sinusoidal led to significant force decrement up to 45 minutes recovery. Twitch force may provide further evidence, where sustained isometric and 1 Percent led to prolonged twitch force depression while MinMax and Sinusoidal led to twitch recovery in the first 15 to 30 minutes. MinMax and Sinusoidal contractions based on the lack of tetanic and twitch force generation change and MVF decline, may have resulted in fatigue that can be explained by processes in the CNS. Evidently, the development of both central and peripheral fatigue may be dependent on the variation of force of an intermittent contraction. As a result, conditions may be described by changes in response during exercise, but the long-term effects may be important to consider since real occupational work tasks are continuous and cumulative.

Conclusion

A central question in the current literature is the effects of mechanical exposure diversity, by varying the amplitude, and its relationship with a sustained isometric effort and classical intermittent contraction. This study identified the physiological responses during a 60-minute exercise protocol (or up to exhaustion), during 60 minutes recovery, and at 24-hours post-exercise. A continuum of responses was established where sustained isometric and classical intermittent contractions (On/Off) were at opposite ends of the spectrum. Based on a series of measurement parameters, variation characterized by +/- ½ force amplitude (MinMax: 7.5% MVF – 22.5% MVF) may result in magnitude of responses that were midway between On/Off and sustained isometric. During exercise, Sinusoidal and 1 Percent share similar physiological responses to On/Off. Recovery, on the other hand, showed prolonged fatigue-effects in sustained isometric, On/Off, and 1 Percent. Sinusoidal and MinMax led to central fatigue but long-term peripheral fatigue was less apparent.

Chapter VI

Overview and Addressing the Hypotheses

General Overview

As the landscape of work changes towards an increase in sedentary and computer tasks, interventions must be sought to reduce local fatigue and the risk of musculoskeletal disorders attributed to these low and minimally varying exposures. Past ergonomic guidelines that emphasize decreasing forces may be ineffective, as a meaningful reduction of force may not be feasible. Consequently, it has been suggested that physical variation is an effective intervention (Mathiassen, 2006). However, little is known in how variation within and diversity between physical variation patterns affects both physiological and psychophysical responses.

In this study, multiple measurements of fatigue were collected. As there is no one gold standard measure, the agreement between multiple measurement parameters may indicate the extent and possible mechanism of fatigue. Although each method reacted somewhat differently to the experimental conditions, there was enough commonality to show clear differences between conditions. It was demonstrated that each parameter provided limited information on *mechanisms* associated with fatigue development. This study also used forces, cycle times, and duty cycles that are relevant to occupation and to longer-term health outcomes such as local fatigue.

Chapter IV addressed the central postulate that forces during isometric contractions, which demonstrated variability differ from a sustained isometric exertion. The classical intermittent contraction pattern (On/Off: 0% to 30% MVF) was compared to the 15% MVF sustained isometric condition. It was demonstrated using 32 individual parameters that there was an overwhelming increase in fatigue response when exposed to a submaximal sustained isometric contraction compared to the intermittent contraction. The intermittent contraction led to a lower rate of fatigue response but showed a long-term fatigue response despite less exhaustion.

Chapter V investigated the effects of three intermittent sustained isometric contractions with varying force amplitudes and their relationship to sustained isometric and On/Off conditions. Variation, based on changes in force, may have a positive effect during exercise and recovery. The Minmax (7.5% to 22.5% MVF) contraction pattern showed responses that were between those of the sustained isometric and On/Off conditions. The 1 percent (1% to 29% MVF) condition resulted in exercise responses that were similar to On/Off but also led to prolonged peripheral fatigue. The Sinusoidal contraction (0% to 30% MVF) condition, on the other hand, resulted in delayed fatigue response and improved recovery, possibly due to potentiation effects. Such a continuum of fatigue response is represented in Figure 6.1.

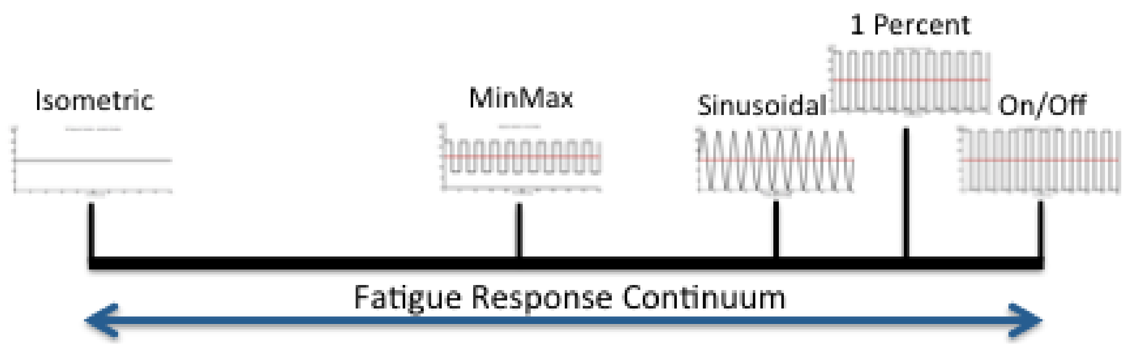

Figure 6.1 Fatigue Response Continuum
The relationship between sustained and intermittent conditions based on fatigue response during exercise. Sustained isometric and On/Off are positioned at opposing ends of the spectrum, MinMax is approximately mid-way (close to sustained isometric), and Sinusoidal and 1 Percent responds similarly to On/Off.

Addressing the Hypotheses

The goal of this study was to address the following hypotheses:

1. The classical intermittent isometric contraction pattern, a repeated cycle of zero mechanical force and 30% of the participant's maximum voluntary force, +/-1 from mean force amplitude, will show a lower rate of fatigue response when compared to a submaximal sustained isometric contraction.

This study has demonstrated the effects of sustained isometric contraction and intermittent contraction patterns over 60 minutes, or up to exhaustion. Through the agreement of response measures, 25 of 32 parameters suggest a significantly different rate of response during the sustained contraction when compared to the intermittent contractions. Each response measure represents a process in the development of longer-term responses, such as fatigue, and its collective agreement may suggest the extent and type of fatigue. Although a slower rate of fatigue response was seen, there remain questions as to the longer-term effects of intermittent contractions during recovery. Although the sustained condition led to quicker recovery rates, the magnitude of the response variable at cessation was greater than the intermittent contraction, thus intrinsically affecting the rate at which the measure has to recover. It was shown that despite the different workloads, long-term fatigue effects were observed after intermittent contractions. This may suggest longer lasting effects will occur if intermittent exercise was done until exhaustion.

2. Activity with +/- ½ mean force amplitude (between 7% and 22.5% of the participant's maximum voluntary force) will show:
    a. A slower rate of fatigue response when compared to a submaximal sustained isometric contraction.
    b. A quicker rate of fatigue response than the classical intermittent on/off contraction pattern.

The MinMax condition was significantly different from the sustained isometric condition in 8 measurement parameters. The MinMax condition was also significantly different from the On/Off contraction in 12 of 32 measurement parameters. Of these measurement parameters, 4 were statistically different from both sustained and On/Off conditions. The observed differences may suggest that MinMax fatigue response is between sustained and On/Off conditions, with values closer to the sustained condition. MinMax recovery, based on the lack of tetanic and twitch force generation change and MVF decline, may involve central fatigue processes. Prolonged low-frequency fatigue effects were not as apparent as those seen in both sustained isometric and On/Off conditions.

3. Exposure to forces between 1% and 29% of the participant's maximum voluntary force will result in:
    a. A slower rate of fatigue response when compared to a submaximal sustained isometric condition.
    b. A quicker rate of fatigue response when compared to the classical intermittent on/off contraction pattern

This study showed that the 1 Percent condition led to fatigue responses that were comparable to the classical intermittent contraction. The 1 Percent condition also led to responses that were significantly different from sustained isometric. Of the 32 measured parameters, 16 responses indicated slower fatigue development during exercise. During recovery, the 1 Percent condition was also comparable to the classical intermittent contraction, resulting in longer-term peripheral fatigue effects. However, according to mechanical force output and EMG gaps analysis, it may have been difficult for participants to discriminate and exert precise forces, i.e. 1%. Nonetheless, the 1 Percent condition had mean force outputs, at the lower contraction phase, that were 1.5% MVF greater than the On/Off condition.

4. Sinusoidal wave patterns will show:
    a. A slower rate of fatigue response when compared to a submaximal sustained isometric contraction.
    b. A quicker rate of muscle fatigue response when compared to the classical intermittent contraction pattern.

The Sinusoidal contraction resulted in 16 measurement responses that were significantly different from the sustained isometric effort. Of these 16 parameters, 2 were also significantly different from the On/Off condition. It can be argued that the Sinusoidal condition led to slower fatigue responses than the sustained isometric contraction but also led to a slightly quicker response rate when compared to On/Off. Sinusoidal, however, may result in better recovery outcomes than sustained isometric and the classical intermittent pattern.

Addressing Additional Issues

Another aim of this study was to address questions pertaining to the response measures:

1) Vøllestad (1997), defined fatigue as the "exercise-induced reduction in the maximal capacity to generate force or power output". Maximum voluntary force and endurance time are often used to detect fatigue. This study showed decreased force output over time in all conditions. Although correlation of measures were not utilized in this study to determine relationships between response parameters, a decrease in force occurred with an increase in EMG amplitude, increase in MMG amplitude, decrease in EMG Hi-Lo ratio, decrease in twitch force, decrease in low-frequency fatigue ratio, increase in heart rate, and increase in ratings of perceived exertion. The preceding responses are commonly indicative of fatigue. Maximum voluntary force may be reflective of the neurophysiological and physiological mechanisms that relate to fatigue development (Vøllestad, 1997).

   Endurance time may be representative of the performance attributes of the particular exercise protocol. A longer endurance time reflects the capabilities to generate the desired forces for a prolonged period of time. A shorter endurance time may be interpreted as failure to exert the desired forces, and thus exhaustion. The sustained isometric condition had a shorter endurance time, consistent with past studies, while the intermittent was performed almost to the entirety of the 60-minute protocol.

2) Muscle twitch and low-frequency force assessments may be a good indicator of the loss of force generating capacity. These assessments may coincide with a reduction in maximum voluntary force production.

   Muscle twitch force and low-frequency fatigue (LFF) decreased in sustained isometric, On/Off, and 1 Percent conditions during exercise and increased to baseline values during recovery. Potentiated twitch force may be more sensitive to muscle fatigue than the tetanic stimulations at 20 Hz and 100 Hz as a prolonged recovery time (up to 24 hours) was observed in the twitch force response. However, previous studies have disputed the accuracy and reliability of twitch response as a measure of fatigue. Low-frequency fatigue ratio (20/100 Hz) decreased during the sustained isometric and 1 Percent contraction and recovered 45 to 60 minutes post-exercise. Intermittent contractions, on the other hand, led to significant depressed ratios at 15 minutes recovery. LFF values returned to baseline 30 minutes after the intermittent contraction. Both twitch force and LFF may be more reflective of the impairment of the cross-bridge cycling mechanism and impairment of calcium handling. In contrast, maximum voluntary force may reflect the gross fatigue process, both neurophysiological and physiological, but non-specific to one particular fatigue process or mechanism. It may be the case that the metabolic accumulation, which impairs cross-bridge cycling, is recovered and the prolonged effects of LFF and twitch response are due to calcium handling impairment.

3) Electromyography and mechanomyogaphy have both been used as indirect tools to measure physiological responses accompanying fatigue. Changes in the time domain and frequency spectra have been widely used parameters during fatigue from prolonged exercise. It is generally accepted that mechanomyography is a more sensitive measure than electromyography and was also apparent in this study. Both electromyography and mechanomyography showed shifts in both amplitude and frequency.

   Analysis of EMG and MMG in both its time and frequency domains revealed increases in amplitude and shifts to lower frequencies with increasing time. These trends were observed in all conditions. MMG root mean square values, a measure of the amplitude of the signal, increased rapidly in all conditions, reflecting the motor unit strategy and possibly the cross-bridge cycling mechanism. EMG RMS also increased in both lateral and medial heads of the triceps brachii. Increases in EMG amplitude may reveal the spatial and temporal summation of motor unit potentials. It was observed that MMG amplitude increased dramatically over time when compared to EMG RMS, possibly due to its improved sensitivity when detecting muscle fatigue. Past literature has shown that 'muscle wisdom' may be an important factor in the interpretation of EMG amplitude activity, signifying a limitation of EMG as a sensitive fatigue measure. Frequency domain variables, however, showed inconsistent results. Shifts to lower frequencies using mean and median measures were observed for most conditions and heads of the triceps. However, values at cessation varied in achieving statistical significance based on mean and median power frequencies. There are limitations and assumptions when using central tendency measures such as mean and median power frequencies, when characterizing the changes in the frequency spectrum. It has been argued that median power frequency is a sensitive to high frequency fatigue yet a poor indicator of low-frequency fatigue. An alternative method is EMG Hi-Lo ratios that describe the changes in EMG frequency by a ratio of two frequency bands. An increase in low-frequency power (reduced Hi-Lo ratio) is indicative of fatigue. This study showed a significant difference in the rate of EMG Hi-Lo ratio response between conditions.

4) Ratings of perceived exertion and heart rate showed increased responses with time for all conditions and mixed relationships between methods were be apparent.

   Both ratings of perceived exertion (RPE) and heart rate increased over the 60 minutes exercise period or until exhaustion. Sustained contractions led to a higher rate of response in both measures when compared to the remaining conditions. In this study, the mean slopes of RPE and heart rate were similar to one another in most conditions. Although no statistical measure was used to analyze the relationship between response measures/tools, previous studies have identified a strong correlation between RPE and heart rate.

5) It was expected that physiological measures (triceps blood flow velocity and heart rate) would show immediate responses during activity and gradually plateau to an equilibrium point and would subsequently show a decrease in response, an indication of fatigue.

   An increase in triceps blood flow was observed in all conditions, with a steep increase during the first few minutes of activity, a gradual increase over time, and a decrease in blood flow in the last few minutes before exhaustion. This was particularly true in the sustained contraction condition.

The intermittent contractions led to a rapid increase blood flow and a gradual increase over time. It is possible that the observed response in the Sustained condition was due to the compression of a vessel supplying the muscle or an impedance of microcirculation as a consequence of increased intramuscular pressure. The intermittent contraction blood flow profile, on the other hand, may be due to a facilitation of blood flow due to rhythmic perfusion. The mean blood velocity was calculated and differences were found between conditions. Sustained isometric, as expected, led to a mean blood flow velocity that was below baseline. The intermittent contractions of varying amplitudes led to increased blood flow velocity over the entire exercise period. Heart rate can also be described as a rapid immediate response followed by a gradual increase until the conclusion of the exercise protocol.

Implications

This study shows that implementing physical variation may provide preferable responses over a set time course when compared to sustained low-level contractions.

In work physiology, intermittent contractions of varying force amplitude requires further study, as this study shows different responses in work patterns beyond sustained isometric and the on/off intermittent exercise.

In ergonomics, implementing physical variation may be in the form of engineering interventions to increase variation when using tools or equipment or changes in job design to implement continuous changes in force, including muscular breaks. Current ergonomic research has focused on reducing high peak forces, which although is important, diverts attention from low and less varying tasks that are also problematic. Time varying forces may provide the necessary mechanism to encourage optimal blood flow and optimal motor unit firing patterns that would benefit the long-term health and well being of workers. These findings may also support the transition from recommending "optimal" postures at a computer workstation in favor of varying forces.

Strengths, Limitations, and Future Research

Strengths of this study included testing the effects of force variation other than on/off cycles. This is the first time this has been done and is important in understanding local fatigue under loading cycles similar to those found in occupational settings. The data will likely be important in helping set local fatigue guidelines. The study used multiple outcome measures to evaluate the effects of the different force profiles. This not only gave better documentation of the fatigue level but similarities and differences between the measures gave insights into possible fatigue mechanisms. The inclusion of a 24hr follow-up to test for long term effects was also valuable as few studies measure such longer-term responses. Prolonged fatigue response (i.e. twitch response) at 24-hours post-exercise was observed in the isometric condition.

The study also had limitations. As each condition was completed until 60 minutes or exhaustion, this was not a classical fatigue study. In fact, each condition was assessed based on responses during a set time course and thus should be considered a performance study. Recovery after sustained isometric and intermittent conditions cannot be compared directly, as the total workload was different between exercise modalities. Nonetheless, the recovery for both conditions was characterized in this study while being cognizant of the difference in exercise workload. It would be beneficial, however, to determine the differences in recovery by exhausting each individual under every condition. This may reveal long-term effects that could be compared between all conditions.

The muscle used in this study was the triceps brachii. This muscle was chosen for its type II fibre dominancy to avoid the effects of daily exposures to different force patterns, which are possibly linked to type I dominant, postural muscles. However, its type II fibre dominancy may influence the results from this study to generalize to all muscles. Although it is recognized that triceps brachii are not common sites of occupationally related injury, its large size and dominant role in the generation of elbow torque makes it very suitable to study fatigue phenomena. Future research should be conducted on muscles consisting of a more equal proportion of type I and type II muscle fibres and that are more subject to work-related fatigue and musculoskeletal disorders. Blood velocity of the brachial artery was also used to assess blood perfusion to the triceps as the artery profunda brachii was difficult to access. It was assumed that the blood kinetics of the brachial artery reflected the course of blood feeding into the triceps brachii. However, it was possible that a proportion of the observed blood kinetics was a reflection of other muscles (i.e forearm musculature) that are fed by the brachial artery.

A limitation of this study was that 15 university-age males were participants in this study. As the number of females and older-aged workers rise (Statistics Canada, 2006), interventions that are applicable to work should include these population groups. Further research should include both genders and persons of varying age.

It would also be interesting to observe continuous responses without the inclusion of test batteries every 15 minutes. In this study, 4 minutes of data were removed after each test battery to ensure no residual effect from the test battery itself and the recovery time between the test battery constituents. However, it is not known whether the carry-over effect from the test battery, by contributing to the fatigue response or imposing rest breaks, was longer than the data removed. According to Krajcarski and Wells (2008), given the time-dependent nature of many biological systems, rest breaks have an effect on the overall risk estimates. The removal of test batteries, on the other hand, may not allow for assessments based on test contractions, twitch force, low-frequency fatigue, and maximum voluntary force.

The elbow torque was created using position control; the participant held their arm in a fixed position against the moment created by the apparatus. It has been shown that the response differs when compared to force control. Position control has been shown to create more fatigue however it may better reflect common tasks.

Participants had difficulty in quickly reducing force output in the On/Off and 1% conditions. A rapid drop followed by a slow reduction to the desired force was observed. This meant that the average force over the desired force period was greater than desired however the 1% condition did show average forces about 1.5 % MVF higher that the On/Off condition.

Although 8 methods were used to measure muscle response, there are a wealth of methods and measurements that may provide additional information of the fatigue process. Invasive measures, such as blood sampling, may provide detailed information of the biomarkers associated with the physiological response. Another possible response is the change in interjoint and intermuscular coordination to compensate for local effects of fatigue and to maintain key movement characteristics. According to Côté and colleagues (2008), there may be a modification of motion at proximal joints to compensate for fatigue-elicited displacements at distal segments.

Future research should be dedicated to performing each condition to exhaustion and to explore the influence of duty cycle and cycle time in longer-term physiological responses. Although intermittent contraction studies based on duty cycle and cycle time changes have been conducted, very few studies have investigated these effects in addition to amplitude changes.

The cumulative research may lead to further studies to explore whether training and possible interventions can decisively influence adaptation response to work.

Appendix A

Statistical Analysis for Chapter IV

Endurance Time

Measure	Condition	Mean (SD)	Z	P
Completion Time (seconds)	Sustained	1066.62 (1162.68)	Z = 2.93	P = 0.003
	On/Off	2918.23 (1234.69)		

EMG Gaps

Measure	Condition	Mean (SD)	Z	P
# Gaps/Minute Lateral Head	Sustained	0.052 (0.130)	Z = 2.824	P = 0.005
	On/Off	12.237 (11.806)		
#Gaps/Minute Medial Head	Sustained	0.0523 (0.103)	Z = 2.667	P = 0.008
	On/Off	11.874 (13.427)		
Mean Duration Lateral Head	Sustained	0.045 (0.111)	Z = 2.746	P = 0.006
	On/Off	0.382 (0.271)		
Mean Duration Medial Head	Sustained	0.088 (0.175	Z = 2.845	P = 0.004
	On/Off	0.318 (0.192)		
Median Duration Lateral Head	Sustained	0.046 (0.113)	Z = 2.667	P = 0.008
	On/Off	0.290 (0.163)		
Median Duration Medial Head	Sustained	0.085 (0.172)	Z = 2.756	P = 0.006
	On/Off	0.271 (0.152)		

Mechanical Force (%MVF) – Comparison at Different Force Levels

Condition	Low Force Level			High Force Level		
	Pre	Post	Paired T-Test	Pre	Post	Paired T-Test
Sustained	15.034 (2.838)	14.748 (1.574)	T(13) = 0.399 P = 0.697	-	-	-
On/Off	4.213 (1.935)	3.670 (1.469)	T(13) = 0.965 P = 0.353	29.666 (1.767)	28.894 (2.300)	T(13) = 1.854 P = 0.088

Test Battery Measurements – Rate of Response (%/Test Contraction)

Time Int.	Condition	Mean Slope (SD)	T	P	Cohen's D
MMG RMS Med	Sustained	112.030 (79.186)	T(13) = 3.489	P = 0.002	D = 1.461
	On/Off	21.166 (38.218)			
EMG RMS Lat	Sustained	49.244 (55.992)	T(13) = 1.865	P = 0.043	D = 0.704
	On/Off	15.312 (38.844)			
EMG RMS Med	Sustained	64.757 (87.952)	T(13) = 2.920	P = 0.007	D = 1.023
	On/Off	0.117 (15.827)			
MMG MnPF Med	Sustained	-20.664 (23.653)	T(13) = 2.025	P = 0.033	D = 0.753
	On/Off	-6.699 (11.370)			
MMG MdPF Med	Sustained	-18.215 (32.951)	T(13) = 1.033	P = 0.161	D = 0.317
	On/Off	-9.548 (20.332)			
EMG MnPF Lat	Sustained	-14.989 (12.669)	T(13) = 2.335	P = 0.019	D = 0.788
	On/Off	-3.347 (16.630)			
EMG MdPF Lat	Sustained	-12.548 (11.483)	T(13) = 1.404	P = 0.093	D = 0.582
	On/Off	-5.703 (12.044)			
EMG MnPF Med	Sustained	-6.432 (10.040)	T(13) = 1.142	P = 0.138	D = 0.527
	On/Off	-2.645 (5.902)			
EMG MdPF Med	Sustained	-3.854 (9.754)	T(13) = 0.503	P = 0.312	D = 0.201
	On/Off	-2.329 (4.482)			
Twitch	Sustained	-22.460 (26.216)	T(13) = 2.229	P = 0.023	D = 0.598
	On/Off	-9.757 (14.668)			
LFF	Sustained	-18.274 (15.885)	T(13) = 3.945	P = 0.001	D = 1.026
	On/Off	-4.556 (10.250)			
Force	Sustained	-14.408 (14.279)	T(13) = 2.681	P = 0.010	D = 0.647
	On/Off	-6.754 (8.716)			

Continuous Measurements – Rate of Response (%/min)

Time Int.	Condition	Mean Slope (SD)	T	P	Cohen's D
MMG RMS Med	Sustained	8.514 (7.525)	T(13) = 3.724	P = 0.002	D = 1.383
	On/Off	0.979 (1.669)			
EMG RMS Lat	Sustained	10.450 (13.378)	T(13) = 2.254	P = 0.022	D = 0.903
	On/Off	1.592 (3.698)			
EMG RMS Med	Sustained	6.319 (8.905)	T(13) = 2.175	P = 0.025	D = 0.861
	On/Off	0.762 (2.004)			
EMG HiLo Ratio Lat	Sustained	-13.067 (18.033)	T(13) = 2.376	P = 0.018	D = 0.795
	On/Off	-0.311 (13.793)			
EMG HiLo Ratio Med	Sustained	-15.108 (15.555)	T(13) = 3.430	P = 0.003	D = 0.965
	On/Off	-2.400 (10.253)			
MMG MnPF	Sustained	-4.513 (12.106)	T(13) = 0.869	P = 0.201	D = 0.303
	On/Off	-1.896 (5.579)			
MMG MdPF	Sustained	-3.051 (22.746)	T(13) = 0.164	P = 0.437	D = 0.062
	On/Off	-1.981 (8.935)			
EMG MnPF Lat	Sustained	-8.444 (10.762)	T(13) = 1.670	P = 0.061	D = 0.549
	On/Off	-3.100 (8.596)			
EMG MdPF Lat	Sustained	-4.068 (12.925)	T(13) = 0.930	P = 0.186	D = 0.367
	On/Off	1.695 (18.058)			
EMG MnPF Med	Sustained	-8.405 (7.173)	T(13) = 2.793	P = 0.008	D = 1.056
	On/Off	-1.987 (4.732)			
EMG MdPF Med	Sustained	-6.420 (7.946)	T(13) = 2.149	P = 0.014	D = 0.945
	On/Off	0.138 (5.755)			
RPE	Sustained	18.237 (12.874)	T(13) = 4.731	P = 0.000	D = 1.282
	On/Off	5.287 (6.195)			
Mean Blood Vel.	Sustained	87.551 (82.812)	T(13) = 8.440	P = 0.000	D = 0.380
	On/Off	189.87 (155.440)			
Heart Rate	Sustained	19.698 (14.791)	T(13) = 4.206	P = 0.001	D = 0.849
	On/Off	9.457 (8.497)			

Appendix B

Statistical Analysis for Chapter V (Exercise)

Endurance Time (seconds)

Condition	Mean (SD)	Percentiles			Chi-Square	P
		25th	50th (Median)	75th		
Sustained	1066.62 (1162.68)	408.000	579.000	1191.500	χ^2 (4) = 24.661	P = 0.000
On/Off	2918.23 (1234.69)	2274.000	3600.000	3600.000		
1 Percent	2411.46 (1400.16)	650.000	3202.000	3600.000		
MinMax	1727.46 (1219.63)	694.000	1474.000	2901.000		
Sinusoidal	2128.46 (1357.54)	711.000	2205.000	3600.000		

Condition (A)	Condition (B)	Z	P
Sustained	1 Percent	Z = -2.580	P = 0.005
	MinMax	Z = -1.851	P = 0.032
	Sinusoidal	Z = -2.353	P = 0.010
On/Off	MinMax	Z = -2.756	P = 0.003
	Sinusoidal	Z = -2.521	P = 0.006

Maximum Voluntary Force – Rate of Response (%/Test Battery)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-14.408 (14.279)	χ^2 (9) = 17.121 P = 0.049 ε = 0.536	F(0,4) = 7.054	P = 0.002	η_p^2 = 0.370
On/Off	-6.754 (8.716)				
1 Percent	-3.739 (3.501)				
MinMax	-11.162 (9.406)				
Sinusoidal	-6.369 (5.207)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	10.669	P = 0.000
	MinMax	3.246	P = 0.209
	Sinusoidal	8.038	P = 0.002
On/Off	MinMax	-4.408	P = 0.087
	Sinusoidal	0.385	P = 0.500

Twitch Force – Rate of Response (%/Test Battery)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-22.460 (26.216)	χ^2 (9) = 22.779 P = 0.007 ε = 0.548	F(0,4) = 2.047	P = 0.029	η_p^2 = 0.270
On/Off	-9.757 (14.668)				
1 Percent	-11.737 (19.222)				
MinMax	-14.516 (23.114)				
Sinusoidal	-10.095 (14.104)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	10.723	P = 0.068
	MinMax	7.944	P = 0.180
	Sinusoidal	12.365	P = 0.035
On/Off	MinMax	-4.759	P = 0.384
	Sinusoidal	-0.338	P = 0.500

Continuous MMG RMS – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	8.514 (7.525)	χ^2 (9) = 45.924 P = 0.000 ε = 0.365	F(0,4) = 11.796	P = 0.001	$\eta_p^2 = 0.496$
On/Off	0.979 (1.669)				
1 Percent	0.715 (1.733)				
MinMax	1.463 (2.236)				
Sinusoidal	0.501 (1.864)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	-7.800	P = 0.000
	MinMax	-7.052	P = 0.000
	Sinusoidal	-8.014	P = 0.000
On/Off	MinMax	0.484	P = 0.500
	Sinusoidal	-0.478	P = 0.500

Continuous EMG RMS Medial Head – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	6.319 (8.905)	χ^2 (9) = 85.670 P = 0.000 ε = 0.272	F(0,4) = 4.984	P = 0.041	$\eta_p^2 = 0.293$
On/Off	0.762 (2.004)				
1 Percent	0.709 (1.104)				
MinMax	1.266 (1.759)				
Sinusoidal	-0.108 (1.402)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	-5.610	P = 0.003
	MinMax	-5.053	P = 0.006
	Sinusoidal	-6.427	P = 0.001
On/Off	MinMax	0.504	P = 0.500
	Sinusoidal	-0.870	P = 0.478

Continuous EMG RMS Lateral Head – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	10.450 (13.378)	χ^2 (9) = 46.345 P = 0.000 ε = 0.369	F(0,4) = 4.792	P = 0.015	η_p^2 = 0.285
On/Off	1.592 (3.698)				
1 Percent	1.038 (1.873)				
MinMax	3.416 (5.824)				
Sinusoidal	0.278 (3.216)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	-9.412	P = 0.002
	MinMax	-7.034	P = 0.019
	Sinusoidal	-10.172	P = 0.001
On/Off	MinMax	1.825	P = 0.450
	Sinusoidal	-1.314	P = 0.483

Test Contractions MMG RMS – Rate of Response (%/Test Contraction)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	112.028 (79.186)	χ^2 (9) = 14.116 P = 0.122 ε = 0.657	F(0,4) = 8.233	P = 0.000	η_p^2 = 0.407
On/Off	21.166 (38.218)				
1 Percent	16.061 (24.409)				
MinMax	104.201 (66.780)				
Sinusoidal	57.175 (54.235)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	-95.967	P = 0.000
	MinMax	-7.827	P = 0.495
	Sinusoidal	-54.853	P = 0.028
On/Off	MinMax	83.035	P = 0.001
	Sinusoidal	36.009	P = 0.155

Test Contractions EMG RMS Lateral Head – Rate of Response (%/Test Contraction)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	49.244 (55.992)	χ^2 (9) = 30.504 P = 0.000 ε = 0.405	F(0,4) = 3.641	P = 0.006	η_p^2 = 0.233
On/Off	15.312 (38.844)				
1 Percent	15.794 (58.542)				
MinMax	119.187 (158.260)				
Sinusoidal	34.336 (49.811)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	-33.450	P = 0.338
	MinMax	69.943	P = 0.052
	Sinusoidal	-14.908	P = 0.486
On/Off	MinMax	103.875	P = 0.004
	Sinusoidal	19.024	P = 0.467

Test Contractions EMG RMS Medial Head – Rate of Response (%/Test Contraction)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	64.757 (87.953)	χ^2 (9) = 36.382 P = 0.000 ε = 0.405	F(0,4) = 4.111	P = 0.003	η_p^2 = 0.255
On/Off	0.117 (15.827)				
1 Percent	-1.848 (21.729)				
MinMax	76.065 (122.886)				
Sinusoidal	18.204 (47.726)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	-66.605	P = 0.021
	MinMax	11.308	P = 0.489
	Sinusoidal	-46.554	P = 0.110
On/Off	MinMax	75.948	P = 0.008
	Sinusoidal	18.086	P = 0.444

Continuous MMG MnPF – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-4.513 (12.106)	χ^2 (9) = 23.533 P = 0.006 ε = 0.634 (Huynh-Feldt)	F(0,4) = 1.041	P = 0.195	$\eta_p^2 = 0.080$
On/Off	-1.896 (5.579)				
1 Percent	2.933 (13.871)				
MinMax	-4.420 (8.908)				
Sinusoidal	-0.595 (11.151)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	7.446	P = 0.127
	MinMax	0.093	P = 0.500
	Sinusoidal	3.918	P = 0.384
On/Off	MinMax	-2.523	P = 0.469
	Sinusoidal	1.301	P = 0.497

Continuous MMG MdPF – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-3.051 (22.746)	χ^2 (9) = 23.578 P = 0.006 ε = 0.618 (Huynh-Feldt)	F(0,4) = 0.685	P = 0.287	$\eta_p^2 = 0.054$
On/Off	-1.981 (8.935)				
1 Percent	3.838 (15.235)				
MinMax	-6.855 (14.735)				
Sinusoidal	0.585 (19.542)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	6.889	P = 0.354
	MinMax	-3.804	P = 0.475
	Sinusoidal	3.636	P = 0.478
On/Off	MinMax	-4.875	P = 0.443
	Sinusoidal	2.566	P = 0.494

Continuous EMG MnPF Lateral Head – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-8.444 (10.762)	χ^2 (9) = 5.611 P = 0.781 ε = 0.808 (Huynh-Feldt)	F(0,4) = 1.550	P = 0.102	$\eta_p^2 = 0.114$
On/Off	-3.096 (8.596)				
1 Percent	-3.853 (5.918)				
MinMax	-4.380 (6.730)				
Sinusoidal	-2.229 (8.707)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	4.590	P = 0.141
	MinMax	4.064	P = 0.193
	Sinusoidal	6.215	P = 0.044
On/Off	MinMax	-1.284	P = 0.486
	Sinusoidal	0.867	P = 0.497

Continuous EMG MdPF Lateral Head – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-4.068 (12.925)	χ^2 (9) = 13.062 P = 0.165 ε = 0.599 (Huynh-Feldt)	F(0,4) = 0.653	P = 0.294	$\eta_p^2 = 0.052$
On/Off	1.695 (18.058)				
1 Percent	-4.518 (6.147)				
MinMax	-0.975 (12.308)				
Sinusoidal	-2.876 (7.828)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	-0.450	P = 0.500
	MinMax	3.093	P = 0.448
	Sinusoidal	1.192	P = 0.499
On/Off	MinMax	-2.669	P = 0.467
	Sinusoidal	-4.571	P = 0.348

Continuous EMG MnPF Medial Head – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-8.405 (7.173)	χ^2 (9) = 28.294 P = 0.001 ε = 0.479	F(0,4) = 2.357	P = 0.060	η_p^2 = 0.164
On/Off	-1.987 (4.732)				
1 Percent	-3.629 (4.037)				
MinMax	-4.538 (2.938)				
Sinusoidal	-2.539 (10.289)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	4.776	P = 0.072
	MinMax	3.867	P = 0.148
	Sinusoidal	5.867	P = 0.026
On/Off	MinMax	-2.551	P = 0.325
	Sinusoidal	-0.552	P = 0.499

Continuous EMG MdPF Medial Head – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-6.420 (7.946)	χ^2 (9) = 28.849 P = 0.001 ε = 0.570	F(0,4) = 2.724	P = 0.002	η_p^2 = 0.185
On/Off	0.138 (5.755)				
1 Percent	-2.914 (2.827)				
MinMax	-3.174 (3.560)				
Sinusoidal	-0.878 (7.037)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	3.505	P = 0.156
	MinMax	3.246	P = 0.189
	Sinusoidal	5.541	P = 0.023
On/Off	MinMax	-3.312	P = 0.180
	Sinusoidal	-1.016	P = 0.486

Test Contraction MMG MnPF – Rate of Response (%/Test Contraction)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-20.664 (23.653)	χ^2 (9) = 13.199 P = 0.159 ε = 0.635 (Huynh-Feldt)	F(0,4) = 1.473	P = 0.118	η_p^2 = 0.109
On/Off	-6.699 (11.370)				
1 Percent	-9.715 (26.227)				
MinMax	0.145 (36.682)				
Sinusoidal	2.455 (27.788)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	10.949	P = 0.347
	MinMax	20.810	P = 0.086
	Sinusoidal	23.120	P = 0.056
On/Off	MinMax	6.845	P = 0.459
	Sinusoidal	9.155	P = 0.403

Test Contraction MMG MdPF – Rate of Response (%/Test Contraction)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-18.215 (32.951)	χ^2 (9) = 15.331 P = 0.086 ε = 0.614	F(0,4) = 1.570	P = 0.099	$\eta_p^2 = 0.116$
On/Off	-9.548 (20.332)				
1 Percent	-11.790 (35.401)				
MinMax	10.030 (53.694)				
Sinusoidal	10.137 (47.540)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	6.425	P = 0.489
	MinMax	28.245	P = 0.094
	Sinusoidal	28.352	P = 0.093
On/Off	MinMax	19.578	P = 0.246
	Sinusoidal	19.685	P = 0.243

Test Contraction EMG MnPF Lateral Head – Rate of Response (%/Test Contraction)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-14.989 (12.669)	χ^2 (9) = 14.261 P = 0.118 ε = 0.590	F(0,4) = 1.272	P = 0.147	η_p^2 = 0.096
On/Off	-3.347 (16.630)				
1 Percent	-8.536 (12.141)				
MinMax	-13.587 (26.541)				
Sinusoidal	-14.995 (10.692)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	6.453	P = 0.352
	MinMax	1.402	P = 0.499
	Sinusoidal	-0.006	P = 0.500
On/Off	MinMax	-10.240	P = 0.160
	Sinusoidal	-11.648	P = 0.109

Test Contraction EMG MdPF Lateral Head – Rate of Response (%/Test Contraction)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-12.548 (11.483)	χ^2 (9) = 8.106 P = 0.529 ε = 0.762	F(0,4) = 1.569	P = 0.099	η_p^2 = 0.116
On/Off	-5.703 (12.044)				
1 Percent	-7.952 (9.568)				
MinMax	-13.655 (15.805)				
Sinusoidal	-14.730 (11.622)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	4.595	P = 0.341
	MinMax	-1.108	P = 0.499
	Sinusoidal	-2.182	P = 0.483
On/Off	MinMax	-7.962	P = 0.113
	Sinusoidal	-9.027	P = 0.071

Test Contraction EMG MnPF Medial Head – Rate of Response (%/Test Contraction)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-6.432 (10.040)	χ^2 (9) = 14.848 P = 0.099 ε = 0.611 (Huynh-Feldt)	F(0,4) = 4.804	P = 0.001	$\eta_p^2 = 0.286$
On/Off	-2.645 (5.902)				
1 Percent	-3.300 (6.011)				
MinMax	-18.090 (17.223)				
Sinusoidal	-6.165 (8.787)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	3.131	P = 0.413
	MinMax	-11.659	P = 0.010
	Sinusoidal	0.267	P = 0.500
On/Off	MinMax	-15.446	P = 0.001
	Sinusoidal	-3.520	P = 0.398

Test Contraction EMG MdPF Medial Head – Rate of Response (%/Test Contraction)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-3.854 (9.754)	χ^2 (9) = 12.562 P = 0.189 ε = 0.673	F(0,4) = 2.942	P = 0.015	η_p^2 = 0.197
On/Off	-2.329 (4.482)				
1 Percent	-2.434 (5.525)				
MinMax	-12.457 (10.922)				
Sinusoidal	-5.893 (12.981)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	1.420	P = 0.491
	MinMax	-8.602	P = 0.028
	Sinusoidal	-2.038	P = 0.469
On/Off	MinMax	-10.127	P = 0.010
	Sinusoidal	-3.564	P = 0.347

Continuous EMG Hi-Lo Ratio Lateral Head – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-13.067 (18.033)	χ^2 (9) = 15.183 P = 0.090 ε = 0.740	F(0,4) = 1.686	P = 0.085	$\eta_p^2 = 0.123$
On/Off	-0.311 (13.793)				
1 Percent	-5.831 (13.559)				
MinMax	-1.804 (20.414)				
Sinusoidal	-2.109 (6.090)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	7.236	P = 0.255
	MinMax	11.263	P = 0.766
	Sinusoidal	10.959	P = 0.852
On/Off	MinMax	-1.493	P = 0.499
	Sinusoidal	-1.797	P = 0.497

Continuous EMG Hi-Lo Ratio Medial Head – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-15.108 (15.555)	χ^2 (9) = 10.664 P = 0.305 ε = 0.776	F(0,4) = 5.329	P = 0.001	η_p^2 = 0.307
On/Off	-2.400 (10.253)				
1 Percent	-3.628 (9.719)				
MinMax	-7.166 (7.632)				
Sinusoidal	-2.304 (9.016)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	11.480	P = 0.002
	MinMax	7.942	P = 0.034
	Sinusoidal	12.804	P = 0.001
On/Off	MinMax	-4.766	P = 0.207
	Sinusoidal	0.096	P = 0.500

Test Battery Low-Frequency Fatigue Ratio – Rate of Response (%/Test Battery)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	-18.274 (15.885)	χ^2 (9) = 33.320 P = 0.000 ε = 0.491	F(0,4) = 2.056	P = 0.076	$\eta_p^2 = 0.146$
On/Off	-4.556 (10.250)				
1 Percent	-4.275 (15.951)				
MinMax	0.1302 (25.279)				
Sinusoidal	6.556 (42.021)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	13.999	P = 0.174
	MinMax	18.405	P = 0.072
	Sinusoidal	24.830	P = 0.014
On/Off	MinMax	4.687	P = 0.480
	Sinusoidal	11.112	P = 0.274

Mean Blood Flow Velocity For All Conditions (Normalized to Baseline, 100%)

Condition	Mean (SD)	F Value	P	Partial Eta Squared η_p^2
Sustained	87.551 (82.812)	F(0,4) = 28.91	P = 0.0001	$\eta_p^2 = 0.173$
On/Off	189.87 (155.440)			
1 Percent	121.85 (46.496)			
MinMax	107.78 (88.025)			
Sinusoidal	121.28 (93.169)			

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	34.299	P = 0.0028
	MinMax	20.226	P = 0.1446
	Sinusoidal	33.729	P = 0.0034
On/Off	MinMax	-82.091	P = 0.0001
	Sinusoidal	-68.587	P = 0.0001

Continuous Ratings of Perceived Exertion – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	18.237 (12.873)	χ^2 (9) = 22.233 P = 0.009 ε = 0.594	F(0,4) = 11.091	P = 0.000	η_p^2 = 0.480
On/Off	5.287 (6.195)				
1 Percent	6.501 (7.756)				
MinMax	10.853 (9.446)				
Sinusoidal	9.484 (11.108)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	-11.736	P = 0.000
	MinMax	-7.384	P = 0.003
	Sinusoidal	-8.752	P = 0.001
On/Off	MinMax	5.566	P = 0.022
	Sinusoidal	4.198	P = 0.087

Continuous Heart Rate – Rate of Response (%/min)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Sustained	19.698 (14.791)	χ^2 (9) = 37.811 P = 0.000 ε = 0.417	F(0,4) = 2.375	P = 0.063	η_p^2 = 0.165
On/Off	9.458 (8.497)				
1 Percent	10.049 (8.665)				
MinMax	20.225 (32.025)				
Sinusoidal	8.577 (9.176)				

Condition (A)	Condition (B)	Mean Difference	P
Sustained	1 Percent	-9.649	P = 0.114
	MinMax	0.527	P = 0.500
	Sinusoidal	-11.121	P = 0.067
On/Off	MinMax	10.767	P = 0.077
	Sinusoidal	-0.881	P = 0.500

EMG Gaps – Number of Gaps Per Minute

Measure	Condition	Mean (SD)	Percentiles			Chi-Square	P
			25th	50th (Median)	75th		
# Gaps/Min Lat Head	Sustained	0.052 (0.130)	0	0	0	χ^2 (4) = 24.909	P = 0.000
	On/Off	12.237 (11.806)	0.156	8.696	21.600		
	1 Percent	10.540 (11.183)	0.085	5.882	22.550		
	MinMax	0.219 (0.234)	0	0.217	0.444		
	Sinusoidal	7.690 (10.512)	0	4.182	12.546		
# Gaps/Min Med Head	Isometric	0.052 (0.103)	0	0	0.09	χ^2 (4) = 25.886	P = 0.000
	On/Off	11.874 (13.427)	0.05	8.717	25.117		
	1 Percent	17.346 (14.661)	0.811	16.133	33.134		
	MinMax	1.195 (1.876)	0	0.550	1.842		
	Sinusoidal	5.623 (6.692)	0.050	2.933	10.877		

Gaps/Minute Lateral Head

Condition (A)	Condition (B)	Z	P
Sustained	1 Percent	Z = -2.934	P = 0.002
	MinMax	Z = -1.960	P = 0.025
	Sinusoidal	Z = -2.497	P = 0.006
On/Off	MinMax	Z = -2.934	P = 0.002
	Sinusoidal	Z = -1.334	P = 0.182

#

Gaps/Minute Medial Head

Condition (A)	Condition (B)	Z	P
Sustained	1 Percent	Z = -2.803	P = 0.003
	MinMax	Z = -2.521	P = 0.006
	Sinusoidal	Z = -2.803	P = 0.003
On/Off	MinMax	Z = -2.667	P = 0.004
	Sinusoidal	Z = -1.334	P = 0.182

Bonferroni Correction: P = 0.05/5 = 0.01. Condition (B) – Condition (A)

EMG Gaps – Mean and Median Duration (per gap) of Lateral Head (seconds)

Measure	Condition	Mean (SD)	Percentiles			Chi-Square	P
			25th	50th (Median)	75th		
Mean Duration	Sustained	0.045 (0.111)	0	0	0	χ^2 (4) = 20.760	P = 0.000
	On/Off	0.382 (0.271)	0.222	0.350	0.574		
	1 Percent	0.337 (0.264)	0.209	0.302	0.407		
	MinMax	0.182 (0.162)	0	0.237	0.297		
	Sinusoidal	0.218 (0.188)	0	0.280	0.386		
Median Duration	Sustained	0.046 (0.113)	0	0	0	χ^2 (4) = 23.182	P = 0.000
	On/Off	0.290 (0.163)	0.222	0.326	0.369		
	1 Percent	0.274 (0.170)	0.209	0.267	0.334		
	MinMax	0.155 (0.129)	0	0.224	0.262		
	Sinusoidal	0.184 (0.154)	0	0.258	0.305		

Mean Duration

Condition (A)	Condition (B)	Z	P
Sustained	1 Percent	Z = -2.934	P = 0.002
	MinMax	Z = -1.955	P = 0.025
	Sinusoidal	Z = -2.191	P = 0.014
On/Off	MinMax	Z = -2.667	P = 0.004
	Sinusoidal	Z = -1.600	P = 0.065

Median Duration

Condition (A)	Condition (B)	Z	P
Sustained	1 Percent	Z = -2.845	P = 0.002
	MinMax	Z = -1.599	P = 0.065
	Sinusoidal	Z = -1.682	P = 0.047
On/Off	MinMax	Z = -2.934	P = 0.002
	Sinusoidal	Z = -1.784	P = 0.037

EMG Gaps – Mean and Median Duration (per gap) of Medial Head (seconds)

Measure	Condition	Mean (SD)	Percentiles 25th	50th (Median)	75th	Chi-Square	P
Mean Duration	Sustained	0.088 (0.175)	0	0	0.144	χ^2 (4) = 10.655	P = 0.031
	On/Off	0.318 (0.192)	0.218	0.290	0.461		
	1 Percent	0.301 (0.197)	0.140	0.333	0.443		
	MinMax	0.214 (0.188)	0	0.258	0.365		
	Sinusoidal	0.243 (0.151)	0.122	0.279	0.320		
Median Duration	Sustained	0.085 (0.172)	0	0	0.119	χ^2 (4) = 12.646	P = 0.013
	On/Off	0.272 (0.152)	0.218	0.262	0.360		
	1 Percent	0.251 (0.160)	0.114	0.296	0.355		
	MinMax	0.178 (0.151)	0	0.245	0.301		
	Sinusoidal	0.216 (0.130)	0.114	0.269	0.289		

Mean Duration

Condition (A)	Condition (B)	Z	P
Sustained	1 Percent	Z = -2.497	P = 0.006
	MinMax	Z = -2.100	P = 0.018
	Sinusoidal	Z = -2.191	P = 0.014
On/Off	MinMax	Z = -1.956	P = 0.025
	Sinusoidal	Z = -1.334	P = 0.091

Median Duration

Condition (A)	Condition (B)	Z	P
Sustained	1 Percent	Z = -2.499	P = 0.007
	MinMax	Z = -1.820	P = 0.035
	Sinusoidal	Z = -1.988	P = 0.024
On/Off	MinMax	Z = -2.223	P = 0.013
	Sinusoidal	Z = -1.098	P = 0.136

Mechanical Force – Comparison at Different Force Levels (% MVF)

Condition	Low Force Level			High Force Level		
	Pre	Post	Paired T-Test	Pre	Post	Paired T-Test
Sustained	15.034 (2.838)	14.748 (1.574)	T(13) = 0.399 P = 0.697	-	-	-
On/Off	4.213 (1.935)	3.670 (1.469)	T(13) = 0.965 P = 0.353	29.666 (1.767)	28.894 (2.300)	T(13) = 1.854 P = 0.088
1 Percent	4.528 (1.981)	3.501 (0.816)	T(13) = 1.624 P = 0.130	27.528 (2.572)	26.717 (2.383)	T(13) = 2.687 P = 0.020
MinMax	9.062 (2.267)	8.989 (2.737)	T(13) = 0.157 P = 0.878	21.736 (1.881)	20.137 (2.765)	T(13) = 2.543 P = 0.026
Sinusoidal	6.481 (1.527)	4.420 (1.410)	T(13) = 5.521 P = 0.000	29.018 (2.389)	27.126 (3.779)	T(13) = 2.681 P = 0.020

Appendix C

Statistical Analysis for Chapter IV and V (Recovery)

Maximum Voluntary Force Recovery – Sustained (%MVF)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	80.470 (15.143)	χ^2 (9) = 71.579 P = 0.000 ε = 0.373	F(0,6) = 3.465	P = 0.021	η_p^2 = 0.224
Cessation	66.749 (14.278)				
Recovery 15 mins	73.219 (10.074)				
Recovery 30 mins	71.777 (12.315)				
Recovery 45 mins	74.796 (10.850)				
Recovery 60 mins	76.960 (9.938)				
24 Hrs Post Ex	76.450 (14.340)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-13.694	P = 0.001
	R15	-8.693	P = 0.025
	R30	-7.251	P = 0.069
	R45	-5.674	P = 0.170
	R60	-3.511	P = 0.392
	24 Hrs	-4.020	P = 0.339

Maximum Voluntary Force Recovery – On/Off (%MVF)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	78.512 (13.268)	χ^2 (9) = 24.374 P = 0.250 ε = 0.660	F(0,6) = 4.510	P = 0.001	η_p^2 = 0.273
Cessation	66.891 (12.539)				
Recovery 15 mins	69.465 (11.123)				
Recovery 30 mins	72.370 (12.976)				
Recovery 45 mins	74.512 (13.053)				
Recovery 60 mins	75.737 (13.619)				
24 Hrs Post Ex	79.897 (9.753)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-11.621	P = 0.001
	R15	-9.047	P = 0.013
	R30	-6.142	P = 0.108
	R45	-3.999	P = 0.314
	R60	-2.775	P = 0.442
	24 Hrs	-1.385	P = 0.498

Maximum Voluntary Force Recovery – 1 Percent (%MVF)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	77.101 (12.333)	χ^2 (9) = 34.092 P = 0.032 ε = 0.547 (Huynh-Feldt)	F(0,6) = 5.475	P = 0.000	$\eta_p^2 = 0.313$
Cessation	68.807 (10.362)				
Recovery 15 mins	72.092 (13.387)				
Recovery 30 mins	72.511 (12.738)				
Recovery 45 mins	76.369 (11.779)				
Recovery 60 mins	77.001 (13.488)				
24 Hrs Post Ex	79.623 (12.808)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-8.294	P = 0.002
	R15	-5.008	P = 0.066
	R30	-4.589	P = 0.096
	R45	-0.732	P = 0.500
	R60	-0.099	P = 0.500
	24 Hrs	2.523	P = 0.374

Maximum Voluntary Force Recovery – MinMax (%MVF)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	76.813 (13.039)	χ^2 (9) = 42.174 P = 0.004 ε = 0.547 (Huynh-Feldt)	F(0,6) = 3.951	P = 0.003	$\eta_p^2 = 0.248$
Cessation	64.149 (14.905)				
Recovery 15 mins	72.604 (14.393)				
Recovery 30 mins	74.321 (15.943)				
Recovery 45 mins	72.792 (17.076)				
Recovery 60 mins	73.431 (12.960)				
24 Hrs Post Ex	79.054 (15.089)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-12.664	P = 0.001
	R15	-4.209	P = 0.319
	R30	-2.491	P = 0.471
	R45	-4.020	P = 0.340
	R60	-3.382	P = 0.405
	24 Hrs	2.241	P = 0.482

Maximum Voluntary Force Recovery – Sinusoidal (%MVF)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	81.531 (10.090)	χ^2 (9) = 52.158 P = 0.000 ε = 0.369	F(0,6) = 8.381	P = 0.000	η_p^2 = 0.411
Cessation	69.310 (11.362)				
Recovery 15 mins	69.111 (11.397)				
Recovery 30 mins	71.738 (8.837)				
Recovery 45 mins	75.033 (11.678)				
Recovery 60 mins	76.218 (12.825)				
24 Hrs Post Ex	92.951 (17.759)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-12.221	P = 0.011
	R15	-12.420	P = 0.009
	R30	-9.793	P = 0.046
	R45	-6.498	P = 0.211
	R60	-5.313	P = 0.311
	24 Hrs	11.420	P = 0.018

Twitch Force Recovery – Sustained (%MVF)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	5.234 (2.271)	χ^2 (9) = 42.744 P = 0.003 ε = 0.465	F(0,6) = 7.346	P = 0.000	η_p^2 = 0.380
Cessation	3.222 (1.570)				
Recovery 15 mins	3.465 (1.882)				
Recovery 30 mins	3.981 (1.961)				
Recovery 45 mins	3.999 (2.220)				
Recovery 60 mins	3.663 (1.987)				
24 Hrs Post Ex	4.316 (2.390)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-2.012	P = 0.000
	R15	-1.769	P = 0.000
	R30	-1.253	P = 0.002
	R45	-1.236	P = 0.002
	R60	-1.571	P = 0.001
	24 Hrs	-0.918	P = 0.023

Twitch Force Recovery – On/Off (%MVF)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	5.609 (2.639)	χ^2 (9) = 78.649 P = 0.000 ε = 0.383	F(0,6) = 3.047	P = 0.015	η_p^2 = 0.203
Cessation	4.326 (4.152)				
Recovery 15 mins	4.557 (4.055)				
Recovery 30 mins	4.463 (3.966)				
Recovery 45 mins	4.467 (3.519)				
Recovery 60 mins	4.688 (3.910)				
24 Hrs Post Ex	5.866 (4.182)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-1.282	P = 0.030
	R15	-1.051	P = 0.083
	R30	-1.146	P = 0.056
	R45	-1.141	P = 0.057
	R60	-0.921	P = 0.137
	24 Hrs	0.258	P = 0.495

Twitch Force Recovery – 1 Percent (%MVF)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	5.239 (2.526)	χ^2 (9) = 50.446 P = 0.000 ε = 0.429	F(0,6) = 6.113	P = 0.002	η_p^2 = 0.338
Cessation	4.026 (2.155)				
Recovery 15 mins	3.675 (1.681)				
Recovery 30 mins	3.826 (2.138)				
Recovery 45 mins	3.814 (2.160)				
Recovery 60 mins	4.020 (2.526)				
24 Hrs Post Ex	4.769 (2.594)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-1.213	P = 0.002
	R15	-1.564	P = 0.000
	R30	-1.413	P = 0.000
	R45	-1.425	P = 0.000
	R60	-1.219	P = 0.002
	24 Hrs	-0.470	P = 0.266

Twitch Force Recovery – MinMax (%MVF)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	5.067 (2.323)	χ^2 (9) = 53.045 P = 0.000 ε = 0.424	F(0,6) = 1.760	P = 0.091	η_p^2 = 0.128
Cessation	3.848 (2.221)				
Recovery 15 mins	3.993 (1.337)				
Recovery 30 mins	3.979 (1.465)				
Recovery 45 mins	4.105 (1.670)				
Recovery 60 mins	3.796 (1.513)				
24 Hrs Post Ex	4.396 (2.082)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-1.219	P = 0.028
	R15	-1.074	P = 0.057
	R30	-1.088	P = 0.053
	R45	-0.962	P = 0.093
	R60	-1.271	P = 0.021
	24 Hrs	-0.670	P = 0.263

Twitch Force Recovery – Sinusoidal (%MVF)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	4.790 (1.685)	χ^2 (9) = 91.876 P = 0.000 ε = 0.258	F(0,6) = 4.426	P = 0.018	$\eta_p^2 = 0.269$
Cessation	3.755 (1.435)				
Recovery 15 mins	3.748 (1.804)				
Recovery 30 mins	3.999 (1.807)				
Recovery 45 mins	4.058 (2.116)				
Recovery 60 mins	3.965 (1.934)				
24 Hrs Post Ex	5.524 (2.403)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-1.034	P = 0.050
	R15	-1.042	P = 0.048
	R30	-0.790	P = 0.149
	R45	-0.732	P = 0.186
	R60	-0.825	P = 0.130
	24 Hrs	0.734	P = 0.185

Low-Frequency Fatigue Ratio Recovery – Sustained (Ratio)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	0.626 (0.136)	χ^2 (9) = 45.096 P = 0.002 ε = 0.452	F(0,6) = 7.213	P = 0.001	η_p^2 = 0.375
Cessation	0.468 (0.170)				
Recovery 15 mins	0.475 (0.214)				
Recovery 30 mins	0.523 (0.248)				
Recovery 45 mins	0.545 (0.192)				
Recovery 60 mins	0.554 (0.281)				
24 Hrs Post Ex	0.668 (0.192)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-0.158	P = 0.001
	R15	-0.151	P = 0.001
	R30	-0.103	P = 0.023
	R45	-0.081	P = 0.085
	R60	-0.072	P = 0.132
	24 Hrs	0.042	P = 0.387

Low-Frequency Fatigue Ratio Recovery – On/Off (Ratio)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	0.625 (0.161)	χ^2 (9) = 20.922 P = 0.428 ε = 0.577	F(0,6) = 2.993	P = 0.006	$\eta_p^2 = 0.200$
Cessation	0.569 (0.150)				
Recovery 15 mins	0.497 (0.184)				
Recovery 30 mins	0.488 (0.169)				
Recovery 45 mins	0.557 (0.147)				
Recovery 60 mins	0.571 (0.153)				
24 Hrs Post Ex	0.578 (0.172)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-0.058	P = 0.262
	R15	-0.128	P = 0.005
	R30	-0.137	P = 0.002
	R45	-0.067	P = 0.168
	R60	-0.054	P = 0.279
	24 Hrs	-0.047	P = 0.344

Low-Frequency Fatigue Ratio Recovery – 1 Percent (Ratio)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	0.567 (0.151)	χ^2 (9) = 30.237 P = 0.078 ε = 0.566	F(0,6) = 5.127	P = 0.000	$\eta_p^2 = 0.299$
Cessation	0.466 (0.104)				
Recovery 15 mins	0.419 (0.149)				
Recovery 30 mins	0.418 (0.167)				
Recovery 45 mins	0.462 (0.147)				
Recovery 60 mins	0.502 (0.152)				
24 Hrs Post Ex	0.569 (0.144)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-0.101	P = 0.030
	R15	-0.148	P = 0.001
	R30	-0.149	P = 0.001
	R45	-0.106	P = 0.022
	R60	-0.066	P = 0.185
	24 Hrs	0.002	P = 0.500

Low-Frequency Fatigue Ratio Recovery – MinMax (Ratio)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	0.643 (0.188)	χ^2 (9) = 28.897 P = 0.104 ε = 0.566 (Huynh – Feldt)	F(0,6) = 2.339	P = 0.027	η_p^2 = 0.163
Cessation	0.575 (0.261)				
Recovery 15 mins	0.548 (0.199)				
Recovery 30 mins	0.540 (0.220)				
Recovery 45 mins	0.526 (0.201)				
Recovery 60 mins	0.608 (0.257)				
24 Hrs Post Ex	0.672 (0.204)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-0.068	P = 0.294
	R15	-0.095	P = 0.132
	R30	-0.103	P = 0.097
	R45	-0.117	P = 0.480
	R60	-0.068	P = 0.056
	24 Hrs	0.028	P = 0.493

Low-Frequency Fatigue Ratio Recovery – Sinusoidal (Ratio)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	0.610 (0.156)	χ^2 (9) = 55.615 P = 0.000 ε = 0.414	F(0,6) = 0.926	P = 0.241	η_p^2 = 0.072
Cessation	0.602 (0.334)				
Recovery 15 mins	0.539 (0.281)				
Recovery 30 mins	0.522 (0.192)				
Recovery 45 mins	0.515 (0.144)				
Recovery 60 mins	0.526 (0.195)				
24 Hrs Post Ex	0.594 (0.124)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-0.008	P = 0.500
	R15	-0.071	P = 0.357
	R30	-0.088	P = 0.258
	R45	-0.095	P = 0.221
	R60	-0.085	P = 0.277
	24 Hrs	-0.017	P = 0.499

Test Contraction MMG RMS Recovery – Sustained (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	15.797 (10.455)	χ^2 (9) = 44.950 P = 0.002 ε = 0.534 (Huynh-Feldt)	F(0,6) = 7.152	P = 0.000	$\eta_p^2 = 0.373$
Cessation	32.276 (21.208)				
Recovery 15 mins	32.507 (23.656)				
Recovery 30 mins	24.325 (16.043)				
Recovery 45 mins	23.486 (16.670)				
Recovery 60 mins	21.336 (12.376)				
24 Hrs Post Ex	18.200 (18.483)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	16.479	P = 0.000
	R15	16.710	P = 0.000
	R30	8.528	P = 0.034
	R45	7.690	P = 0.060
	R60	5.539	P = 0.196
	24 Hrs	2.403	P = 0.478

Test Contraction MMG RMS Recovery – On/Off (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	16.850 (8.532)	χ^2 (9) = 21.752 P = 0.380 ε = 0.601	F(0,6) = 3.455	P = 0.003	$\eta_p^2 = 0.224$
Cessation	22.568 (15.730)				
Recovery 15 mins	20.863 (12.409)				
Recovery 30 mins	19.709 (9.354)				
Recovery 45 mins	19.337 (11.964)				
Recovery 60 mins	19.572 (9.284)				
24 Hrs Post Ex	15.826 (9.534)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	5.717	P = 0.004
	R15	4.013	P = 0.054
	R30	2.859	P = 0.191
	R45	2.486	P = 0.261
	R60	2.721	P = 0.215
	24 Hrs	-1.024	P = 0.491

Test Contraction MMG RMS Recovery – 1 Percent (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	15.271 (8.871)	χ^2 (9) = 45.818 P = 0.001 ε = 0.442	F(0,6) = 2.967	P = 0.026	η_p^2 = 0.198
Cessation	18.878 (12.102)				
Recovery 15 mins	18.334 (11.184)				
Recovery 30 mins	16.393 (8.041)				
Recovery 45 mins	15.620 (8.276)				
Recovery 60 mins	17.205 (8.781)				
24 Hrs Post Ex	13.834 (5.609)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	3.608	P = 0.037
	R15	3.063	P = 0.083
	R30	1.122	P = 0.467
	R45	0.349	P = 0.500
	R60	1.934	P = 0.297
	24 Hrs	-1.434	P = 0.416

Test Contraction MMG RMS Recovery – MinMax (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	14.121 (5.159)	χ^2 (9) = 62.505 P = 0.000 ε = 0.368	F(0,6) = 4.493	P = 0.001	η_p^2 = 0.272
Cessation	29.275 (15.576)				
Recovery 15 mins	25.206 (10.767)				
Recovery 30 mins	24.355 (15.425)				
Recovery 45 mins	19.415 (10.849)				
Recovery 60 mins	21.285 (18.268)				
24 Hrs Post Ex	15.556 (8.205)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	15.154	P = 0.000
	R15	11.085	P = 0.008
	R30	10.234	P = 0.015
	R45	5.294	P = 0.249
	R60	7.164	P = 0.105
	24 Hrs	1.436	P = 0.499

Test Contraction MMG RMS Recovery – Sinusoidal (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	17.300 (10.467)	χ^2 (9) = 62.505 P = 0.000 ε = 0.368	F(0,6) = 4.493	P = 0.001	η_p^2 = 0.272
Cessation	26.427 (17.900)				
Recovery 15 mins	23.074 (14.064)				
Recovery 30 mins	18.795 (12.545)				
Recovery 45 mins	20.558 (16.077)				
Recovery 60 mins	19.006 (12.537)				
24 Hrs Post Ex	15.152 (9.053)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	9.127	P = 0.002
	R15	5.774	P = 0.062
	R30	1.495	P = 0.491
	R45	3.259	P = 0.319
	R60	1.706	P = 0.483
	24 Hrs	-2.148	P = 0.455

Test Contraction EMG RMS Medial Head Recovery – Sustained (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	18.516 (6.444)	χ^2 (9) = 60.163 P = 0.000 ε = 0.392	F(0,6) = 2.610	P = 0.042	η_p^2 = 0.279
Cessation	26.714 (10.590)				
Recovery 15 mins	24.950 (10.934)				
Recovery 30 mins	23.325 (8.248)				
Recovery 45 mins	22.481 (10.315)				
Recovery 60 mins	21.924 (9.450)				
24 Hrs Post Ex	20.666 (10.421)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	8.198	P = 0.003
	R15	6.434	P = 0.020
	R30	4.808	P = 0.095
	R45	3.965	P = 0.181
	R60	3.408	P = 0.257
	24 Hrs	2.149	P = 0.437

Test Contraction EMG RMS Medial Head Recovery – On/Off (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	16.825 (5.823)	χ^2 (9) = 46.943 P = 0.001 ε = 0.460	F(0,6) = 1.150	P = 0.171	η_p^2 = 0.087
Cessation	18.070 (6.760)				
Recovery 15 mins	18.501 (8.162)				
Recovery 30 mins	17.531 (6.760)				
Recovery 45 mins	17.970 (7.551)				
Recovery 60 mins	17.936 (7.075)				
24 Hrs Post Ex	19.422 (8.963)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	1.245	P = 0.352
	R15	1.676	P = 0.209
	R30	0.707	P = 0.483
	R45	1.146	P = 0.385
	R60	1.112	P = 0.395
	24 Hrs	2.597	P = 0.039

Test Contraction EMG RMS Medial Head Recovery – 1 Percent (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	17.885 (4.757)	χ^2 (9) = 32.037 P = 0.052 ε = 0.549	F(0,6) = 1.087	P = 0.189	η_p^2 = 0.083
Cessation	16.549 (4.904)				
Recovery 15 mins	18.245 (4.955)				
Recovery 30 mins	17.762 (3.988)				
Recovery 45 mins	17.871 (4.119)				
Recovery 60 mins	18.132 (4.038)				
24 Hrs Post Ex	16.479 (5.053)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-1.336	P = 0.291
	R15	0.360	P = 0.500
	R30	-0.123	P = 0.500
	R45	-0.014	P = 0.500
	R60	0.247	P = 0.500
	24 Hrs	-1.406	P = 0.266

Test Contraction EMG RMS Medial Head Recovery – MinMax (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	15.936 (5.815)	χ^2 (9) = 72.623 P = 0.000 ε = 0.362	F(0,6) = 4.697	P = 0.008	η_p^2 = 0.281
Cessation	25.065 (12.124)				
Recovery 15 mins	23.683 (9.784)				
Recovery 30 mins	22.934 (8.268)				
Recovery 45 mins	20.390 (6.458)				
Recovery 60 mins	20.202 (8.270)				
24 Hrs Post Ex	15.297 (6.135)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	9.129	P = 0.001
	R15	7.748	P = 0.006
	R30	6.998	P = 0.014
	R45	4.455	P = 0.143
	R60	4.267	P = 0.163
	24 Hrs	-0.639	P = 0.500

Test Contraction EMG RMS Medial Head Recovery – Sinusoidal (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	18.253 (6.104)	χ^2 (9) = 58.155 P = 0.000 ε = 0.339	F(0,6) = 0.748	P = 0.613	η_p^2 = 0.059
Cessation	21.098 (11.973)				
Recovery 15 mins	19.134 (8.244)				
Recovery 30 mins	19.814 (9.179)				
Recovery 45 mins	19.168 (8.801)				
Recovery 60 mins	19.188 (8.619)				
24 Hrs Post Ex	17.074 (7.325)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	2.844	P = 0.545
	R15	0.881	P = 0.996
	R30	1.561	P = 0.936
	R45	0.915	P = 0.995
	R60	0.935	P = 0.995
	24 Hrs	-1.179	P = 0.982

Test Contraction EMG RMS Lateral Head Recovery – Sustained (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	16.066 (7.126)	χ^2 (9) = 44.852 P = 0.002 ε = 0.427	F(0,6) = 5.232	P = 0.007	η_p^2 = 0.304
Cessation	24.877 (12.523)				
Recovery 15 mins	24.217 (9.947)				
Recovery 30 mins	22.637 (10.123)				
Recovery 45 mins	21.502 (8.443)				
Recovery 60 mins	20.661 (8.670)				
24 Hrs Post Ex	19.068 (7.025)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	8.810	P = 0.000
	R15	8.151	P = 0.000
	R30	6.571	P = 0.003
	R45	5.436	P = 0.013
	R60	4.595	P = 0.040
	24 Hrs	3.002	P = 0.413

Test Contraction EMG RMS Lateral Head Recovery – On/Off (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	17.870 (7.434)	χ^2 (9) = 76.923 P = 0.000 ε = 0.368	F(0,6) = 0.672	P = 0.533	η_p^2 = 0.053
Cessation	16.021 (9.188)				
Recovery 15 mins	19.343 (10.861)				
Recovery 30 mins	18.112 (10.316)				
Recovery 45 mins	18.267 (10.743)				
Recovery 60 mins	20.190 (13.736)				
24 Hrs Post Ex	17.920 (9.336)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-1.848	P = 0.914
	R15	1.473	P = 0.968
	R30	0.242	P = 1.000
	R45	0.397	P = 1.000
	R60	2.320	P = 0.801
	24 Hrs	0.051	P = 1.000

Test Contraction EMG RMS Lateral Head Recovery – 1 Percent (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	14.684 (5.318)	χ^2 (9) = 43.368 P = 0.003 ε = 0.501	F(0,6) = 1.225	P = 0.315	$\eta_p^2 = 0.093$
Cessation	15.550 (6.133)				
Recovery 15 mins	17.413 (6.749)				
Recovery 30 mins	16.753 (5.841)				
Recovery 45 mins	16.968 (5.868)				
Recovery 60 mins	16.836 (5.706)				
24 Hrs Post Ex	16.466 (6.825)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	0.866	P = 0.951
	R15	2.729	P = 0.117
	R30	2.069	P = 0.337
	R45	2.284	P = 0.246
	R60	2.152	P = 0.300
	24 Hrs	1.782	P = 0.486

Test Contraction EMG RMS Lateral Head Recovery – MinMax (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	13.268 (7.033)	χ^2 (9) = 71.795 P = 0.000 ε = 0.397	F(0,6) = 4.047	P = 0.002	$\eta_p^2 = 0.252$
Cessation	25.931 (15.897)				
Recovery 15 mins	22.156 (13.751)				
Recovery 30 mins	20.111 (11.390)				
Recovery 45 mins	18.980 (11.985)				
Recovery 60 mins	19.425 (11.953)				
24 Hrs Post Ex	18.788 (10.416)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	12.663	P = 0.000
	R15	8.888	P = 0.004
	R30	6.843	P = 0.031
	R45	5.712	P = 0.157
	R60	6.157	P = 0.111
	24 Hrs	5.520	P = 0.182

Test Contraction EMG RMS Lateral Head Recovery – Sinusoidal (% MVC)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	16.471 (6.848)	χ^2 (9) = 75.550 P = 0.000 ε = 0.298	F(0,6) = 2.951	P = 0.079	η_p^2 = 0.197
Cessation	23.207 (12.939)				
Recovery 15 mins	21.085 (8.090)				
Recovery 30 mins	21.188 (8.983)				
Recovery 45 mins	19.679 (8.465)				
Recovery 60 mins	19.455 (8.664)				
24 Hrs Post Ex	16.779 (8.084)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	6.736	P = 0.004
	R15	4.614	P = 0.054
	R30	4.717	P = 0.048
	R45	3.208	P = 0.405
	R60	2.984	P = 0.478
	24 Hrs	0.307	P = 1.000

Test Contraction EMG MnPF Lateral Head Recovery – Sustained (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	131.898 (27.091)	χ^2 (9) = 42.287 P = 0.004 ε = 0.537 (Huynh-Feldt)	F(0,6) = 4.408	P = 0.003	η_p^2 = 0.269
Cessation	115.893 (19.055)				
Recovery 15 mins	121.640 (20.232)				
Recovery 30 mins	122.052 (19.887)				
Recovery 45 mins	124.790 (20.002)				
Recovery 60 mins	129.879 (29.184)				
24 Hrs Post Ex	136.369 (22.497)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-16.005	P = 0.003
	R15	-10.258	P = 0.143
	R30	-9.845	P = 0.172
	R45	-7.108	P = 0.470
	R60	-2.018	P = 0.996
	24 Hrs	4.472	P = 0.853

Test Contraction EMG MnPF Lateral Head Recovery – On/Off (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	127.823 (25.514)	χ^2 (9) = 61.967 P = 0.000 ε = 0.407	F(0,6) = 0.456	P = 0.676	$\eta_p^2 = 0.037$
Cessation	121.122 (29.287)				
Recovery 15 mins	124.619 (19.536)				
Recovery 30 mins	125.745 (20.492)				
Recovery 45 mins	124.987 (22.481)				
Recovery 60 mins	121.665 (20.965)				
24 Hrs Post Ex	129.694 (24.973)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-6.022	P = 0.834
	R15	-3.204	P = 0.990
	R30	-2.078	P = 0.999
	R45	-2.836	P = 0.994
	R60	-6.158	P = 0.820
	24 Hrs	1.871	P = 0.999

Test Contraction EMG MnPF Lateral Head Recovery – 1 Percent (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	123.929 (22.559)	χ^2 (9) = 55.958 P = 0.000 ε = 0.519	F(0,6) = 3.487	P = 0.024	η_p^2 = 0.225
Cessation	113.794 (19.726)				
Recovery 15 mins	117.082 (17.935)				
Recovery 30 mins	119.262 (15.719)				
Recovery 45 mins	116.505 (18.009)				
Recovery 60 mins	116.310 (20.205)				
24 Hrs Post Ex	128.225 (20.227)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-10.135	P = 0.024
	R15	-6.847	P = 0.298
	R30	-4.668	P = 0.671
	R45	-7.425	P = 0.226
	R60	-7.619	P = 0.205
	24 Hrs	4.295	P = 0.740

Test Contraction EMG MnPF Lateral Head Recovery – MinMax (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	147.622 (49.829)	χ^2 (9) = 63.026 P = 0.000 ε = 0.416	F(0,6) = 2.749	P = 0.018	η_p^2 = 0.186
Cessation	116.972 (20.835)				
Recovery 15 mins	117.757 (22.432)				
Recovery 30 mins	121.999 (23.389)				
Recovery 45 mins	129.034 (31.479)				
Recovery 60 mins	126.514 (23.599)				
24 Hrs Post Ex	139.177 (38.261)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-30.651	P = 0.006
	R15	-29.865	P = 0.008
	R30	-25.623	P = 0.024
	R45	-18.588	P = 0.236
	R60	-21.108	P = 0.141
	24 Hrs	-8.445	P = 0.891

Test Contraction EMG MnPF Lateral Head Recovery – Sinusoidal (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	130.441 (25.594)	χ^2 (9) = 95.160 P = 0.000 ε = 0.250	F(0,6) = 3.434	P = 0.005	$\eta_p^2 = 0.222$
Cessation	110.849 (29.058)				
Recovery 15 mins	120.569 (28.519)				
Recovery 30 mins	120.570 (39.883)				
Recovery 45 mins	122.375 (31.995)				
Recovery 60 mins	125.721 (37.627)				
24 Hrs Post Ex	134.654 (30.102)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-19.592	P = 0.004
	R15	-9.872	P = 0.358
	R30	-9.871	P = 0.359
	R45	-8.066	P = 0.561
	R60	-4.720	P = 0.922
	24 Hrs	4.213	P = 0.952

Test Contraction EMG MdPF Lateral Head Recovery – Sustained (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	98.432 (12.454)	χ^2 (9) = 29.374 P = 0.094 ε = 0.643	F(0,6) = 4.264	P = 0.001	η_p^2 = 0.262
Cessation	89.405 (8.134)				
Recovery 15 mins	94.712 (10.619)				
Recovery 30 mins	93.586 (10.342)				
Recovery 45 mins	94.079 (10.883)				
Recovery 60 mins	97.978 (12.914)				
24 Hrs Post Ex	99.279 (9.166)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-8.949	P = 0.001
	R15	-3.719	P = 0.423
	R30	-4.845	P = 0.183
	R45	-4.352	P = 0.271
	R60	-0.454	P = 1.000
	24 Hrs	0.848	P = 0.999

Test Contraction EMG MdPF Lateral Head Recovery – On/Off (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	98.083 (29.330)	χ^2 (9) = 87.618 P = 0.000 ε = 0.365	F(0,6) = 0.276	P = 0.781	η_p^2 = 0.022
Cessation	96.952 (33.243)				
Recovery 15 mins	94.426 (13.512)				
Recovery 30 mins	94.300 (14.718)				
Recovery 45 mins	92.309 (14.253)				
Recovery 60 mins	92.037 (12.708)				
24 Hrs Post Ex	98.107 (15.709)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-1.132	P = 1.000
	R15	-3.657	P = 0.989
	R30	-3.785	P = 0.987
	R45	-5.775	P = 0.909
	R60	-6.046	P = 0.890
	24 Hrs	0.024	P = 1.000

Test Contraction EMG MdPF Lateral Head Recovery – 1 Percent (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	93.220 (12.721)	χ^2 (9) = 44.419 P = 0.002 ε = 0.548 (Huynh-Feldt)	F(0,6) = 5.144	P = 0.001	η_p^2 = 0.300
Cessation	84.489 (11.669)				
Recovery 15 mins	90.225 (14.098)				
Recovery 30 mins	89.932 (13.015)				
Recovery 45 mins	87.103 (15.365)				
Recovery 60 mins	88.770 (13.929)				
24 Hrs Post Ex	97.632 (13.270)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-8.732	P = 0.004
	R15	-2.995	P = 0.736
	R30	-3.288	P = 0.658
	R45	-6.12	P = 0.107
	R60	-4.450	P = 0.360
	24 Hrs	4.412	P = 0.368

Test Contraction EMG MdPF Lateral Head Recovery – MinMax (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	107.701 (21.883)	χ^2 (9) = 73.902 P = 0.000 ε = 0.446	F(0,6) = 3.077	P = 0.046	η_p^2 = 0.204
Cessation	88.732 (14.304)				
Recovery 15 mins	91.750 (16.099)				
Recovery 30 mins	94.370 (16.590)				
Recovery 45 mins	95.483 (15.839)				
Recovery 60 mins	95.424 (16.076)				
24 Hrs Post Ex	100.739 (26.040)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-18.968	P = 0.001
	R15	-15.951	P = 0.006
	R30	-13.331	P = 0.024
	R45	-12.218	P = 0.040
	R60	-12.277	P = 0.039
	24 Hrs	-6.962	P = 0.551

Test Contraction EMG MdPF Lateral Head Recovery – Sinusoidal (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	102.283 (25.491)	χ^2 (9) = 87.721 P = 0.000 ε = 0.374	F(0,6) = 5.052	P = 0.000	$\eta_p^2 = 0.296$
Cessation	84.359 (14.185)				
Recovery 15 mins	91.511 (15.065)				
Recovery 30 mins	92.025 (15.651)				
Recovery 45 mins	91.237 (14.735)				
Recovery 60 mins	93.485 (19.292)				
24 Hrs Post Ex	98.167 (13.922)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-17.924	P = 0.000
	R15	-10.772	P = 0.009
	R30	-10.258	P = 0.014
	R45	-11.046	P = 0.008
	R60	-8.798	P = 0.038
	24 Hrs	-4.116	P = 0.720

Test Contraction EMG MnPF Medial Head Recovery – Sustained (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	103.062 (17.863)	χ^2 (9) = 38.434 P = 0.010 ε = 0.433	F(0,6) = 3.975	P = 0.002	η_p^2 = 0.249
Cessation	97.751 (18.203)				
Recovery 15 mins	102.244 (15.919)				
Recovery 30 mins	103.225 (15.886)				
Recovery 45 mins	104.142 (16.641)				
Recovery 60 mins	104.792 (19.305)				
24 Hrs Post Ex	108.500 (16.130)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-5.311	P = 0.101
	R15	-0.818	P = 0.999
	R30	0.163	P = 1.000
	R45	1.080	P = 0.994
	R60	1.731	P = 0.938
	24 Hrs	5.438	P = 0.089

Test Contraction EMG MnPF Medial Head Recovery – Sustained (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	116.014 (28.308)	χ^2 (9) = 69.679 P = 0.000 ε = 0.293	F(0,6) = 1.370	P = 0.238	η_p^2 = 0.102
Cessation	108.757 (19.541)				
Recovery 15 mins	115.736 (20.936)				
Recovery 30 mins	115.139 (21.322)				
Recovery 45 mins	114.345 (18.786)				
Recovery 60 mins	114.014 (18.575)				
24 Hrs Post Ex	109.819 (22.408)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-7.257	P = 0.179
	R15	-0.278	P = 1.000
	R30	-0.875	P = 0.999
	R45	-1.669	P = 0.994
	R60	-2.000	P = 0.984
	24 Hrs	-6.195	P = 0.314

Test Contraction EMG MnPF Medial Head Recovery – 1 Percent (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	108.924 (14.368)	χ^2 (9) = 99.669 P = 0.000 ε = 0.256	F(0,6) = 2.999	P = 0.085	η_p^2 = 0.200
Cessation	103.745 (12.494)				
Recovery 15 mins	107.661 (13.114)				
Recovery 30 mins	109.872 (13.171)				
Recovery 45 mins	107.628 (12.855)				
Recovery 60 mins	107.625 (15.137)				
24 Hrs Post Ex	118.739 (25.136)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-5.179	P = 0.563
	R15	-1.263	P = 0.999
	R30	0.948	P = 0.999
	R45	-1.296	P = 0.999
	R60	-1.298	P = 0.999
	24 Hrs	9.815	P = 0.054

Test Contraction EMG MnPF Medial Head Recovery – MinMax (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	123.926 (38.841)	χ^2 (9) = 121.328 P = 0.000 ε = 0.348	F(0,6) = 2.405	P = 0.050	η_p^2 = 0.167
Cessation	100.308 (20.773)				
Recovery 15 mins	106.709 (18.448)				
Recovery 30 mins	106.312 (17.812)				
Recovery 45 mins	107.686 (14.818)				
Recovery 60 mins	110.395 (16.810)				
24 Hrs Post Ex	120.922 (40.232)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-23.618	P = 0.008
	R15	-17.218	P = 0.127
	R30	-17.614	P = 0.114
	R45	-16.240	P = 0.166
	R60	-13.532	P = 0.319
	24 Hrs	-3.005	P = 0.998

Test Contraction EMG MnPF Medial Head Recovery – Sinusoidal (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	113.202 (19.365)	χ^2 (9) = 35.476 P = 0.022 ε = 0.400	F(0,6) = 1.331	P = 0.282	η_p^2 = 0.100
Cessation	103.022 (18.579)				
Recovery 15 mins	103.710 (31.130)				
Recovery 30 mins	108.951 (21.545)				
Recovery 45 mins	112.986 (27.854)				
Recovery 60 mins	111.622 (21.468)				
24 Hrs Post Ex	110.557 (18.328)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-10.179	P = 0.213
	R15	-9.492	P = 0.273
	R30	-4.251	P = 0.913
	R45	-0.215	P = 1.000
	R60	-1.579	P = 0.999
	24 Hrs	-2.645	P = 0.990

Test Contraction EMG MdPF Medial Head Recovery – Sustained (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	80.220 (10.094)	χ^2 (9) = 22.151 P = 0.358 ε = 0.627	F(0,6) = 4.004	P = 0.002	η_p^2 = 0.250
Cessation	77.429 (8.692)				
Recovery 15 mins	80.596 (8.963)				
Recovery 30 mins	82.352 (9.338)				
Recovery 45 mins	81.176 (8.301)				
Recovery 60 mins	81.965 (9.488)				
24 Hrs Post Ex	82.938 (8.267)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-2.792	P = 0.146
	R15	0.376	P = 0.999
	R30	2.132	P = 0.376
	R45	0.956	P = 0.945
	R60	1.745	P = 0.577
	24 Hrs	2.718	P = 0.165

Test Contraction EMG MdPF Medial Head Recovery – On/Off (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	90.199 (24.779)	χ^2 (9) = 69.602 P = 0.000 ε = 0.321	F(0,6) = 1.361	P = 0.276	$\eta_p^2 = 0.102$
Cessation	84.777 (19.147)				
Recovery 15 mins	89.742 (16.262)				
Recovery 30 mins	88.349 (15.575)				
Recovery 45 mins	88.040 (14.888)				
Recovery 60 mins	89.744 (18.122)				
24 Hrs Post Ex	83.119 (13.013)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-5.422	P = 0.382
	R15	-0.458	P = 1.000
	R30	-1.851	P = 0.985
	R45	-2.159	P = 0.969
	R60	-0.455	P = 1.000
	24 Hrs	-7.081	P = 0.152

Test Contraction EMG MdPF Medial Head Recovery – 1 Percent (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	82.715 (8.993)	χ^2 (9) = 59.250 P = 0.000 ε = 0.382	F(0,6) = 1.717	P = 0.195	η_p^2 = 0.125
Cessation	78.340 (8.458)				
Recovery 15 mins	82.515 (9.483)				
Recovery 30 mins	83.875 (9.503)				
Recovery 45 mins	80.996 (7.926)				
Recovery 60 mins	82.069 (10.040)				
24 Hrs Post Ex	84.628 (11.770)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-4.375	P = 0.214
	R15	-0.199	P = 1.000
	R30	1.160	P = 0.989
	R45	-1.718	P = 0.933
	R60	-0.646	P = 0.999
	24 Hrs	1.913	P = 0.896

Test Contraction EMG MdPF Medial Head Recovery – MinMax (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	86.739 (13.518)	χ^2 (9) = 127.244 P = 0.000 ε = 0.202	F(0,6) = 1.591	P = 0.232	η_p^2 = 0.117
Cessation	76.321 (11.270)				
Recovery 15 mins	81.100 (10.916)				
Recovery 30 mins	82.180 (10.171)				
Recovery 45 mins	82.442 (8.063)				
Recovery 60 mins	83.792 (9.742)				
24 Hrs Post Ex	88.779 (33.060)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-10.418	P = 0.053
	R15	-5.639	P = 0.650
	R30	-4.559	P = 0.815
	R45	-4.298	P = 0.849
	R60	-2.947	P = 0.968
	24 Hrs	2.040	P = 0.995

Test Contraction EMG MdPF Medial Head Recovery – Sinusoidal (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	84.723 (11.748)	χ^2 (9) = 46.997 P = 0.001 ε = 0.405	F(0,6) = 1.497	P = 0.239	$\eta_p^2 = 0.111$
Cessation	75.688 (11.403)				
Recovery 15 mins	85.584 (14.270)				
Recovery 30 mins	87.205 (22.211)				
Recovery 45 mins	85.739 (20.269)				
Recovery 60 mins	85.841 (19.113)				
24 Hrs Post Ex	86.501 (16.260)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-9.035	P = 0.210
	R15	0.861	P = 1.000
	R30	2.482	P = 0.987
	R45	1.016	P = 0.999
	R60	1.118	P = 0.999
	24 Hrs	1.778	P = 0.998

Test Contraction MMG MnPF Recovery – Sustained (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	26.065 (6.084)	χ^2 (9) = 45.973 P = 0.001 ε = 0.418	F(0,6) = 1.665	P = 0.201	$\eta_p^2 = 0.122$
Cessation	22.610 (5.542)				
Recovery 15 mins	25.169 (5.721)				
Recovery 30 mins	23.470 (4.980)				
Recovery 45 mins	23.806 (5.133)				
Recovery 60 mins	23.595 (5.306)				
24 Hrs Post Ex	23.529 (6.517)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-3.455	P = 0.022
	R15	-0.895	P = 0.958
	R30	-2.595	P = 0.195
	R45	-2.258	P = 0.315
	R60	-2.469	P = 0.235
	24 Hrs	-2.536	P = 0.213

Test Contraction MMG MnPF Recovery – On/Off (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	25.077 (5.242)	χ^2 (9) = 21.221 P = 0.410 ε = 0.660	F(0,6) = 2.032	P = 0.072	η_p^2 = 0.145
Cessation	23.055 (5.972)				
Recovery 15 mins	22.292 (5.858)				
Recovery 30 mins	22.209 (5.926)				
Recovery 45 mins	22.037 (6.151)				
Recovery 60 mins	21.561 (4.926)				
24 Hrs Post Ex	22.589 (5.480)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-2.022	P = 0.306
	R15	-2.785	P = 0.040
	R30	-2.868	P = 0.033
	R45	-3.040	P = 0.023
	R60	-3.516	P = 0.008
	24 Hrs	-2.488	P = 0.140

Test Contraction MMG MnPF Recovery – 1 Percent (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	27.475 (7.193)	χ^2 (9) = 53.592 P = 0.000 ε = 0.463	F(0,6) = 1.784	P = 0.172	η_p^2 = 0.129
Cessation	24.687 (5.201)				
Recovery 15 mins	24.695 (4.516)				
Recovery 30 mins	23.513 (4.965)				
Recovery 45 mins	24.282 (5.050)				
Recovery 60 mins	22.580 (4.436)				
24 Hrs Post Ex	25.872 (9.694)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-2.788	P = 0.371
	R15	-2.780	P = 0.374
	R30	-3.962	P = 0.048
	R45	-3.194	P = 0.123
	R60	-4.895	P = 0.012
	24 Hrs	-1.603	P = 0.849

Test Contraction MMG MnPF Recovery – MinMax (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	23.645 (5.883)	χ^2 (9) = 58.007 P = 0.000 ε = 0.350	F(0,6) = 0.320	P = 0.739	η_p^2 = 0.026
Cessation	22.305 (7.268)				
Recovery 15 mins	22.567 (6.510)				
Recovery 30 mins	23.105 (6.640)				
Recovery 45 mins	24.945 (6.402)				
Recovery 60 mins	23.956 (6.172)				
24 Hrs Post Ex	24.130 (11.740)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-1.341	P = 0.983
	R15	-1.078	P = 0.994
	R30	-0.540	P = 0.999
	R45	1.300	P = 0.985
	R60	0.311	P = 1.000
	24 Hrs	0.485	P = 0.999

Test Contraction MMG MnPF Recovery – Sinusoidal (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	23.816 (6.753)	χ^2 (9) = 52.202 P = 0.000 ε = 0.506	F(0,6) = 2.607	P = 0.066	η_p^2 = 0.178
Cessation	21.959 (5.591)				
Recovery 15 mins	25.872 (7.917)				
Recovery 30 mins	22.082 (6.166)				
Recovery 45 mins	21.875 (6.243)				
Recovery 60 mins	22.118 (5.827)				
24 Hrs Post Ex	21.902 (5.209)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-1.858	P = 0.543
	R15	2.055	P = 0.441
	R30	-1.734	P = 0.610
	R45	-1.941	P = 0.499
	R60	-1.698	P = 0.630
	24 Hrs	-1.914	P = 0.513

Test Contraction MMG MdPF Recovery – Sustained (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	22.076 (8.508)	χ^2 (9) = 37.732 P = 0.012 ε = 0.436	F(0,6) = 1.350	P = 0.276	η_p^2 = 0.101
Cessation	19.344 (7.301)				
Recovery 15 mins	23.134 (8.173)				
Recovery 30 mins	20.815 (6.481)				
Recovery 45 mins	21.218 (7.203)				
Recovery 60 mins	20.747 (7.133)				
24 Hrs Post Ex	19.385 (7.519)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-2.732	P = 0.380
	R15	1.058	P = 0.972
	R30	-1.262	P = 0.938
	R45	-0.858	P = 0.990
	R60	-1.329	P = 0.923
	24 Hrs	-2.691	P = 0.395

Test Contraction MMG MdPF Recovery – On/Off (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	22.392 (6.255)	χ^2 (9) = 33.248 P = 0.039 ε = 0.649 (Huynh-Feldt)	F(0,6) = 1.486	P = 0.195	$\eta_p^2 = 0.110$
Cessation	19.232 (6.544)				
Recovery 15 mins	18.328 (6.839)				
Recovery 30 mins	19.275 (7.415)				
Recovery 45 mins	19.059 (7.248)				
Recovery 60 mins	18.219 (6.114)				
24 Hrs Post Ex	18.408 (6.226)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-3.160	P = 0.251
	R15	-4.065	P = 0.042
	R30	-3.117	P = 0.132
	R45	-3.334	P = 0.104
	R60	-4.173	P = 0.035
	24 Hrs	-3.985	P = 0.046

Test Contraction MMG MdPF Recovery – 1 Percent (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	25.496 (9.247)	χ^2 (9) = 46.715 P = 0.001 ε = 0.471	F(0,6) = 1.783	P = 0.172	$\eta_p^2 = 0.129$
Cessation	21.592 (5.947)				
Recovery 15 mins	21.383 (6.147)				
Recovery 30 mins	20.082 (6.052)				
Recovery 45 mins	20.774 (6.412)				
Recovery 60 mins	19.035 (4.325)				
24 Hrs Post Ex	22.453 (11.120)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-3.904	P = 0.299
	R15	-4.113	P = 0.252
	R30	-5.415	P = 0.036
	R45	-4.722	P = 0.073
	R60	-6.461	P = 0.011
	24 Hrs	-3.043	P = 0.546

Test Contraction MMG MdPF Recovery – MinMax (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	19.435 (7.013)	χ^2 (9) = 63.900 P = 0.000 ε = 0.384	F(0,6) = 0.729	P = 0.510	η_p^2 = 0.057
Cessation	17.869 (6.779)				
Recovery 15 mins	17.916 (6.499)				
Recovery 30 mins	20.536 (7.309)				
Recovery 45 mins	22.685 (8.196)				
Recovery 60 mins	21.093 (7.694)				
24 Hrs Post Ex	20.995 (14.441)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-1.566	P = 0.998
	R15	-1.518	P = 0.990
	R30	1.102	P = 0.998
	R45	3.251	P = 0.751
	R60	1.658	P = 0.984
	24 Hrs	1.560	P = 0.988

Test Contraction MMG MdPF Recovery – Sinusoidal (Hz)

Condition	Mean (SD)	Mauchly's Test	F Value	P	Partial Eta Squared η_p^2
Baseline	21.229 (9.384)	χ^2 (9) = 41.383 P = 0.005 ε = 0.557 (Huynh-Feldt)	F(0,6) = 2.776	P = 0.028	η_p^2 = 0.188
Cessation	19.259 (6.594)				
Recovery 15 mins	24.484 (8.831)				
Recovery 30 mins	18.463 (6.708)				
Recovery 45 mins	18.533 (6.829)				
Recovery 60 mins	18.839 (6.822)				
24 Hrs Post Ex	17.871 (5.703)				

Condition (A)	Condition (B)	Mean Difference	P
Baseline	Cessation	-1.970	P = 0.821
	R15	3.255	P = 0.375
	R30	-2.766	P = 0.538
	R45	-2.696	P = 0.564
	R60	-2.391	P = 0.676
	24 Hrs	-3.358	P = 0.344

References

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