# Localized Muscle Fatigue: Theoretical and Practical Aspects in Occupational Environments

Ehsan Rashedi

Dissertation submitted to the faculty of

Virginia Polytechnic Institute and State University

in partial fulfillment of the requirements for the degree of

Doctor of Philosophy
in
Industrial and Systems Engineering

Maury A. Nussbaum, Chair
Nathan K. Lau
Michael L. Madigan
Ting Xia

December 3, 2015

Blacksburg, Virginia

Keywords: Localized Muscle Fatigue; Recovery; Modeling; History Dependency; Performance

Localized Muscle Fatigue: Theoretical and Practical Aspects in Occupational Environments

Ehsan Rashedi

ABSTRACT

Localized muscle fatigue (LMF) is a complex, multifactorial phenomenon that involves exercise-induced decrements in the ability to generate force or power. LMF can adversely affect performance and may increase the risk of work-related musculoskeletal disorders (WMSDs), and is thus of contemporary occupational relevance. Despite considerable progress in understanding and predicting muscle fatigue, there are many uncertainties and unresolved issues that are principally associated with the physiological complexity of LMF and the diverse mechanisms that underlie LMF development. This research thus aimed to address some of the theoretical and practical issues related to muscle fatigue and recovery. Regarding the theoretical aspects, two specific muscle fatigue models (MFMs) were directly compared and some important differences in their predictions were identified. These differences were used, in part, as a basis for developing testable hypotheses and designing associated experiments. Further theoretical evaluations were conducted to explore the sensitivity of these models to the model parameters and their ability to predict endurance time in both prolonged and intermittent exertions. Sensitivity to inherent model parameters was quantified, which was relatively high in conditions involving lower to moderate levels of effort. Further assessments indicated substantial variability related to model recovery parameters, which might be related to the inability of these MFMs in simulating the recovery process. From a practical viewpoint, the effect of cycle time on the development and consequences of LMF was determined during intermittent isometric exertions. A shorter cycle time led to less fatigue development as reflected by rates of change in perceived discomfort, performance, and muscle capacity. Lastly, the dependency of muscle recovery on these different histories of fatiguing muscle contractions was explored. How a muscle recovers appeared to depend only on the state from which it starts to recover, though not the exertion history that led to that state. In summary, results of these studies may help in enhancing our understanding of fatigue and recovery processes, and in improving existing models of muscle fatigue and recovery. More accurate predictions of LMF development may help in enhancing muscle performance and in reducing the risk of musculoskeletal injuries and their associated healthcare costs.

## Dedication

He always believed in me and encouraged me toward my goals. Death took his body, but he will remain in my heart forever. To the memory of my dad, Gholamhossein Rashedi, who passed on the values of hard work, and respect for education.

**Acknowledgements**

First, I would like to sincerely and gratefully thank my advisor, Dr. Nussbaum, for his guidance, support, admirable personality, and most importantly, his mentorship during my graduate studies at Virginia Tech. Thank you for all the knowledge you shared with me over the years. I would also like to thank my current committee members, Dr. Lau, Dr. Madigan, and Dr. Xia, also former committee members, Dr. Agnew, and Dr. Lockhart for their comments, and feedback on this work.

In addition, I would like to thank the members of the Industrial Ergonomics and Biomechanics Lab for their friendship and support, in particular my officemate Dr. Kim for his advices and willingness to help. I am so thankful to Randy Waldron, Will Vest, and all the faculty and staff in the ISE department for their help over the past few years.

This journey would not have been possible without the support of my friends. I have many happy memories from the past few years of living in Blacksburg with all of you. We were like a family and I am so grateful to have you all, Behrouz T., Mohamadsaeed D., Ali N., Hodjat P., Mohammad B., Alireza S., Farid J., Hesam B., Alireza K., Mojtaba T., Masoud A., and so many others.

To my family: You encouraged me to follow my goal of pursuing a PhD. Thank you mom for your unconditional love. To my father and mother-in-law, thank you for all your support and kindness during the past years.  Thanks to my brother, sister, sister-in-law, and their families. And to my wife Nasibeh: this academic career would not have been possible without your love, patience, support, sacrifice, and understanding. You have inspired me with your dedication, drive, and positive outlook.

Table of Contents

## List of Figures

## List of Tables

# 1. Introduction

## 1.1. Localized muscle fatigue

In the context of muscular activity/exertion, localized muscle fatigue (LMF) is described/defined as “a loss of maximal force generating capacity” (Bigland-Ritchie et al., 1986; Gandevia et al., 1995), or “failure to maintain the required or expected force” (Edwards, 1981). This definition is associated with the notion of endurance time (ET) and an inability to sustain a given task (aka “task failure”). However, diverse aspects of the neuromuscular system are altered by muscular activity, and such alterations occur prior to ET or task failure. Therefore, LMF has been more specifically defined as “any exercise-induced reduction in the ability of a muscle to generate force or power, regardless of the ability to sustain the task” (Bigland-Ritchie & Woods, 1984). LMF is a complex, multifactorial phenomenon that is widely used as an indicator of physiological processes, since such fatigue leads to a decline in desired performance and muscle force production capacity during various exertions in occupational settings (Chaffin, 1973), and more generally during diverse activities involving voluntary muscle force generation (Chaffin et al., 1991; Gates & Dingwell, 2008; Vøllestad, 1997). Symptoms of LMF include subjective and objective changes, such as an increased perceived exertion, reduced strength, discomfort or a sensation of fatigue, diminished neuromuscular control, muscle tremors, and altered electromyographic (EMG) signals. In contrast to LMF, whole-body fatigue results from task-related metabolic demands involving multiple muscles or muscle groups. In extreme circumstances, such demands could exceed the cardiovascular system capacity (Chaffin, 1973).

LMF has often been described based on the limiting conditions, in particular, the duration that a task or a set of demand can be continued (i.e., ET). Similarly, exhaustion, and the resultant task failure is well characterized for prolonged static exertions. However, the ability of predicting LMF under a range of task characteristics (i.e., intensity, duration, and speed of exercise) is yet to be developed. Facilitating such predictions – a primary focus of our going work and the emphasis in this research – is imperative because of the potential for LMF to adversely affect performance and to increase the risk of work-related musculoskeletal disorders (WMSD). A worker's physical performance is important since it directly impacts the efficiency of the task and the effectiveness of the outcome. Additionally, WMSDs represents a potentially costly issue for employers since they can result in long-term disability, lost manpower hours, and compensation payouts.

### 1.2. Mechanisms and measurement of LMF

Muscle fibers and the motoneurons that innervate them comprise the basic functional unit known as a motor unit (MU). Diverse processes occur in the central nervous system (CNS) that activate motoneurones (Enoka, 1995), thereby leading to action potential generation (Gandevia et al., 1995). Therefore, to produce muscular force several sequential processes are required, each of which may act as a potential limiting factor and contributor to fatigue (Place et al., 2010; Vøllestad, 1997). These processes are categorized in two major types, namely central and peripheral mechanisms, both may occur as a result of either maximal or submaximal contractions (Fitts, 1994; Taylor & Gandevia, 2008; Taylor et al., 2006).

Central fatigue, defined as an exercise-induced degradation of the muscle voluntary activation (Gandevia et al., 1995), involves all central factors (i.e., mechanisms located in the CNS) that activate motoneurones (Enoka, 1995), including motivational factors and integration of sensory information. This activation transmits along the motor neuron and leads to action potential generation in the sarcolemma (Gandevia et al., 1995). Reduced MU recruitment and impaired firing rate modulation during a fatiguing exercise are among the related mechanisms that contribute to central fatigue (Bigland-Ritchie et al., 1983; Löscher et al., 1996; Peters & Fuglevand, 1999). While central fatigue is considered a contributing factor to LMF (Avela et al., 2001), some reports describe peripheral fatigue as being more substantial in LMF development and requiring a longer duration for recovery. In fact, peripheral fatigue can contribute to more than 80% of muscular fatigue (Peters & Fuglevand, 1999; Søgaard et al., 2006).

Peripheral fatigue involves mechanisms such as neuromuscular transmission, muscle action potential propagation, excitation-contraction coupling, and contractile elements in muscles (Bigland-Ritchie, 1981; Gandevia, 2001). Neuromuscular transmission (i.e., transformation of the nerve action potential into the muscle action potential) takes place at the neuromuscular junction. Several fatigue-related factors such as neurotransmitter depletion, reduced neurotransmitter release, and insufficient propagation of the nerve potential at the nerve ending can impair this transmission.

As several diverse mechanisms appear to play a role in LMF, there are a number of approaches that have been used to identify or quantify the presence of LMF. These approaches can be broadly categories as direct and indirect measures (Vøllestad, 1997). By definition (as noted above), a direct method for assessing LMF is by measuring maximum force or power, typically using either maximum voluntary contractions (MVCs) or electrical stimulation of a muscle or nerve. Measuring tetanic force or responses to brief electrical stimulations (i.e., twitch force) are common examples of the latter method. In one approach, a disproportionate force decline in the presence of muscular fatigue has been demonstrated in response to low-frequency stimulation, a phenomenon also known as low-frequency fatigue (LFF). LFF has been linked to repetitive submaximal contractions and high intensity exercises (Bigland-Ritchie et al., 1986). Comparing the results of muscle electrical stimulation and MVC outcomes can provide an estimation of central fatigue, since force generation during stimulated responses is independent of the CNS drive. In contrast to the direct assessment methods of LMF, several indirect, objective measures have been used to assess LMF development, including alterations in the EMG signal, changes in performance (e.g., production rate and work accuracy), ET, and twitch interpolation (TI). Moreover, indirect subjective measures such as ratings of perceived discomfort (for example, using Borg's CR-10 scale (Borg, 1990)) have also been used widely to identify and quantify LMF.

### 1.3. Effects of LMF on performance

Repetitive and/or sustained muscle exertions can lead to LMF, which in turn may decrease the performance of workers in a variety of ways. For example, fatigue-induced declines in performance can result from compromised coordination capabilities (Billaut et al., 2005; Corcos et al., 2002; Sparto et al., 1997) and longer reaction times (Häkkinen &

Komi, 1986; Wilder et al., 1996). More specifically, reaction time is slowed due to higher activity in the prefrontal areas of the brain induced by fatigue (van Duinen et al., 2007), and muscle response time increases as a result of decreasing muscle fiber conduction velocity (Eberstein & Beattie, 1985). Furthermore, increased muscle tremors (Hunter & Enoka, 2003), and impaired proprioceptive ability (Björklund et al., 2000; Carpenter et al., 1998; Chaffin, 1973; Pedersen et al., 1999; Skinner et al., 1986)—especially the ability to sense joint angle—represent other important effects of LMF related to human performance. A compromised sense of joint angle could affect other aspects of worker performance, such as the ability to accurately reproduce a movement (Jaric et al., 1999). LMF has also been reported to lead to impairments in one's sense of effort (Enoka & Stuart, 1992); in other words, LMF can alter an individual's judgment about the magnitude of the exerted force, which could directly impact motor performance.

Although muscle fatigue can negatively impact muscle timing (Strang & Berg, 2007; Wilder et al., 1996) and coordination (Corcos et al., 2002; Gorelick et al., 2003) — thereby contributing to compromising task performance — there is also evidence that humans can adapt to fatigue and still maintain overall performance (Gates & Dingwell, 2008). Other studies have shown an ability of participants to achieve an overall task goal by changing neural control, as well as by increasing muscle cocontraction and joint stiffness (Lucidi & Lehman, 1992; Potvin & O'Brien, 1998; Selen et al., 2007). It must be noted, however, that cocontraction in and of itself will increase muscle activity, and thereby contributing to fatigue development. Despite a number of reported complexities associated with LMF and task performance, the available literature as a whole suggests a causal relationship between LMF and performance declines for a variety of tasks.

### 1.4. LMF and work-related musculoskeletal disorders

LMF has received increasing attention among ergonomists, and has been recognized as an important measure in research and interventions aimed at reducing musculoskeletal disorders (MSDs) in the workplace. Its specific role in the development of MSDs is not clear yet, and the existence of LMF does not necessarily imply an increased risk of WMSDs (Mathiassen & Winkel, 1992). However, factors such as working posture and sustained muscle contraction that are closely related with muscle fatigue have been shown to contribute to soft tissue injuries (Sommerich et al., 1993; K Veiersted, 1994). Furthermore, several theories and conceptual model have been proposed describing the potential links between LMF and WMSD-related mechanisms (Armstrong et al., 1993; Forde et al., 2002; Sejersted & Vøllestad, 1993). For example, fatigue caused by prolonged static postures may lead to muscular imbalances, both with underused and overused muscles.

Other occupational exposures, such as from tasks that involve high levels of repetition, are also correlated with LMF. Such motions are reported to increase the pressure around peripheral nerves and initiate chronic nerve compression, as well as cause subsequent swelling and impairment of the vascular supply (Byl et al., 1997; Forde et al., 2002; Stauber, 2004). Sustained low-level isometric contractions recruit the same low threshold MUs according to the size principle (Henneman et al., 1965). It is notable that each time an MU is activated, it generates at least 30% of its maximal capacity (Sjøgaard & Jensen, 1997), even when the overall task demand is less than 30% MVC. Considering contractions that are sustained for a long period of time, it is expected to have some overloaded muscle fibers that are susceptible to loss of calcium homeostasis. This phenomena is known as the “Cinderella” response of MUs (Sjøgaard & Jensen, 1997), that can occur with or following to fatigue, and has been linked to soft tissue inflammation (Lovlin et al., 1987). Another injury mechanism that can occur in presence of LMF involves ischemia-reperfusion (Appell et al., 1999). For example, tasks that cause intermittent blood flow blockage, any subsequent reperfusion is ischemic tissue can be responsible for an inflammatory response in muscle with potential contribution to WMSDs.

Support for the above theories comes from several causal mechanisms associated with LMF as a potential contributor to WMSD (Armstrong et al., 1993; Kumar, 2001; Mathiassen & Winkel, 1992; NIOSH, 2001). For example, LMF can put workers at risk for a variety of disorders, including chronic rotator cuff tendonitis (Bjelle et al., 1981; Herberts et al., 1984), trapezius myalgia (Veiersted et al., 1993), and cervicobrachical disorder (Maeda, 1977). These and other studies indicate that occupational tasks with exposures that are closely related to fatigue (e.g., sustained muscle tension, non-neutral body posture, and highly repetitive motions) may contribute to the development of muscular problems (Bernard et al., 1997; Björkstén et al., 2001; Forde et al., 2002; Malchaire et al., 2001; Sommerich et al., 1993; Veiersted, 1994), and which in turn may heighten the risk for injury. Furthermore, inadequate recovery time could result in accumulated fatigue that may lead to subsequent health problems, particularly tissue disorders (Byström & Fransson-Hall, 1994; Hagberg, 1984; Kumar, 2001; Sommerich et al., 1993; Vøllestad & Sejersted, 1988). In summary, these pieces of evidence support the role of LMF as at least a surrogate measure of injury risk, and possibly in causal relationship with WMSDs.

### 1.5. Background on muscle fatigue modeling and research needs

Despite considerable progress in understanding and predicting muscle fatigue, there are many uncertainties and unresolved issues (Enoka, 1995)—principally associated with the physiological complexity of muscle fatigue (Bigland-Ritchie et al., 1995) and the diverse mechanisms that underlie the initiation and development of LMF (Allen et al., 2008; Bigland-Ritchie et al., 1995; Enoka, 1995; Fitts, 1994; Gandevia, 2001; Kirkendall, 1990; Place et al., 2010; Westerblad et al., 1998). The specific impact(s) of these various mechanisms (see next section) on muscle performance impairment is not well understood (Vøllestad et al., 1988), and these impacts can vary according to task characteristics (Bigland-Ritchie et al., 1995). Therefore, it is impossible to introduce a unique mechanism responsible for the decline in force generating capacity in all types of exertions (Barry & Enoka, 2007; Fitts, 1994). Any prediction of human muscle fatigue, such as using a model as described below, is likely to have inevitable limitations; ideally, though, such a model should account for a range of specific task parameters or conditions. Of particular interest for ergonomic applications are task type and frequency, load magnitude, and personal factors as they related to individual differences in fatigue.

### 1.5.1. Major factors affecting fatigue/endurance

A number of studies have identified task type/parameters and individual differences as important factors affecting the development of LMF. Prolonged static/dynamic and sustained/intermittent tasks, as well as task parameters such as intensity, duration, cycle time, and work-rest proportion (indicates by duty cycle), are among the former factors (Horton et al., 2012; Iridiastadi & Nussbaum, 2006; Nussbaum et al., 2001; Yassierli & Nussbaum, 2009). Greater effort level and duty cycle have been found to have consistent effects involving higher rates of LMF development and/or decreased ETs (Björkstén & Jonsson, 1977; Jørgensen et al., 1988). More specifically, a greater effort level results in higher demand on each recruited muscle fiber and more recruitment of MUs, both leading to a faster rate of muscular fatigue.

Quantitatively, a curvilinear relationship has been recognized for several decades between muscle contraction level and ET (Rohmert, 1960). Low intensity tasks can be sustained for a long duration; in contrast, ET exponentially (or as a power function) decreases with higher levels of muscle contraction. Further, a lower duty cycle (i.e., the ratio between the contraction period and the cycle time) results in a longer ET (Björkstén & Jonsson, 1977; Kahn & Monod, 1989). Cycle time may also impact the development of LMF, but apparently with less substantial effects compared with effort level (Iridiastadi & Nussbaum, 2006; Yassierli & Nussbaum, 2009). In fact, no consensus has been reached about the benefits of shorter cycle times if other aspects of a task are kept consistent. While prolonged static exertions can be harmful to workers, high repetitions with very short cycle time can also be unfavorable (Silverstein et al., 1986; Sommerich et al., 1993). In contrast, some studies indicate that shorter cycle time is linked to slower development of LMF (Petrofsky et al., 2000; Yassierli & Nussbaum, 2009).

More generally, effort level, exertion duration, repetitiveness, task type, and interactive effects among these are related, in the occupational context, to the concept of job rotation. Job rotation has the potential to alter many of the noted factors during a job shift, by reducing LMF and the risk of injury. For example, performing shoulder abduction was reported to be more fatiguing comparing to rotating between shoulder abduction and flexion (Raina & Dickerson, 2009). However, existing studies have shown inconsistent effects of job rotation on physical demands (Horton et al., 2012; Kuijer et al., 2005) or rates of WMSD occurrence (Aptel et al., 2008).

LMF development can be influenced by the specific muscle(s) or muscle group(s) involved, or more generally the joint involved in force/moment generation. Frey Law and Avin (2010) reviewed a large number of studies involving isometric tasks performed until volitional failure, and found that endurance times (and hence fatigue rate) is dependent on both contraction intensity and joint. These authors concluded that no single generalized fatigue model can adequately represent most individual joints.

As noted, LMF has also been investigated in terms of individual differences, with age and gender being two important aspects of interest (Callahan & Kent-Braun, 2011; Clark & Manini, 2003; Hicks et al., 2001; Hunter, 2009). Age-related differences are of particular interest given the increasing proportion of older workers (Harrigan, 2004). Among the potential reasons for these differences, a loss of muscle fibers in the elderly, especially Type II muscle fibers (i.e., fast twitch) has been reported (Lexell et al., 1983). Consistent with this age-related changes of muscle fiber composition, as well as altered contractile properties and associated loss of strength, some studies have reported a lower rate of LMF development (i.e., higher fatigue resistance) among older adults at the same normalized effort level (Adamo et al., 2009; Allman & Rice, 2002; Ditor & Hicks, 2000; Yassierli et al., 2007). In contrast, other studies found the opposite effect (Baudry et al., 2007; Petrella et al., 2005), and yet others similar fatigue resistance for different age groups (Bemben et al., 1996; Lindström et al., 1997). Differences in study design and the type of task involved have been suggested to explain the differences in outcomes. More specifically, an age-related increase in fatigue resistance is commonly observed with isometric contractions, while contrary results are shown more often with dynamic contractions. Indeed, systematic reviews have indicated that contraction mode is a major factor influencing fatigue resistance with aging (Avin & Frey Law, 2011; Christie et al., 2011). Thus, the effect of age on LMF mechanisms should be considered with respect to the specific task conditions involved (Allman & Rice, 2002). Aside from the noted discrepancies regarding age-related effects on LMF for fixed levels of relative effort, aging typically leads to a substantial decline in strength (Frontera et al., 1991; Johnson, 1982), due to a loss of muscle mass and deficiencies in central activation of muscle (Stackhouse et al., 2001).

Gender is another personal factor related to LMF (Hicks et al., 2001). Males typically have a higher proportion of Type II muscle fibers (Miller et al., 1993), and the recruitment of these muscle fibers is higher among males. Consistent with this, males generally develop LMF more rapidly than females (Avin et al., 2010; Hunter, 2009). However, the magnitude of gender differences in fatigue resistance depends on what muscle group is involved (Avin et al. (2010). These findings have important implications for ergonomic applications that seek to minimize the extent of LMF for a range of workers, and predicting the effects of such individual differences may help achieve higher productivity and less risk of WMSD occurrence.

### 1.5.2. History dependency of localized muscle fatigue

Diverse sources of variability can influence the development of LMF during task execution. Important individual differences include anthropometry, age, gender, fiber type distribution, and fitness status. Specific tasks demands can also be highly influential, on both LMF and LMF-induced changes in task performance. Extensive existing evidence has assessed the influence of task parameters such as the intensity and duration of work, as examples of factors affecting physical exposure (Horton et al., 2012; Iridiastadi & Nussbaum, 2006; Nussbaum et al., 2001; Yassierli & Nussbaum, 2009). A majority of previous studies on muscle fatigue have involved prolonged static contractions, with well-described relationships between effort level and endurance times (El ahrache et al., 2006; Frey Law & Avin, 2010). However, this type of loading has relatively low occupational relevance, as most work tasks have intermittent resting periods (Adamo et al., 2009; Iridiastadi & Nussbaum, 2006). Broadly, intermittent contractions can be characterized based on three task parameters: the exertion level (EL); duty cycle (DC: the ratio between the exertion period and the CT); and cycle time (CT).

Consistently, higher ELs and DCs have been found to cause higher rates of LMF development and/or decreased endurance times (Björkstén & Jonsson, 1977; Jørgensen et al., 1988), likely a direct result of the increased total effort generated over a given time interval.

In contrast to the clear effects of EL and DC during intermittent efforts, the influence of CT on LMF development is less clear. However, the extremes of cycle time, specifically very short and long values, are not favorable. Tasks with high levels of repetition are reported to increase the pressure around peripheral nerves, initiate chronic nerve compression, and cause subsequent swelling and impairment of the vascular supply (Byl et al., 1997; Forde et al., 2002; Stauber, 2004). In contrast, sustained, low-level, isometric contractions recruit the same low threshold motor units (MUs) according to the size principle (Henneman et al., 1965). Notably, each time that a MU is activated it generates at least 30% of its maximal capacity (Sjøgaard & Jensen, 1997), even when the overall task demand is less than 30% of capacity. Considering contractions that are sustained for a long period, it is expected to have some overloaded muscle fibers that experience a loss of calcium homeostasis, or the "Cinderella" response of MUs (Sjøgaard & Jensen, 1997). This phenomenon can occur with or following fatigue, and has been linked to soft tissue inflammation (Lovlin et al., 1987). As such, while prolonged static exertions can be harmful, high repetitions with very short cycle times may also be unfavorable (Silverstein et al., 1986; Sommerich et al., 1993). In between these two extremes, a range of CTs can be considered, and is a potential design parameter for occupational tasks. A practical question remains: given a fixed "amount" of workload that has to be generated intermittently for a specific duration (aka the "tension-time product"), how the work and rest should be distributed? In other words, should frequent short rest breaks be given or longer infrequent ones?

Some studies have linked shorter CTs (i.e., more task variation) with a slower development of LMF (Dickerson et al., 2015; Petrofsky et al., 2000; Yassierli & Nussbaum, 2007), while others provided evidence in favor of longer CTs (Byström et al., 1991; Westgaard, 1988). Still other studies (Engström et al., 1999; Moore & Wells, 2005) have indicated that the effect of CT was not significant at controlled levels of EL and DC. As such, the specific effects of cycle time on muscular fatigue development remain unclear. Previous work on intermittent exertions has investigated CT effects for several intermittent tasks, but using biomechanically complex systems such as the shoulder joint (Dickerson et al., 2015; Iridiastadi & Nussbaum, 2006; Mathiassen, 1993). This complexity has potential effects on LMF monitoring, which may lead to unclear or inconsistent outcomes. For example, there may be changes in the synergistic or antagonistic contraction of multiple muscles. To our knowledge, the influence of CT during low to moderate level of exertions has not yet been reported for a biomechanically "simple" joint, for which the specific effects of CT might be more easily identifiable. This study thus investigated the effect of CT for such a simple system, using both subjective and objective measures to assess the development and consequences of LMF. Based on the weight of existing evidence, and the noted adverse physiological effects of sustained contractions, it was hypothesized that under conditions of similar workload a longer CT will negatively influence perceived discomfort, performance, and muscle capacity.

### 1.5.3. Muscle fatigue and the recovery process

Subsequent to a decrease in muscular strength during exercise, the recovery is initially rapid and with full recovery completed within a relatively short period of time or up to few days, depending upon the experimental protocol (Edwards et al., 1977; Fitts, 1994; Sahlin & Ren, 1989). As mentioned before, inadequate recovery could result in accumulated fatigue that may lead to subsequent health problems. Regarding existing work-rest allowance models, endurance time and recovery models have been used with the end-goal of controlling fatigue development.

Given the diverse approaches used in building existing muscle fatigue models, it is not surprising that substantial discrepancies in work-rest allowances have been reported (Khalid El ahrache and Imbeau (2009). Information to guide the selection of the most appropriate rest allowance model is lacking. Earlier models were based on experimental data and did not consider underlying physiological aspects. In more recent theoretical models (James & Green, 2012; Liu et al., 2002; Xia & Frey Law, 2008), although physiological processes have been partially incorporated in model development, potential effects of muscle work history were not considered. These models, though, have not explicitly considered task-dependency of muscle fatigue development. They only consider the current muscle "state" (force generation capacity), ignoring the "path" taken or loading history leading to a given level of force generation capacity. As such, these models do not predict different recovery patterns subsequent to different muscle exertions. Considering this history-dependency of recovery has the potential to further enhance recovery modeling, and thus facilitate better modeling, especially for intermittent contraction.

### 1.5.4. Muscle fatigue modeling approaches

LMF is influenced by many potential factors with different impacts under possible conditions. Although LMF quantification is essential for many reasons such as work-rest scheduling, task assessment, and assessing an individuals' physical capacity for work, it is not practical to measure LMF under all possible situations. Even for a given situation or task, directly measuring LMF can be time consuming and costly. As such, the use of muscle fatigue models (MFMs) to predict muscle fatigue has broad potential application. These models can be categorized into two groups, i.e., empirical and theoretical, and existing work on each is summarized below.

#### 1.5.4.1. Empirical muscle fatigue models

Empirical MFMs are based on empirical observations and fitting experimental data, rather than mathematical relationships between underlying system parameters. The origin of this type of MFMs is often credited to Rohmert (1960) and Monod and Scherrer (1965), with substantial subsequent expansion or modification. These models are based on either ET (El ahrache et al., 2006) or on decreases in strength during successive work cycles, with some approaches considering cycle time, submaximal force level, and duty cycle as modifying task parameters (Roman-Liu et al., 2005; Wood et al., 1997). A principal advantage of these models is the ability to modify them for a specific situation.

For example, empirical job rotation models were applied to a series of specific tasks with a goal of finding the best setting with minimum occupational exposure (Carnahan et al., 2000). However, such empirical models are often not accurate in situations other than the ones used for model development, with resulting limitations in generalizing model results to other conditions. Further, most models based on ET have used either power or negative exponential functions. Recently, Frey Law and Avin (2010) fitted joint-specific power and exponential functions based on a meta-analysis of extensive previous results, and from this obtained MFM parameter values for several different joints. Although such models have yielded high correlations between ET and effort level (as %MVC), many of them have not provide good estimates of one or both of two asymptotic tendencies: predicting long ETs for low (~0%MVC) efforts and short ETs for high (~100 %MVC) efforts (El ahrache et al., 2006). In more recent work, power function models by Frey Law and Avin (2010) yielded very long ETs for low efforts, yet exponential models appeared to under-predict ETs for very low %MVCs (i.e., 11 to 29 min. at ~0% MVC). Furthermore, time-dependent changes in physical capacity cannot be predicted using ET models, these models are mainly relevant for static tasks, and such models cannot capture the recovery process. Finally, distinct models may be needed for accurate predictions of ET among different individuals and for different muscles.

Another approach for predicting muscle fatigue—monitoring strength decline during successive work cycles—is also dependent on the specific tasks for which the model is developed (Iridiastadi & Nussbaum, 2006; Roman-Liu et al., 2005; Wood et al., 1997). Such models have been used principally for intermittent static tasks (including different levels of load, duty cycle, and/or cycle time), and which are more relevant to occupational task demands versus single, prolonged static exertions. A majority of these models have been developed based on manipulating one or two of the noted task parameters while keeping the other(s) constant. Due to the number of parameter levels and combinations, comprehensive evaluations are difficult to perform, and interactive effects are not reported in most cases. Iridiastadi and Nussbaum (2006) investigated LMF across a wide range of intermittent static task demands, and subsequently developed an empirical MFM that could predict ET and LMF in the context of shoulder abduction. While this model considered a fairly broad set of muscular efforts, substantial data collection was required and only a single muscle group and physical function was included. Further, the model could only account for a portion of the inherent variability in several outcome measures (i.e., ET, strength, perceived discomfort).

#### 1.5.4.2. Theoretical muscle fatigue models

Theoretical MFMs are models based on fundamental physiological processes, some of which are supported by existing evidence while others are presumed. These models have utilized several approaches for predicting declines in muscle force during fatiguing tasks. Some prior work has developed physiological MFMs, from which it was possible to identify theoretical relationships between the state of muscle metabolic and force decay as a result of muscle activation and fatigue. For example, Giat et al (1993, 1996) developed a MFM based on intramuscular pH levels during the course of stimulation and recovery. Wexler et al. (1997) and colleagues also developed a physiological MFM, based on $Ca^{2+}$ cross-bridge mechanisms to determine muscle force declines.

Subsequently, a series of studies using mathematical models was conducted by this group (Ding et al. 2000a, 2000b, 2005).

Although all of these theoretical models are useful at the single muscle level, they cannot take into account task-related biomechanical factors such as joint angle and velocity. Furthermore, these models are based on physiological mechanisms and tend to be structurally complex; thus, they are likely not useful for ergonomics applications. For example, in order to quantify muscle fatigue in only the quadriceps, Ding et al. (2000b) incorporated 15 parameters (Ding et al., 2000b), while Giat et al. (1993) employed over 30 parameters that had to be fitted for each individual. It is also not practical to implement these models in multiple muscles since they are not computationally efficient. Moreover, the model developed by Ding et al. (2000b) predicts responses to muscle stimulation, which is likely not applicable to occupational tasks, since muscle contraction induced by stimulation is different during voluntary muscle contraction.

Other researchers have developed MFMs at the muscle fiber level, and calculated overall muscle force as the sum of individual fiber forces (Deeb et al., 1992; Hawkins & Hull, 1993). Parameters in the Deeb et al. (1992) model were obtained using curve fitting to experimental data, however this model was not experimentally validated for other conditions. Regarding the Hawkins and Hull model, very low power (<0.18) for experimental validation was provided (Hawkins & Hull, 1993). Although these MFMs are easier to implement, they employ muscle subunits with different recruitment thresholds and muscle properties. Therefore, these approaches provide only rough approximations of temporal changes in muscle force generation, since the motor pool has a continuum of thresholds and firing rates for cells (Winters, 1995).

Another group of theoretical MFMs is based on the general patterns of MU recruitment within a muscle. Liu et al. (2002), established a model based on compartmental theory for understanding fatigue and recovery as a result of muscle stimulation by voluntary drive. Using brain effort as an input variable is one of the advantages for this MFM. However, they assumed a constant and maximal brain effort and utilized this assumption as a basis for the model's rationale. Hence, this model cannot predict fatigue for submaximal or dynamic tasks. Xia and Frey Law (2008) exploited the same MU-based approach in their model, in which they attempted to optimize computational efforts using task-related factors. These investigators characterized MUs as being in, and moving between, three states (i.e., rest, active, and fatigued), and estimated total muscle force according to the number of active MUs. Based on prior literature, they also utilized the joint torque relationship by employing different joint angles and velocities to predict the muscle fatigue at a specific joint level. No experimental validation was provided for this model, and fatigability for different muscle groups was not discussed; nor did the researchers consider interactions between MUs and muscle fibers in their predictions of muscle force generating capacity. Yet another MFM using general principals of MU recruitment is based on the hypothesis that maximum voluntary contraction will decline with each muscle contraction, with higher force generation leading to more rapid fatigue development (Ma et al., 2009). This MFM considers a logarithmic relationship between fatigue level and the external load. It incorporates a simple differential equation, but does not consider limb dynamics (Marion et al., 2010) nor has it been subjected to thorough experimental validation (Ma et al., 2011).

MFMs based on power-endurance relationships have recently been developed that employ the type, recruitment, and fatigability of MUs during exercise (James & Green, 2012; Sih et al., 2012). Sih et al. (2012) modified a prior model by Liu et al. (2002) and utilized it to assess muscle fatigue during submaximal exercises such as cycling and running. They considered four discrete MU states (i.e., all combinations of fatigued and unfatigued, active and inactive MUs) for predicting human power-endurance curves. For validation, they provided only ET predictions based on the fitted model parameters for the experimental data. Further, they assumed similar contractile properties for all MUs. A limitation of this approach is that, since MUs are assumed to only have four possible discrete states, the model is not capable of expressing fatigue as a continuous function of time. Conversely, the recent model developed by James and Green (2012) has been used to predict muscular power output during maximal exertion and ET during submaximal contractions. The contractile properties of muscle are assumed to vary as a continuous function of time and MU type; i.e., the continuum of MUs includes the full range of twitch speed from the slowest to the fastest. This MFM can only account for recruitment of MUs and essentially ignores the influence of MU firing rate on power output. Model parameters were obtained using curve fitting to experimental data; however, this model was not experimentally validated for other conditions.

#### 1.5.5. Existing research gaps

As argued above, it is of value to assess LMF in the context of occupational task performance, with the broader goal of understanding and the effects of diverse task conditions on injury risk and/or performance decrements. However, given the number of diverse variables that likely impact LMF, it is not practical to measure LMF in all possible conditions and for each worker. Therefore, of interest and practical utility is to predict LMF development and/or endurance capability given a specific set of task demands, as opposed to relying on direct measurements (e.g., from an actual worker or by using a mock-up). Several research needs will be addressed in the current research:

- Improving fatigue assessment by measuring the muscle force generating capacity independent of voluntary drive, using electrical stimulation (Gandevia et al., 1995), instead of relying only on MVCs.
- Avoiding (for now) the inherent biomechanical complexity of the human musculoskeletal system (e.g., the shoulder or low back (Gribble et al., 2003)) to help in generating more reliable outcomes. This is addressed in Aim 2, for example by using index finger abduction and the first dorsal interosseous (FDI) muscle (Johnson et al., 1973).
- Preventing simultaneous agonist and antagonist activation by applying the electrical stimulation over the muscle belly.
- Comparing predictions of fatigue and recovery using the two existing MFMs with the most direct occupational relevance (addressed in Chapter 3).
- Evaluating the effects of task parameters such as exertion level and cycle time on muscle fatigue development during intermittent isometric contractions (addressed in Chapter 4).
- Quantifying the “task dependency” or “history dependency” of recovery from LMF (addressed in Chapter 5).

## References


Adamo, D, Khodaee, M, Barringer, S, Johnson, P, & Martin, B. Low mean level sustained and intermittent grip exertions: Influence of age on fatigue and recovery. *Ergonomics* 2009;52(10):1287-1297.

Allen, DG, Lamb, GD, & Westerblad, H. Skeletal Muscle Fatigue: Cellular Mechanisms. *Physiol Rev* 2008;88(1):287-332.

Allman, BL, & Rice, CL. Neuromuscular fatigue and aging: central and peripheral factors. *Muscle & Nerve* 2002;25(6):785-796.

Appell, HJ, Gloser, S, Soares, J, & Duarte, JA. Structural alterations of skeletal muscle induced by ischemia and reperfusion. *BAM-PADOVA* 1999;9(5):263-268.

Aptel, M, Cail, F, Gerling, A, & Louis, O. Proposal of parameters to implement a workstation rotation system to protect against MSDs. *International Journal of Industrial Ergonomics* 2008;38:900-909.

Armstrong, T, Buckle, P, Fine, L, Hagberg, M, Jonsson, B, Kilbom, A, Kuorinka, I, Silverstein, B, Sjogaard, G, & Viikari-Juntura, E. A conceptual model for work-related neck and upper-limb musculoskeletal disorders. *Scand J Work Environ Health* 1993;19:73-84.

Avela, J, Kyröläinen, H, & Komi, PV. Neuromuscular changes after long-lasting mechanically and electrically elicited fatigue. *European Journal of Applied Physiology* 2001;85:317-325.

Avin, KG, & Frey Law, LA. Age-Related Differences in Muscle Fatigue Vary by Contraction Type: A Meta-analysis. *Physical Therapy* 2011;91(8):1153-1165.

Avin, KG, Naughton, MR, Ford, BW, Moore, HE, Monitto-Webber, MN, Stark, AM, Gentile, AJ, & Law, LAF. Sex differences in fatigue resistance are muscle group dependent. *Medicine and science in sports and exercise* 2010;42(10):1943.

Barry, BK, & Enoka, RM. The neurobiology of muscle fatigue: 15 years later. *Integrative and Comparative Biology* 2007;47(4):465-473.

Baudry, S, Klass, M, Pasquet, B, & Duchateau, J. Age-related fatigability of the ankle dorsiflexor muscles during concentric and eccentric contractions. *European Journal of Applied Physiology* 2007;100(5):515-525.

Bemben, MG, Massey, BH, Bemben, DA, Misner, JE, & Boileau, RA. Isometric intermittent endurance of four muscle groups in men aged 20-74 yr. *Medicine and science in sports and exercise* 1996;28(1):145.

Bernard, BP, Putz-Anderson, V, National Institute for Occupational, S, & Health. *Musculoskeletal disorders and workplace factors : a critical review of epidemiologic evidence for work-related musculoskeletal disorders of the neck, upper extremity, and low back*. Cincinnati, OH: U.S. Dept. of Health and Human Services, Public Health Service, Centers for Disease Control and Prevention, National Institute for Occupational Safety and Health ; National Institute for Occupational Safety and Health, 1997.

Bigland-Ritchie, B, EMG/force relations and fatigue of human voluntary contractions. *Exerc Sport Sci Rev* 1981;9:75-117.

Bigland-Ritchie, B, Cafarelli, E, & Vøllestad, NK. Fatigue of submaximal static contractions. *Acta Physiologica Scandinavica* 1986;128 (Suppl 556):137-148.

Bigland-Ritchie, B, Johansson, R, Lippold, OCJ, & Woods, JJ. Contractile speed and EMG changes during fatigue of sustained maximal voluntary contractions. *Journal of neurophysiology* 1983;50(1):313-324.
Bigland-Ritchie, B, Rice, C, Garland, S, & Walsh, M. Task-dependent factors in fatigue of human voluntary contractions. *Advances in experimental medicine and biology* 1995;384:361-380.
Bigland-Ritchie, B, & Woods, JJ. Changes in muscle contractile properties and neural control during human muscular fatigue. *Muscle & Nerve* 1984;7(9):691-699.
Billaut, F, Basset, FA, & Falgairette, G. Muscle coordination changes during intermittent cycling sprints. *Neuroscience Letters* 2005;380(3):265-269.
Bjelle, A, Hagberg, M, & Michaelson, G. Occupational and individual factors in acute shoulder-neck disorders among industrial workers. *British Journal of Industrial Medicine* 1981;38:356-363.
Björklund, M, Crenshaw, AG, Djupsjöbacka, M, & Johansson, H. Position sense acuity is diminished following repetitive low-intensity work to fatigue in a simulated occupational setting. *European Journal of Applied Physiology* 2000;81(5):361-367.
Björkstén, M, & Jonsson, B. Endurance limit of force in long-term intermittent static contractions. *Scand. J Work Environ. & Health* 1977;3:23-27.
Björkstén, MG, Boquist, B, Talbäck, M, & Edling, C. Reported neck and shoulder problems in female industrial workers: the importance of factors at work and at home. *International Journal of Industrial Ergonomics* 2001;27:159-170.
Borg, G. Psychophysical scaling with applications in physical work and the perception of exertion. *Scand J Work Environ Health* 1990;16(Suppl 1):55-58.
Byl, N, Merzenich, M, Cheung, S, Bedenbaugh, P, Nagarajan, S, & Jenkins, W. A primate model for studying focal dystonia and repetitive strain injury: effects on the primary somatosensory cortex. *Physical Therapy* 1997;77(3):269-284.
Byström, S, & Fransson-Hall, C. Acceptability of intermittent handgrip contractions based on physiological response. *Human Factors* 1994;36(1):158-171.
Byström, S, Mathiassen, S, & Fransson-Hall, C. Physiological effects of micropauses in isometric handgrip exercise. *European Journal of Applied Physiology and Occupational Physiology* 1991;63(6):405-411.
Callahan, DM, & Kent-Braun, JA. Effect of old age on human skeletal muscle force-velocity and fatigue properties. *Journal of Applied Physiology* 2011;111(5):1345-1352.
Carnahan, BJ, Redfern, MS, & Norman, B. Designing safe job rotation schedules using optimization and heuristic search. *Ergonomics* 2000;43(4):543-560.
Carpenter, JE, Blasier, RB, & Pellizzon, GG. The effects of muscle fatigue on shoulder joint position sense. *The American Journal of Sports Medicine* 1998;26(2):262-265.
Chaffin, DB. Localized muscle fatigue–definition and measurement. *Journal of Occupational Medicine* 1973;15(4):346-354.
Chaffin, DB, Andersson, G, & Martin, BJ. *Occupational biomechanics*: Wiley New York, 1991.
Christie, A, Snook, EM, & Kent-Braun, JA. Systematic review and meta-analysis of skeletal muscle fatigue in old age. *Med Sci Sports Exerc* 2011;43(4):568-577.

Clark, BC, & Manini, TM. Gender differences in skeletal muscle fatigability are related to contraction type and EMG spectral compression. *Journal of Applied Physiology* 2003;94(6):2263.

Corcos, DM, Jiang, H-Y, Wilding, J, & Gottlieb, GL. (2002). Fatigue induced changes in phasic muscle activation patterns for fast elbow flexion movements *Experimental Brain Research* (Vol. 142, pp. 1-12).

Deeb, JM, Drury, CG, & Pendergast, DR. An exponential model of isometric muscular fatigue as a function of age and muscle groups. *Ergonomics* 1992;35(7/8):899-918.

Dickerson, C, Meszaros, K, Cudlip, A, Chopp-Hurley, J, & Langenderfer, J. The influence of cycle time on shoulder fatigue responses for a fixed total overhead workload. *Journal of Biomechanics* 2015;48(11):2911–2918.

Ding, J, Lee, SCK, Johnston, TE, Wexler, AS, Scott, WB, & Binder-Macleod, SA. Mathematical model that predicts isometric muscle forces for individuals with spinal cord injuries. *Muscle & Nerve* 2005;31(6):702-712.

Ding, J, Wexler, AS, & Binder-Macleod, SA. Development of a mathematical model that predicts optimal muscle activation patterns by using brief trains. *J Appl Physiol* 2000a;88(3):917-925.

Ding, J, Wexler, AS, & Binder-Macleod, SA. A predictive model of fatigue in human skeletal muscles. *J Appl Physiol* 2000b;89(4):1322-1332.

Ditor, DS, & Hicks, AL. The effect of age and gender on the relative fatigability of the human adductor pollicis muscle. *Canadian Journal of Physiology and Pharmacology* 2000;78:781-790.

Eberstein, A, & Beattie, B. Simultaneous measurement of muscle conduction velocity and EMG power spectrum changes during fatigue. *Muscle & Nerve* 1985;8:768-773.

Edwards, R, Hill, D, Jones, D, & Merton, P. Fatigue of long duration in human skeletal muscle after exercise. *J Physiology* 1977;272:769-778.

Edwards, RHT. Human muscle function and fatigue. *Human muscle fatigue: physiological mechanisms* 1981:1-18.

El ahrache, K, & Imbeau, D. Comparison of rest allowance models for static muscular work. *International Journal of Industrial Ergonomics* 2009;39(1):73-80.

El ahrache, K, Imbeau, D, & Farbos, B. Percentile values for determining maximum endurance times for static muscular work. *International Journal of Industrial Ergonomics* 2006;36(2):99-108.

Engström, T, Hanse, J, & Kadefors, R. Musculoskeletal symptoms due to technical preconditions in long cycle time work in an automobile assembly plant: a study of prevalence and relation to psychosocial factors and physical exposure. *Applied Ergonomics* 1999;30(5):443-453.

Enoka, R, & Stuart, D. Neurobiology of muscle fatigue. *J Appl Physiol* 1992;72(5):1631-1648.

Enoka, R, Mechanisms of muscle fatigue: Central factors and task dependency. *Journal of Electromyography and Kinesiology* 1995;5(3):141-149.

Fitts, RH, Cellular mechanisms of muscle fatigue. *Physiol Rev* 1994;74(1):49-94.

Forde, M, Punnett, L, & Wegman, D. Pathomechanisms of work-related musculoskeletal disorders: conceptual issues. *Ergonomics* 2002;45(9):619-630.

Frey Law, L, & Avin, K. Endurance time is joint-specific: A modelling and meta-analysis investigation. *Ergonomics* 2010;53(1):109-129.

Frontera, WR, Hughes, VA, Lutz, KJ, & Evans, WJ. A cross-sectional study of muscle strength and mass in 45- to 78-yr-old men and women. *Journal of Applied Physiology* 1991;71(2):644-650.

Gandevia, SC. Spinal and Supraspinal Factors in Human Muscle Fatigue. *Physiol Rev* 2001;81(4):1725-1789.

Gandevia, SC, Allen, GM, & McKenzie, DK. *Central fatigue: critical issues, quantification and practical implications* (Vol. 384: Fatigue). New York: Plenum Press, 1995.

Gates, DH, & Dingwell, JB. The effects of neuromuscular fatigue on task performance during repetitive goal-directed movements. *Exp Brain Res* 2008;187(4):573-585.

Giat, Y, Mizrahi, J, & Levy, M. A model of fatigue and recovery in paraplegic's quadriceps muscle subjected to intermittent FES. *Journal of Biomechanics* 1996;118(3):357.

Giat, Y, Mizrahi, J, & Levy, M. A musculotendon model of the fatigue profiles of paralyzed quadriceps muscle under FES. *IEEE Transactions on Biomedical Engineering* 1993;40(7):664-674.

Gorelick, M, Brown, JMM, & Groeller, H. Short-duration fatigue alters neuromuscular coordination of trunk musculature: implications for injury. *Applied Ergonomics* 2003;34(4):317-325.

Gribble, PL, Mullin, LI, Cothros, N, & Mattar, A. Role of cocontraction in arm movement accuracy. *Journal of neurophysiology* 2003;89(5):2396.

Hagberg, M. Occupational musculoskeletal stress and disorders of the neck and shoulder: a review of possible pathophysiology. *International Archives of Occupational and Environmental Health* 1984;53(3):269-278.

Häkkinen, K, & Komi, PV. Effects of fatigue and recovery on electromyographic and isometric force- and relaxation-time characteristics of human skeletal muscle. *European Journal of Applied Physiology* 1986;55(6):588-596.

Harrigan, MW. Employment projections to 2012: Concepts and context. *Monthly Lab. Rev.* 2004;127:3.

Hawkins, D, & Hull, ML. Muscle force as affected by fatigue: Mathematical model and experimental verification. *Journal of Biomechanics* 1993;26(9):1117-1128.

Henneman, E, Somjen, G, & Carpenter, D. Functional significance of cell size in spinal motoneurons. *Journal of neurophysiology* 1965;28:560-580.

Herberts, P, Kadefors, R, Högfors, C, & Sigholm, G. Shoulder pain and heavy manual labor. *Clinical Orthopaedics and Related Research* 1984;191:166-178.

Hicks, AL, Kent-Braun, J, & Ditor, DS. Sex differences in human skeletal muscle fatigue. *Exercise and Sport Sciences Reviews* 2001;29(3):109-112.

Horton, L, Nussbaum, M, & Agnew, M. Effects of rotation frequency and task order on localised muscle fatigue and performance during repetitive static shoulder exertions. *Ergonomics* 2012;55(10):1205-1217.

Hunter, SK. Sex differences and mechanisms of task-specific muscle fatigue. *Exerc Sport Sci Rev* 2009;37(3):113-122.

Hunter, SK, & Enoka, RM. Changes in muscle activation can prolong the endurance time of a submaximal isometric contraction in humans. *Journal of Applied Physiology* 2003;94(1):108-118.
Iridiastadi, H, & Nussbaum, M. Muscular fatigue and endurance during intermittent static efforts: effects of contraction level, duty cycle, and cycle time. *Hum Factors* 2006;48(4):710-720.
Iridiastadi, H, & Nussbaum, MA. Muscle fatigue and endurance during repetitive intermittent static efforts: development of prediction models. *Ergonomics* 2006;49(4):344-360.
James, A, & Green, S. A phenomenological model of muscle fatigue and the power-endurance relationship. *Journal of Applied Physiology* 2012;113(10):1643-1651.
Jaric, S, Blesic, S, Milanovic, S, Radovanovic, S, Ljubisavljevic, M, & Anastasijevic, R. Changes in movement final position associated with agonist and antagonist muscle fatigue. *European Journal of Applied Physiology and Occupational Physiology* 1999;80(5):467-471.
Johnson, MA, Polgar, J, Weightman, D, & Appleton, D. Data on the distribution of fibre types in thirty-six human muscles: An autopsy study. *Journal of the Neurological Sciences* 1973;18(1):111-129.
Johnson, T. Age-related differences in isometric and dynamic strength and endurance. *Physical Therapy* 1982;62(7):985-989.
Jørgensen, K, Fallentin, N, Krogh-Lund, C, & Jensen, B. Electromyography and fatigue during prolonged, low-level static contractions. *European Journal of Applied Physiology* 1988;57:316-321.
Kahn, JF, & Monod, H. Fatigue induced by static work. *Ergonomics* 1989;32(7):839-846.
Kirkendall, DT. Mechanisms of peripheral fatigue. *Medicine & Science in Sports & Exercise* 1990;22(4):444.
Kuijer, PPFM, van der Beek, AJ, van Dieën, JH, Visser, B, & Frings-Dresen, MH. Effect of job rotation on need for recovery, musculoskeletal complaints, and sick leave due to musculoskeletal complaints: a prospective study among refuse collectors. *American Journal of Industrial Medicine* 2005;47:394-402.
Kumar, S. Theories of musculoskeletal injury causation. *Ergonomics* 2001;44(1):17-47.
Lexell, J, Henriksson-Larsén, K, Winblad, B, & Sjöström, M. Distribution of different fiber types in human skeletal msucles: effects of aging studied in whole muscle cross sections. *Muscle & Nerve* 1983;6(8):588-595.
Lindström, B, Lexell, J, Gerdle, B, & Downham, D. Skeletal muscle fatigue and endurance in young and old men and women. *The Journals of Gerontology, Series A* 1997;52(1):59-67.
Liu, JZ, Brown, RW, & Yue, GH. A Dynamical Model of Muscle Activation, Fatigue, and Recovery. *Biophysical Journal* 2002;82(5):2344-2359.
Löscher, W, Cresswell, AG, & Thorstensson, A. Central fatigue during a long-lasting submaximal contraction of the triceps surae. *Experimental Brain Research* 1996;108(2):305-314.
Lovlin, R, Cottle, W, Pyke, I, Kavanagh, M, & Belcastro, A. Are indices of free radical damage related to exercise intensity. *European Journal of Applied Physiology and Occupational Physiology* 1987;56(3):313-316.

Lucidi, CA, & Lehman, SL. Adaptation to fatigue of long duration in human wrist movements. *J. Appl. Physiol.* 1992;73(6):2596-2603.
Ma, L, Chablat, D, Bennis, F, & Zhang, W. A new simple dynamic muscle fatigue model and its validation. *International Journal of Industrial Ergonomics* 2009;39(1):211-220.
Ma, L, Chablat, D, Bennis, F, Zhang, W, Hu, B, & Guillaume, F. Fatigue evaluation in maintenance and assembly operations by digital human simulation in virtual environment. *Virtual Reality* 2011;15(1):55-68.
Maeda, K. Occupational cervicobrachial disorder and its causative factors. *J Human Ergol* 1977;6:193-202.
Malchaire, J, Cock, N, & Vergracht, S. Review of the factors associated with musculoskeletal problems in epidemiological studies. *International Archives of Occupational and Environmental Health* 2001;74(2):79-90.
Marion, MS, Wexler, AS, & Hull, ML. Predicting fatigue during electrically stimulated non-isometric contractions. *Muscle & Nerve* 2010;41(6):857-867.
Mathiassen, S. The influence of exercise/rest schedule on the physiological and psychophysical response to isometric shoulder-neck exercise. *European Journal of Applied Physiology* 1993;67:528-539.
Mathiassen, S, & Winkel, J. Can occupational guidelines for work-rest schedules be based on endurance time data? *Ergonomics* 1992;35(3):253-259.
Miller, AEJ, MacDougall, JD, Tarnopolsky, MA, & Sale, DG. Gender differences in strength and muscle fiber characteristics. *European Journal of Applied Physiology* 1993;66(3):254-262.
Monod, H, & Scherrer, J. The work capacity of a synergic muscular group. *Ergonomics* 1965;8(3):329-338.
Moore, A, & Wells, R. Effect of cycle time and duty cycle on psychophysically determined acceptable levels in a highly repetitive task. *Ergonomics* 2005;48(7):859-873.
NIOSH. (2001). National Occupational Research Agenda for Musculoskeletal Disorders.
Nussbaum, M, Clark, L, Lanza, M, & Rice, K. Fatigue and endurance limits during intermittent overhead work. *AIHAJ-American Industrial Hygiene Association* 2001;62(4):446-456.
Pedersen, J, Lönn, J, Hellström, F, Djupsjöbacka, M, & Johansson, H. Localized muscle fatigue decreases the acuity of the movement sense in the human shoulder. *Medicine and science in sports and exercise* 1999;31(7):1047-1052.
Peters, EJD, & Fuglevand, AJ. Cessation of human motor unit discharge during sustained maximal voluntary contraction. *Neuroscience Letters* 1999;274(1):66-70.
Petrella, JK, Kim, J-s, Tuggle, SC, Hall, SR, & Bamman, MM. Age differences in knee extension power, contractile velocity, and fatigability. *Journal of Applied Physiology* 2005;98:211-220.
Petrofsky, J, Stacy, R, & Laymon, M. The relationship between exercise work intervals and duration of exercise on lower extremity training induced by electrical stimulation in humans with spinal cord injuries. *European Journal of Applied Physiology* 2000;82(5-6):504-509.

Place, N, Yamada, T, Bruton, J, & Westerblad, H. Muscle fatigue: from observations in humans to underlying mechanisms studied in intact single muscle fibres. *European Journal of Applied Physiology* 2010;110(1):1-15.

Potvin, JR, & O'Brien, PR. Trunk Muscle Co-contraction Increases During Fatiguing, Isometric, Lateral Bend Exertions: Possible Implications for Spine Stability. *Spine* 1998;23(7):774-780.

Raina, SM, & Dickerson, CR. The influence of job rotation and task order on muscle fatigue: A deltoid example. *Work* 2009;34(2):205-213.

Rohmert, W. Ermittlung von erholungspausen für statische arbeit des menschen. *Internationale Zeitschrift für Angewandte Physiologie enschliesslich Arbeitsphysiologie* 1960;18:123-164.

Roman-Liu, D, Tokarski, T, & Kowalewski, R. Decrease of force capabilities as an index of upper limb fatigue. *Ergonomics* 2005;48(8):930-948.

Sahlin, K, & Ren, J. Relationship of contraction capacity to metabolic changes during recovery from a fatiguing contraction. *J Appl Physiol* 1989;67(2):648-654.

Sejersted, O, & Vøllestad, N. Physiology of muscle fatigue and associated pain. In H. Værøy & H. Merskey (Eds.), *Progress in Fibromyalgia and Myofascial Pain* (pp. 41-51). Amsterdam: Elsevier Science Publishers B.V., 1993.

Selen, LPJ, Beek, PJ, & Dieën, JH. Fatigue-induced changes of impedance and performance in target tracking. *Experimental Brain Research* 2007;181(1):99-108.

Sih, B, Ng, L, & Stuhmiller, J. Generalization of a Model Based on Biophysical Concepts of Muscle Activation, Fatigue and Recovery that Explains Exercise Performance. *International journal of sports medicine* 2012;33(4):258.

Silverstein, B, Fine, L, & Armstrong, T. Hand wrist cumulative trauma disorders in industry. *British Journal of Industrial Medicine* 1986;43(11):779-784.

Sjøgaard, G, & Jensen, B. Muscle pathology with overuse. *Chronic Musculoskeletal Injuries in the Workplace. In: Ranney D, editor. WB Saunders Company: Philadelphia, PA* 1997:17-40.

Skinner, HB, Wyatt, MP, Hodgdon, JA, Conard, DW, & Barrack, RL. Effect of fatigue on joint position sense of the knee. *Journal of Orthopaedic Research* 1986;4:112-118.

Søgaard, K, Gandevia, SC, Todd, G, Petersen, NT, & Taylor, JL. The effect of sustained low-intensity contractions on supraspinal fatigue in human elbow flexor muscles. *The Journal of Physiology* 2006;573(2):511-523.

Sommerich, C, McGlothlin, J, & Marras, W. Occupational risk factors associated with soft tissue disorders of the shoulder: a review of recent investigations in the literature. *Ergonomics* 1993;36(6):697-717.

Sparto, PJ, Parnianpour, M, Reinsel, TE, & Simon, S. The effect of fatigue on multijoint kinematics, coordination, and postural stability during a repetitive lifting test. *J Orthop Sports Phys Ther* 1997;25(1):3-12.

Stackhouse, SK, Stevens, JE, Lee, SCK, Pearce, KM, Snyder-Mackler, L, & Binder-Macleod, SA. Maximum voluntary activation in nonfatigued and fatigued muscle of young and elderly individuals. *Physical Therapy* 2001;81:1102-1109.

Stauber, W. Factors involved in strain-induced injury in skeletal muscles and outcomes of prolonged exposures. *Journal of Electromyography and Kinesiology* 2004;14(1):61-70.

Strang, A, & Berg, W. Fatigue-induced adaptive changes of anticipatory postural adjustments. *Experimental Brain Research* 2007;178(1):49-61.

Taylor, JL, & Gandevia, SC. A comparison of central aspects of fatigue in submaximal and maximal voluntary contractions. *J Appl Physiol* 2008;104(2):542-550.

Taylor, JL, Todd, G, & Gandevia, SC. Evidence for a supraspinal contribution to human muscle fatigue. *Clinical and Experimental Pharmacology and Physiology* 2006;33(4):400-405.

van Duinen, H, Renken, R, Maurits, N, & Zijdewind, I. Effects of motor fatigue on human brain activity, an fMRI study. *NeuroImage* 2007;35(4):1438-1449.

Veiersted, K. Sustained muscle tension as a risk factor for trapezius myalgia. *International Journal of Industrial Ergonomics* 1994;14:333-339.

Veiersted, KB, Westgaard, RH, & Andersen, P. Electromyographic evaluation of muscular work patterns as a predictor of trapezius myalgia. *Scand J Work Environ Health* 1993;19:284-290.

Vøllestad, N. Measurement of human muscle fatigue. *Journal of Neuroscience Methods* 1997;74(2):219-227.

Vøllestad, NK, & Sejersted, OM. Biochemical correlates of fatigue: a brief review. *European Journal of Applied Physiology* 1988;57:336-347.

Vøllestad, NK, Sejersted, OM, Bahr, R, Woods, JJ, & Bigland-Ritchie, B. Motor drive and metabolic responses during repeated submaximal contractions in humans. *Journal of Applied Physiology* 1988;64(4):1421-1427.

Westerblad, H, Allen, D, Bruton, J, Andrade, F, & Lännergren, J. Mechanisms underlying the reduction of isometric force in skeletal muscle fatigue. *Acta Physiologica Scandinavica* 1998;162(3):253-260.

Westgaard, R. Measurement and evaluation of postural load in occupational work situations. *European Journal of Applied Physiology* 1988;57:291-304.

Wexler, AS, Jun, D, & Binder-Macleod, SA. A mathematical model that predicts skeletal muscle force. *Biomedical Engineering, IEEE Transactions on* 1997;44(5):337-348.

Wilder, DG, Aleksiev, AR, Magnusson, ML, Pope, MH, Spratt, KF, & Goel, VK. Muscular Response to Sudden Load: A Tool to Evaluate Fatigue and Rehabilitation. *Spine* 1996;21(22):2628-2639.

Winters, JM. How detailed should muscle models be to understand multi-joint movement coordination? *Human Movement Science* 1995;14(4-5):401-442.

Wood, DD, Fisher, DL, & Andres, RO. Minimizing fatigue during repetitive jobs: optimal work-rest schedules. *Human Factors* 1997;39(1):83-101.

Xia, T, & Frey Law, LA. A theoretical approach for modeling peripheral muscle fatigue and recovery. *Journal of Biomechanics* 2008;41(14):3046-3052.

Yassierli, & Nussbaum, M. Effects of age, gender, and task parameters on fatigue development during intermittent isokinetic torso extensions. *International Journal of Industrial Ergonomics* 2009;39(1):185-191.

Yassierli, & Nussbaum, M. Muscle fatigue during intermittent isokinetic shoulder abduction: age effects and utility of electromyographic measures. *Ergonomics* 2007;50(7):1110-1126.

Yassierli, Nussbaum, MA, Iridiastadi, H, & Wojcik, LA. The influence of age on isometric endurance and fatigue is muscle dependent: a study of shoulder abduction and torso extension. *Ergonomics* 2007;50(1):26-45.

# 2. A Review of Occupationally-Relevant Models of Localized Muscle Fatigue

Ehsan Rashedi and Maury A. Nussbaum

**Abstract**

Localized muscle fatigue (LMF) is a complex phenomenon that can differ between individuals, tasks, and muscles. Several muscle-fatigue models (MFMs) have been developed in prior research. MFMs have potential practical value in ergonomics, given that LMF can impair performance, serve as a surrogate measure of injury risk, and may act as a causal factor for work-related musculoskeletal disorders. Existing MFMs are reviewed here, and which are broadly classified as either "empirical" or "theoretical". Two specific MFMs, considered most ergonomically-relevant, were directly compared and some important differences in predictions were found. Identifying such differences is suggested as a useful approach, both for developing testable hypotheses and in guiding subsequent model development or refinement. Other potential approaches for improving future MFMs are also discussed, including expansion of model structure to account for individual differences (e.g., age, gender, and obesity), task related parameters, and variability in motor unit composition.

**Keywords:** localized muscle fatigue; models; performance; ergonomics; recovery; endurance; review

* This chapter was published in *International Journal of Human Factors Modelling and Simulation* 2015;5(1):61-80.

## 2.1. Introduction

The primary goals of this paper are to give an overview of the advantages and limitations of existing muscle fatigue models (MFMs), provide a quantitative comparison between predictions generated by two specific MFMs, and suggest potential directions for future work. In the following sections, we first define localized muscle fatigue (LMF), summarize LMF impacts on performance and work-related musculoskeletal disorders (WMSDs), and highlight current evidence regarding underlying LMF-related mechanisms. Existing "empirical" and "theoretical" MFMs are then described, and predictions from two" ergonomically-relevant" MFMs are compared under different loading conditions. Finally, individual and task-related factors affecting LMF are discussed, and potential future structural changes for MFMs are suggested.

In the context of muscular activity/exertion, LMF has been described/defined as "a loss of maximal force generating capacity" (Gandevia et al., 1995), or a "failure to maintain the required or expected force" (Edwards, 1981). These definitions are associated with the notion of endurance time (ET) and an inability to sustain a given task (aka "task failure"). However, diverse aspects of the neuromuscular system are altered by muscular activity, and such alterations occur prior to ET or task failure. Therefore, LMF has been defined more specifically as "any exercise-induced reduction in the ability of a muscle to generate force or power, regardless of the ability to sustain the task" (Bigland-Ritchie & Woods, 1984). LMF is a complex, multifactorial phenomenon that is widely used as an indicator of underlying physiological processes, since such fatigue leads to a decline in desired performance and muscle force capacity during diverse activities involving voluntary muscle force generation (Vøllestad, 1997) and more specifically a range of activities in the occupational setting (Chaffin, 1973). LMF leads to subjective and objective changes, such as an increased perceived exertion and discomfort, reduced strength, diminished neuromuscular control, muscle tremors, and altered electromyographic (EMG) signals.

LMF has often been described based on the limiting condition, in particular the duration that a task or a set of demands can be continued (i.e., ET). Similarly, exhaustion, and the resultant task failure, is relatively well characterized for prolonged static exertions. However, the ability to predict LMF under a range of task characteristics (e.g., as a function of intensity, duration, or speed) is yet to be developed. Facilitating such predictions is needed, given the potential for LMF to adversely affect performance and to increase the risk of WMSDs. Regarding the former, a worker's physical performance and capacity can directly impact task efficiency and effectiveness. Fatigue-induced declines in performance can result from several sources, including impaired coordination (Sparto et al., 1997), longer reaction times (Häkkinen & Komi, 1986), increased muscle tremors (Hunter & Enoka, 2003), and decrements in proprioceptive ability (Björklund et al., 2000) and sense of effort (Enoka & Stuart, 1992).

WMSDs are prevalent in diverse occupational environments, accounting for about 33% of all nonfatal injuries or illnesses in the U.S. that necessitate missed days from work (Bureau of Labor Statistics, 2012). LMF has received increasing attention among ergonomists, and has been recognized as an important measure in research and interventions aimed at reducing musculoskeletal disorders (MSDs) in the workplace. The specific role of LMF in MSD development is not yet clear, though, and the existence of LMF does not necessarily imply an increased risk of WMSDs (Mathiassen & Winkel, 1992). However, factors such as working posture and repetitive/sustained muscle contraction that are closely associated with muscle fatigue likely contribute to soft tissue injuries (Sommerich et al., 1993). Several theories and conceptual models have also been proposed (e.g., Forde et al., 2002), describing potential links between LMF and WMSD-related mechanisms such as a loss of calcium homeostasis (Sjøgaard & Jensen, 1997) and ischemia-reperfusion (Appell et al., 1999). Support for these theories comes from several causal mechanisms associated with LMF as a potential contributor to WMSDs (Kumar, 2001). Such evidence supports the role of LMF as at least a surrogate measure of injury risk, and possibly acting as a causal factor for WMSDs.

### 2.2. Mechanisms of LMF

Mammalian muscle is comprised of muscle fibers that generate force production and movements resulting from contractions driven by nervous system commands. These commands are ultimately transmitted by motor neurons, which represent the major efferent neurons that innervate muscle fibers. Muscle fibers and the motor neuron that innervates them comprise the basic functional unit known as a motor unit (MU). Several sequential processes in the central nervous system (CNS) occur to activate motor neurones (Enoka, 1995) and to generate action potentials and finally force generation in muscle fibers (Gandevia et al., 1995). Each of these processes can act as a limiting factor and contributor to fatigue (Vøllestad, 1997). These processes are categorized as central and peripheral mechanisms, both of which can be compromised by maximal or submaximal contractions (Fitts, 1994). Central fatigue, defined as an exercise-induced degradation of voluntary muscle activation (Gandevia et al., 1995), involves all central factors (i.e., mechanisms located in the CNS) that activate motor neurones, including motivational aspects and integration of sensory information. Peripheral fatigue involves mechanisms such as neuromuscular transmission, muscle action potential propagation, excitation-contraction coupling, and the contractile elements in muscles (Gandevia, 2001). While central fatigue is considered a contributing factor to LMF, most reports describe peripheral fatigue as being more substantial in LMF development and requiring a longer duration for recovery.

## 2.3. Overview of Muscle Fatigue Modeling

Despite considerable progress in understanding and predicting muscle fatigue, there are many uncertainties and unresolved issues (Enoka, 1995) that are principally associated with the physiological complexity of LMF and the diverse mechanisms that underlie the initiation and development of LMF. The specific impact(s) of these various mechanisms on muscle performance impairment is not well understood, and these impacts can vary substantially depending on task characteristics (Bigland-Ritchie et al., 1995). In other words, there is not a unique mechanism responsible for the decline in force generating capacity in all types of exertions (Fitts, 1994). Any predictions of human muscle fatigue, such as using a model as described subsequently, are thus likely to have inevitable limitations; ideally, though, such a model should account for a range of specific task parameters or conditions. Task duration and frequency, load magnitude, and personal factors are likely of particular relevance for ergonomic applications, and the influences of these are briefly summarized below.

### 2.3.1 Task parameters affecting fatigue/endurance

A number of studies have identified important task types/parameters affecting the development of LMF. Among these factors are prolonged static vs. dynamic and sustained vs. intermittent tasks, and more specific task parameters such as intensity, duration, cycle time, and work-rest proportion or duty cycle (Horton et al., 2012; Iridiastadi & Nussbaum, 2006; Yassierli et al., 2007). Higher effort levels and duty cycles have been found to have consistent effects involving higher rates of LMF development and/or decreased ETs, for example during intermittent shoulder abductions (Björkstén & Jonsson, 1977; Iridiastadi & Nussbaum, 2006). Underlying these effects, greater effort levels/durations result in higher/longer demands on each recruited muscle fiber and more recruitment of MUs, both leading to a faster rate of muscular fatigue.

Quantitatively, a curvilinear relationship has been long recognized between muscle contraction level and ET (Rohmert, 1960). Tasks requiring low levels of effort can be sustained for a long duration, whereas ET decreases exponentially (or as a power function) with higher levels of muscle contraction. Further, a lower duty cycle (i.e., ratio of contraction period to cycle time) results in a longer ET (Björkstén & Jonsson, 1977). Cycle time may also impact the development of LMF, but apparently with less substantial effects compared with effort level (Iridiastadi & Nussbaum, 2006; Yassierli & Nussbaum, 2009). Inconsistent evidence actually exists regarding the benefits of a shorter cycle times if other task aspects are kept consistent. While prolonged static exertions can be harmful, high repetitions with very short cycle time can also be unfavorable (Sommerich et al., 1993). In contrast, one study indicated that a shorter cycle time yields slower development of LMF (Yassierli & Nussbaum, 2009).

### 2.3.2. Muscle fatigue modeling approaches

LMF is influenced by many potential factors, with differing impacts depending on loading conditions. Although LMF quantification is useful for many reasons, such as work-rest scheduling, task assessment, and determining an individuals' physical capacity, it is not practical to measure LMF directly under all possible situations. Even for a given situation or task, directly measuring LMF can be time consuming and costly. As such, the use of MFMs to predict muscle fatigue has broad potential application. These models can be categorized into two types, *empirical* and *theoretical*, and existing work on each is summarized below.

#### 2.3.2.1. Empirical muscle fatigue models

Empirical MFMs are based on empirical observations and fitting experimental data, rather than mathematical relationships between underlying system parameters. The origin of this type of MFMs is often credited to Rohmert (1960) and Monod and Scherrer (1965), with substantial subsequent expansion or modification. These models are based on either ET (El ahrache et al., 2006) or on decreases in strength during successive work cycles, with some approaches considering cycle time, submaximal force level, and duty cycle as modifying task parameters (Roman-Liu et al., 2005; Wood et al., 1997). A principal advantage of these models is the ability to modify them for a specific situation. For example, empirical job rotation models were applied to a series of specific tasks with a goal of finding the best setting with minimum occupational exposure (Carnahan et al., 2000). However, such empirical models are often not accurate in situations other than the ones used for model development, with resulting limitations in generalizing model results to other conditions. Further, most models based on ET have used either power or negative exponential functions. Recently, Frey Law and Avin (2010) fitted joint-specific power and exponential functions based on a meta-analysis of extensive previous results, and from this obtained MFM parameter values for several different joints. Although such models have yielded high correlations between ET and effort level (as %MVC), many of them have not provide good estimates of one or both of two asymptotic tendencies: predicting long ETs for low (~0%MVC) efforts and short ETs for high (~100 %MVC) efforts (El ahrache et al., 2006). In more recent work, power function models by Frey Law and Avin (2010) yielded very long ETs for low efforts, yet exponential models appeared to under-predict ETs for very low %MVCs (i.e., 11 to 29 min. at ~0% MVC). Furthermore, time-dependent changes in physical capacity cannot be predicted using ET models, these models are mainly relevant for static tasks, and such models cannot capture the recovery process. Finally, distinct models may be needed for accurate predictions of ET among different individuals and for different muscles.

Another approach for predicting muscle fatigue—monitoring strength decline during successive work cycles—is also dependent on the specific tasks for which the model is developed (Iridiastadi & Nussbaum, 2006; Roman-Liu et al., 2005; Wood et al., 1997). Such models have been used principally for intermittent static tasks (including different levels of load, duty cycle, and/or cycle time), and which are more relevant to occupational task demands versus single, prolonged static exertions. A majority of these models have been developed based on manipulating one or two of the noted task parameters while keeping the other(s) constant. Due to the number of parameter levels and combinations, comprehensive evaluations are difficult to perform, and interactive effects are not reported in most cases. Iridiastadi and Nussbaum (2006) investigated LMF across a wide range of intermittent static task demands, and subsequently developed an empirical MFM that could predict ET and LMF in the context of shoulder abduction. While this model considered a fairly broad set of muscular efforts, substantial data collection was required and only a single muscle group and physical function was included. Further, the model could only account for a portion of the inherent variability in several outcome measures (i.e., ET, strength, perceived discomfort).

#### 2.3.2.2. Theoretical muscle fatigue models

In part to address the noted limitations of empirical MFMs, more recent efforts have developed theoretical MFMs that are based on mathematical representations of physiological processes that are either presumed or supported by existing evidence. These models have utilized several approaches for predicting declines in muscle force during fatiguing tasks (summarized in Table 2.1). Some prior work has developed physiological MFMs, from which it was possible to identify theoretical relationships between the state of muscle metabolic and force capacity as a result of muscle activation and fatigue. For example, Giat et al. (1996) developed a MFM based on intramuscular pH levels during the course of stimulation and recovery. Wexler et al. (1997) developed a physiological muscle force model based on $Ca^{2+}$ cross-bridge mechanisms. Subsequently, this group also used this model to predict muscle fatigue by completing a series of studies using mathematical models of muscular responses to a range of stimulation scenarios (Ding et al. 2000, 2005). More recently, this model has been expanded for non-isometric contractions (Marion et al., 2010), in which muscle responses to fatiguing electrical stimulations were represented as a reduction of muscle power.

Although these theoretical models are useful at the single muscle level, they do not account for important task-related biomechanical factors such as joint/muscle angle and velocity. Furthermore, these models are based on relatively "low-level" physiological mechanisms and are structurally complex; thus, they are likely not practical for ergonomics applications. For example, while only a few model parameters are directly related to muscle fatigue, 15 parameters still need to be specified to use the MFM by Ding et al. (2000) in only the quadriceps, while Giat et al. (1996) employed over 30 parameters that had to be fitted for each individual. It also may not be practical to implement these models for multiple muscles, since they are not computationally efficient. Moreover, the Ding et al. (2000) model predicts responses to muscle stimulation, which is likely not applicable to occupational tasks, since muscle contraction induced by stimulation is different during voluntary activation. Other researchers have developed MFMs at the muscle fiber level, and have calculated overall muscle force as the sum of individual fiber forces (Deeb et al., 1992; Hawkins & Hull, 1993). Parameters in the Deeb et al. (1992) model were obtained using curve fitting to experimental data, however this model was not experimentally validated for other conditions. Regarding the

Table 2.1: Summary of theoretical muscle fatigue models (MFM). As noted in the text, MFMs are considered to have low relevance to ergonomics if they are highly complex, computationally demanding, or do not predict responses to voluntary contractions.

| **MFM** | **Model output** | **Relevance to ergonomics** | **Model Evaluation Condition** |
|---|---|---|---|
| Deeb et al. (1992) | Force | High | Isometric elbow flexion and knee extension (MVC and SVCs at 20, 40, 60, and 80%) |
| Hawkins and Hull (1993) | Force | Low | Isometric and cyclic elbow extension (MVC) |
| Giat et al. (1996) | Force | Low | Isometric quadriceps contractions (ES) |
| Ding et al. (2000) | Force | Low | Isometric quadriceps femoris contractions (ES) |
| Liu et al. (2002) | Force | Low | Sustained handgrip tasks (MVC) |
| J. Ding et al. (2005) | Force | Low | Isometric quadriceps femoris contractions (ES) |
| Xia and Frey Law (2008) | Force | High | Optimal model parameter values obtained by, and evaluated using, pre-defined exponential and power curves of ET- |
| Ma et al. (2009) | Force | High | Compared to existing static models of ET vs. exertion level (ME). Sustained isometric push tasks with different time intervals using a constant load = 25 N (SVC) |
| Ma et al. (2010) | Torque | High | Compared to existing static models of ET-exertion level (ME). Continuous and intermittent repetitive lifting task using a constant weight = 36 N (SVC) |
| Marion et al. (2010) | Force | Low | Isometric and non-isometric knee extensions (ES) |
| James and Green (2012) | Power | High | Optimal model parameter values obtained by curve fitting to sprint PT and cycling PE profiles (ME) |
| Sih et al. (2012) | Force, power | High | Optimal model parameter values obtained by curve fitting to hand grip task (MVC), isometric quadriceps contractions (ES), and cycling and running PE profiles (ME) |

SVC: submaximal voluntary contraction, ES: electrical stimulation, ME: mathematical evaluation; PE: power-endurance; PT: power-time

Hawkins and Hull (1993) model, relatively low statistical power (<0.18) was reported, suggesting a need for further experimental validation.

Another group of theoretical MFMs is based on the general patterns of MU recruitment within a muscle. Liu et al. (2002) developed a model based on compartmental theory to understand fatigue and recovery as a result of voluntary muscle activation. One feature of this MFM is the use of brain effort as an input variable. However, a constant and maximal brain effort was assumed, and thus this model cannot predict fatigue for submaximal efforts or those with time varying demands. Xia and Frey Law (2008) exploited the same MU-based approach in their model, in which they sought to optimize computational effort using a simplified approach in comparison to other models that utilize explicit muscle length-tension and force-velocity relationships. They characterized MUs as being in, and moving between, three states (i.e., rest, active, and fatigued), and estimated total muscle force from the number of active MUs. Limited experimental validation was provided for this model, since optimal model parameters were only obtained for sustained isometric tasks (Frey-Law et al., 2012), based on the pre-determined exponential and power curves describing the ET-exertion level relationship (Frey Law & Avin, 2010). The identified model parameters were evaluated using a set of exertion levels within the same range as those used for parameter identification, though an independent experimental validation was not completed.

Yet another MFM, using general principals of MU recruitment, is based on the hypothesis that maximum voluntary contraction will decline with each muscle contraction, with higher force generation leading to more rapid fatigue development (Ma et al., 2009). This MFM incorporates two independent, first-order differential equations for fatigue development and recovery (Ma et al., 2010), and is based conceptually on the MU recruitment concept described by Liu et al. (2002). External force and individual differences, such as MVC and fatigue resistance, are implicitly incorporated in this MFM. Recently, Brouillette et al. (2012) evaluated this MFM using other published ET models for the shoulder and elbow joints. Although this model has a simple structure that facilitate its usage in other applications such as digital human modeling (Brouillette et al., 2012), it does not consider limb dynamics, nor has it been subjected to thorough experimental validation.

MFMs based on power-endurance relationships have recently been developed, and that employ the type, recruitment, and fatigability of MUs during exercise. Sih et al. (2012) modified the prior model of Liu et al. (2002), and assessed muscle fatigue during submaximal exercises such as cycling and running. They considered four discrete MU states (i.e., all combinations of fatigued and unfatigued, active and inactive MUs) for predicting human power-endurance curves. ET predictions were only validated based on fitted model parameters from experimental data. A limitation of this approach is that similar contractile properties were assumed for all MUs. Since MUs are assumed to only have four possible discrete states, there is no allowance for muscle fatigue to vary as a continuous function of time. Conversely, another recent model by James and Green (2012) was used to predict muscular power output during maximal exertions and ET during submaximal contractions. In their model, muscle contractile properties are assumed to vary as a continuous function of time and MU type (a continuum of MUs included the full range of twitch speed from the slowest to the fastest). Their MFM accounts for recruitment of MUs, but does not capture the influence of MU firing rate on power output. Further, the model parameters were obtained using curve fitting to experimental data, and the model was not validated for other conditions.

## 2.4. A Direct Comparison of Two Theoretical Muscle Fatigue Models

As noted above, diverse approaches have been used to generate theoretical MFMs, and specifically to predict task-related declines in muscle force capacity. Many of these approaches, though, are considered here as impractical for occupational application, due to their complexity and/or computational inefficiency. Those MFMs intended for predicting responses to muscle stimulation are also considered not applicable, since, as noted, muscular responses to electrical stimulation differ from those during voluntary contractions (Gregory & Bickel, 2005). Two specific MFMs are compared below, and both are considered occupationally relevant (i.e., predict responses to voluntary contractions, computationally efficient, not overly complex). Further, direct comparison between the two is possible, since respective model parameters have been reported for the same muscle group. These two models were compared under different loading conditions, with a goal of identifying conditions in which the models generate similar vs. divergent outputs. The latter could serve as a basis for generating testable hypotheses that can be assessed in subsequent work, and also aid in either refinement of current MFMs or development of a new MFM.

The first modeling approach is that of Ma et al. (2009). Based on their previous work, parameters (rates) for fatigue and recovery of hand-grip muscles were set to be 1.123 $min^{-1}$ (Ma et al., 2011), and 2.4 $min^{-1}$ (Ma et al., 2010), respectively. The second model is that of Xia and Frey Law (2008), for which optimal values of fatigue (0.00980) and recovery (0.00064) parameters were reported for hand-grip muscles using a grid-search approach with modified Monte Carlo simulations (Frey-Law et al., 2012). Simulations using both models were generated using MATLAB™ (R2012a, v.7.14, MathWorks, Natick, MA, USA) and the dynamic system simulation (SIMULINK).

#### 2.4.1. Loading conditions and model outputs

Two distinct isometric loading conditions were examined, including both isotonic and non-isotonic efforts, and with endurance time and time-dependent muscle force capacity as respective outcome measures. First, the models were compared for very simple loading, involving sustained isometric, isotonic exertions. Second, intermittent isometric contractions were used to examine time-varying model outputs under different target load conditions. The latter type of loading (intermittent) can also help to investigate outcomes for different job/task sequences, and also address other occupationally-relevant aspects such as job rotation. Of note, non-isometric contractions were not included, given the challenges involved in accounting for relevant physiological aspects (e.g., length-tension and velocity-tension relationships). Such contractions, though more practically relevant, can be addressed in subsequent research. Preliminary results using both the Ma et al. (2009) and Xia and Frey Law (2008) models are presented below, for the selected loading conditions (all involving the hand-grip muscles).

#### 2.4.1.1. Sustained isometric exertions and endurance time

ET for a range of exertion levels were simulated using both models. Specifically, exertion levels were varied from 10-100% MVC in 10% increments. Model predictions were also compared with the experimental results of Manenica (1986), by estimating ETs at the same exertion levels using the author's reported curve-fit. These results had not been used for parameter identification in the two theoretical MFMs, and therefore were used as an independent source of data for evaluation purposes. Both of the theoretical models predicted similar ETs for each level of exertion (Figure 2.1). Model predictions of ET were highly correlated ($R^2 = 0.98$), and both model predictions had a high correlation with experimental data ($R^2 = 0.89$). Of note, though, both models substantially over predicted ET at the lowest contraction level (10% MVC).

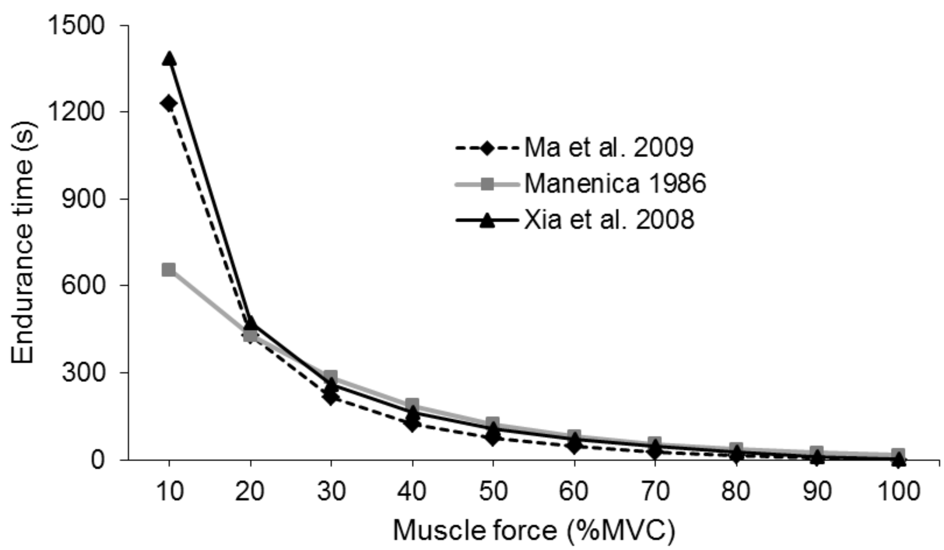



Figure 2.1. Comparison of results from two theoretical (Ma et al. 2009, and Xia et al. 2008) and one empirical model (Manenica 1986), for endurance time during sustained isometric exertions involving hand grip.

#### 2.4.2.2. Intermittent isotonic exertions and job rotation

Two intermittent tasks were simulated, and which were intended to capture a moderate range of occupational task demands. Task ‘A’ required 50% MVC and had a 50% duty cycle, and task ‘B’ required 30% MVC and had a 30% duty cycle. Both tasks had a 50 sec cycle time, and simulations were completed over 400 sec. Two task sequences were simulated that differed in rotation frequency: ‘AABBAABB’ and ‘ABABABAB’. The former sequence was also compared to a sequence with the same rotation frequency but different task order (or starting task): ‘BBAABBAA’. Of note, all three sequences had the same cumulative physical demands (Figure 2.2). Simulation results for both models regarding muscle force capacity over time are depicted in Figures 2.3 and 2.4, respectively comparing the effects of rotation frequency and task order. Model predictions differed over time and at the completion of the task sequences. Force capacity predicted by the Ma et al. (2009) model was about ~20% MVC higher at end of all three sequences, and neither model predicted an effect of rotation frequency or task order on final differences in force capacity.

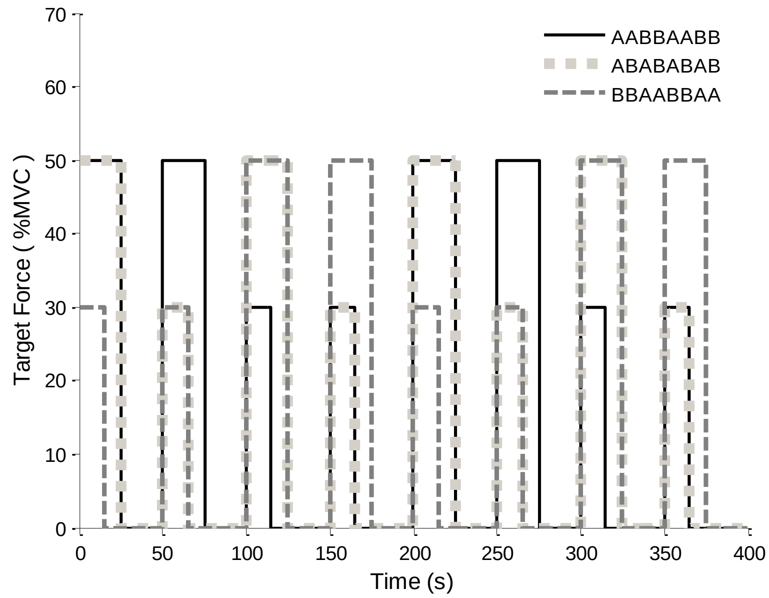


Figure 2.2. Illustration of the three task sequences simulated, involving intermittent isometric target forces (as a % of MVC) for two tasks (A and B; see text for details).

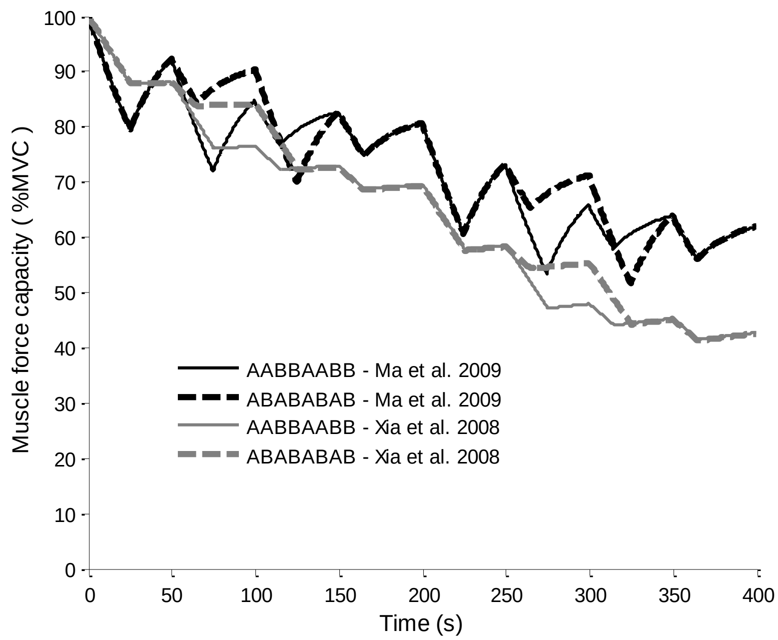


Figure 2.3. Predictions of muscle force capacity (as a % of MVC) in response to two task rotation frequencies, using the Ma et al. (2009) and Xia et al. (2008) MFMs. Task sequences are shown in Figure 2.2.

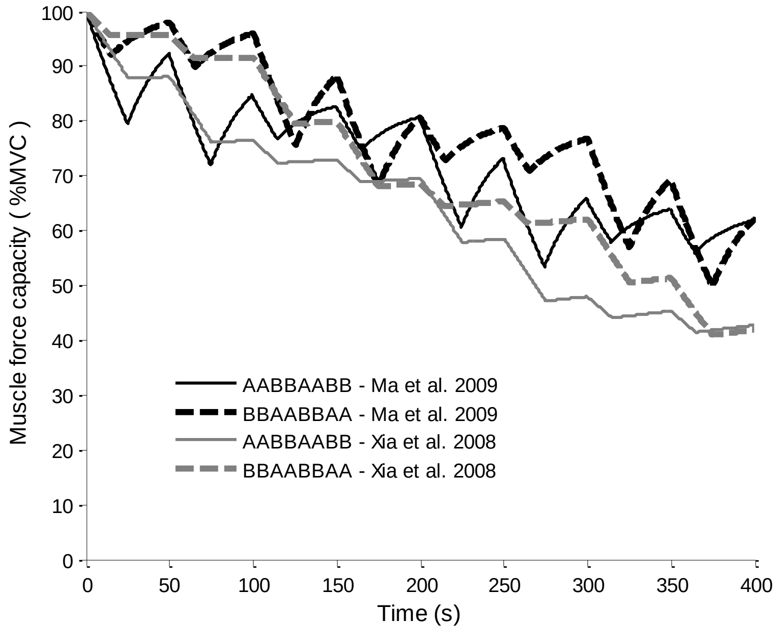



Figure 2.4. Predictions of muscle force capacity (as a % of MVC) in response to two task orders (or staring task), using the Ma et al. (2009) and Xia et al. (2008) MFMs. Task sequences are shown in Figure 2.2.

### 2.4.2. Summary of model results

Two theoretical MFMs were compared under several simulated loading conditions to assess their level of conformity and, more importantly, to identify potential discrepancies between model outcomes. The models generated very similar predictions of ET during sustained isometric exertions (Figure 2.1), though with divergence evident at the lowest levels tested (i.e., 10% MVC, and which both models over-predicted ET). Model predictions were more distinct for intermittent exertions. Neither model predicted final (after 400 sec) differences in force capacity with changes in rotation frequency or task order, though both predicted differences during the simulation period (Figs 3 and 4). Further, the Ma et al. (2009) model predicted higher levels of muscle force capacity for all three sequences and over most of the simulation period. Overall, the current analyses, while preliminary, suggests that there can be important differences in predictions generated by the two MFMs examined, and that such differences may depend on the specific loading conditions or tasks demands. It is also notable that model parameters used here were adopted from prior studies. Subsequent work is recommended to assess the sensitivity of predictions to these model parameters.

## 2.5. Discussion and Conclusions

A variety of approaches have been reported for modeling human muscle fatigue, and which were categorized here as empirical or theoretical. Reviewing these, and highlighting potential advantages and limitations in each approach, may be of benefit for developing and applying practical (i.e., ergonomically relevant) methods in future work. While empirical models can be tailored for a specific situation, existing evidence suggests that they may not be highly accurate in conditions other than those used for developing the models (i.e., lack of generalizability), and are unable to explicitly consider underlying physiological mechanisms involved in muscle fatigue. Theoretical models, on the other hand, may be more useful, as they have the ability to span a broader range of application, and they have been found to make reasonable predictions over a range of loading conditions. Thus, theoretical modeling may be a more promising future approach to facilitate ergonomic assessments, for example by integration within digital human simulations. Theoretical models, though, are based on certain assumptions, which are yet to be well supported and limit applicability, and existing efforts at testing model predictions have been rather limited. Thus far, most theoretical MFMs have only been evaluated using the same data (or datasets) as used for parameter identification (i.e., no independent comparisons or predictions were done). In some cases, these MFMs have been evaluated only under isometric conditions.

In this paper, example comparisons were obtained for two theoretical MFMs that were considered the most ergonomically-relevant of current alternatives, and which also had readily available parameters to enable a direct comparison. While predictions from these models were qualitatively similar, they differed quantitatively, especially during more complex loading conditions. The noted similarities likely resulted from the underlying modeling approaches, in that both involved first-order differential equations and solutions to these types of equations have exponential form. The fact that model parameters were obtained for sustained isometric contractions may account for obtaining similar outcomes for sustained isometric exertions (i.e., endurance times at different levels of exertion). However, the use of similar model parameter values (from isometric conditions) in more complex loading conditions (i.e., intermittent isometric contractions), might explain the divergence in model predictions in the latter. Sustained exertions differ physiologically from intermittent efforts, for example in terms of the recovery processes involved in the more complex loading situations. Future research is needed to facilitate modeling of such complex condition (e.g., intermittent, non-isotonic, and/or non-isometric), though as further discussed below it will likely be challenging to model recovery processes.

Identifying such differences between model outcomes is suggested as a useful approach, both for developing testable hypotheses and in guiding subsequent model development or refinement. In the following, we offer a number of other approaches that may be of value in such future modeling efforts, specifically by expanding model structures either in an implicit or explicit manner. Of particular interest are factors related to individual differences such as age, gender, and obesity, as well as including task-related parameters, accounting for specific muscle groups involved, and utilizing different types of MUs in a model structure."

Given the increasing proportion of older workers (Harrigan, 2004), age-related effects on the development of LMF have received increasing attention. Some studies have reported a lower rate of LMF development (i.e., higher fatigue resistance) among older adults at the same normalized effort level (Allman & Rice, 2002; Yassierli et al., 2007). In contrast, others found the opposite effect (Baudry et al., 2007), and yet others found similar fatigue resistance for different age groups (Lindström et al., 1997). Differences in study design and the type of task involved have been suggested to explain these mixed outcomes. More specifically, an age-related increase in fatigue resistance is commonly observed with isometric contractions, while contrary results are observed more often during dynamic exertions. Indeed, a recent systematic review indicated that contraction mode is a major factor influencing fatigue resistance with aging (Avin & Frey Law, 2011). Loss of muscle fibers with age, especially type II (fast twitch) fibers, likely contributes substantially to age-related differences in LMF development (Lexell et al., 1983). Alterations in muscle fiber composition may also account for age-related differences between different types of muscle contractions (i.e., isometric vs. dynamic contractions), since fast-twitch muscle fibers are more involved during dynamic contractions. Moreover, differing age-related changes in alternative muscle groups may explain the muscle dependency of such age effects in dynamic contractions (Yassierli & Nussbaum, 2009). Further, the effects of age on fatigability can also be influenced by gender (Hicks et al., 2001).

Consistent with a generally lower rate of LMF development (Avin et al., 2010; Hunter, 2009), females typically have a lower proportion of type II muscle fibers (Miller et al., 1993). The magnitude of gender differences in fatigue resistance also depends on what muscle group is involved (Avin et al., 2010). Moreover, females generally have less muscle mass, leading to less vascular occlusion and oxygen requirements for a submaximal effort at a comparable proportion of MVC, both of can result in increased fatigue resistance (Hicks et al., 2001). Notably, some of these gender differences (i.e., in muscle mass and fibers proportions) are also found with aging, and which may explain the observed moderating effect of gender on the capacity of aging muscles (Hicks et al., 2001). Thus, incorporating diversity in fiber types in a MFM could account for gender differences in fatigability and the interactive effects of age and gender.

Obesity is another personal factor that may influence LMF, and which in turn may need to be considered in future MFMs. Additional experimental evidence is necessary, however, since there has been disparity in the results of previous studies. Higher rates of fatigue have been found with obesity (Maffiuletti et al., 2007), whereas another recent study found no significance difference (Cavuoto & Nussbaum, 2013). The latter authors attributed the lack of an obesity effect to potential differences in the specific level of obesity between studies. Further, physiological studies indicate an impairment of oxidative enzyme activity and increased lipid content in obese subjects (He et al., 2001), while others found a reduced percentage of type I and an increased percentage of type IIb muscle fibers in obese individuals (Tanner et al., 2002). Incorporating different muscle fibers in the process of developing MFMs could therefore account for differences in LMF related to obesity.

Existing evidence (briefly summarized above) indicates important differences in fatigue between individuals, as well as different muscle groups within an individual. Alterations and/or differences in muscle fiber composition is a consistent theme underlying such differences, and is thus a promising moderator that can serve in future MFM enhancement. A previous model explicitly simulated muscle fibers in some detail (Hawkins & Hull, 1993), yet is considered not occupationally-relevant given the resultant complexity in model structure. However, MFMs based on the general patterns of MU recruitment could incorporate different type of MUs (slow, fast fatigue-resistant, and fast fatigable), as was proposed by Xia and Frey Law (2008). Doing so would implicitly account for differences between muscle fiber types, since: 1) muscle fibers innervated by an α-motor neuron (within a MU) appear to have identical characteristics, and 2) each muscle fiber is innervated by only one MU (Burke et al., 1973). Thus, instead of using different types of muscle fibers in a MFM, using different type of MUs could be simpler (i.e., modeling a higher level in the neuromuscular system hierarchy). Using diverse types of MUs in a MFM could also explicitly capture classical "size principle" in MU recruitment (Henneman et al., 1965), since the activation order of different MU pools could be defined accordingly. Moreover, adjusting the proportions of different types of

MUs in a MFM, along with their physiological characteristics, could account (at least in part) for individual difference (e.g., age and gender) that influence muscle fatigue. Notably, accounting for differences between muscle groups may be of benefit, or more generally expanding MFMs to the muscle level (vs. only modeling at the joint level). Such an approach could, for example, assess joint physical capacity independently for agonistic vs. antagonistic muscle activities at a given joint.

More generally, force generated by skeletal muscles during a voluntary contraction is controlled by the CNS through modulating the number of recruited MUs and through rate coding, or the discharge rate at which these MUs are activated (Kernell, 2006). Fuglevand et al. (1993) developed a theoretical model of MU recruitment and rate coding that was used in several subsequent studies (Barry et al., 2007; Dideriksen et al., 2010; Keenan & Valero-Cuevas, 2007). This model included two sub-models of motor neurons and MU-force relationships, and provided the capability to manipulate MU properties and predict their effects on the surface EMG signal. While useful for investigating the EMG-force relationship (Keenan & Valero-Cuevas, 2007) and force variability (Barry et al., 2007), this model could only be used to simulate isometric contractions. Expanding this model could address other aspects, however, such as age-related differences in discharge–rate characteristics (Barry et al., 2007), and discharge–rate variability (Moritz et al., 2005).

Of note, EMG signal properties (e.g., median frequency) could be used to experimentally validate these types of models, since arrays of EMG electrodes can be used to obtain the

MU firing rate spectrum using EMG signal decomposition algorithms (De Luca and Contessa, 2012). Moreover, previous studies have shown that EMG can help in determining muscle fiber type composition. In particular, EMG can classify muscle fibers into distinct categories of “fast” and “slow” fibers (Kupa et al., 1995; Wretling et al., 1987). However, most of these studies were conducted *in vitro*, and others have suggested that such an approach is not sufficiently reliable, given the currently unclear role of many confounding factors (Larivière et al., 2003). More recently, Dideriksen et al. (2010) modified the Fuglevand et al. (1993) model to simulate the interaction between motor neurons and muscle fibers during a fatiguing contraction, and which could reproduce some experimental findings regarding fatiguing contractions (e.g., recruitment of MUs and decreases in discharge rate, and relationships between ET and target force). This suggests that future MFMs could be enhanced by incorporating knowledge regarding MU physiology during fatiguing contractions, such as alterations in MU recruitment and rate coding.

Future research should focus on several aspects related to MFMs, such as identifying appropriate parameter values for use in existing MFMs of different muscle groups/joints, and accounting for participant characteristics (e.g., gender and age). Recently, Frey-Law et al. (2012) used their previous meta-analysis of 194 static experiments (Frey Law & Avin, 2010) to identify model parameter values for use in a MFM (Xia & Frey Law, 2008) of joints such as the knee and elbow. Similarly, Ma et al. (2013) and Zhang et al. (2014) respectively determined subject- and gender-specific parameters for another MFM (Ma et al., 2009). Such information facilitates the use of MFMs in practice, and aids in incorporating them in other modeling/simulation approaches such as in models of force generation and digital human simulations.

Future research is also needed to better model recovery, from both theoretical and experimental perspective. A majority of previous studies have focused on the mechanisms related to muscle fatigue development, typically under relatively simple conditions. Yet, a better understanding of recovery will be needed to model more complex conditions such as intermittent work. Recovery, such as during rest, depends substantially on the history of muscle loading, but such history-dependence remains largely unquantified, and rarely modeled. Use of MFMs including a recovery model, such as in Ma et al. (2009) and (Xia & Frey Law, 2008), are a promising approach, yet additional investigation and evaluation are clearly needed given the complex relationships involved.

Future work is also needed in terms of evaluating existing MFMs. Several models of muscle fatigue have been developed using maximum force or power generation capacity. Maximal voluntary contraction force or power output remains one of the principal methods for assessing muscle fatigue, since force or power are the end outcomes of a series of events during force generation (Vøllestad, 1997). Any decline in force or power can indicate impairments in central and/or peripheral fatigue mechanisms, making these measures a logical first choice for assessing muscle fatigue. Typically, MVC is achieved by instructing a participant to generate their highest possible effort. Voluntary force generation, however, can be limited by several factors, such as posture (Mathiowetz et al., 1985), motivation and mechanisms related to inhibitory effects at different levels of the CNS (Gandevia et al., 1995). As such, for any specific posture, achieving "true" measures of muscle force generating capacity may not be feasible using voluntary efforts, even with continuous encouragement and feedback. Therefore, S.C. Gandevia et al. (1995) suggested a distinction between MVC and maximal evocable force (MEF), the latter of which can be measured using muscle or nerve electrical stimulation (Gandevia, 2001). Utilizing both MVC and MEF may help to improve the development of MFMs, and perhaps even allow for separate predictions of fatigue-related force decline caused by central vs. peripheral mechanisms.

Most existing MFMs were developed for isometric contractions. Use of isometric efforts avoids several complexities, such as accounting for physiological aspects such as length-tension and velocity-tension relationships. In contrast, dynamic contractions are more relevant to typical occupational demands, and MFMs would have more potential for practical use if they were applicable for such efforts. Indeed, recent authors of models applicable to dynamic tasks (James & Green, 2012; Ma et al., 2012; Marion et al., 2010; Sih et al., 2012) acknowledge this need, in part due to the failure of older models in predicting muscle fatigue in diverse conditions. Meanwhile, ergonomics approaches and tools are typically more focused on "higher-level" aspects of biomechanical systems (e.g. muscle physical capacity), and for practical application they need to avoid excessive complexities (e.g., of physiological mechanisms). There thus appears to be a conflict between complexity and applicability, for which no available solution is clear at present. However, as knowledge continues to be gained regarding the complex LMF process, it seems likely that improved MFMs, that are occupationally-relevant, will be developed in the future.

In summary, MFMs have the potential to enhance our knowledge regarding the development LMF, and to serve as an ergonomic tool by predicting LMF under a range of loading conditions.

Prior MFMs were reviewed, each of which has advantages and limitations. Two specific ergonomically-relevant MFMs were directly compared under a few loading conditions, to identify differences in model outcomes. Determining such differences is suggested as a useful approach for guiding subsequent model development or refinement. Other potential methods for improving future MFMs were also suggested, including expansion of the model structure using factors related to individual differences and task-related parameters, and utilizing different types of MUs in the model structure. Predicting LMF using a MFM, while accounting for effects of important individual differences, can contribute to future design/evaluation of work tasks, with the end goal of controlling the development of LMF and associated adverse consequences on performance and injury risk. As such, future work in this area is encouraged.

## References

Allman, BL, & Rice, CL. Neuromuscular fatigue and aging: central and peripheral factors. *Muscle & Nerve* 2002;25(6):785-796.

Appell, HJ, Gloser, S, Soares, J, & Duarte, JA. Structural alterations of skeletal muscle induced by ischemia and reperfusion. *BAM-PADOVA* 1999;9(5):263-268.

Avin, KG, & Frey Law, LA. Age-Related Differences in Muscle Fatigue Vary by Contraction Type: A Meta-analysis. *Physical Therapy* 2011;91(8):1153-1165.

Avin, KG, Naughton, MR, Ford, BW, Moore, HE, Monitto-Webber, MN, Stark, AM, Gentile, AJ, & Law, LAF. Sex differences in fatigue resistance are muscle group dependent. *Medicine and science in sports and exercise* 2010;42(10):1943.

Barry, BK, Pascoe, MA, Jesunathadas, M, & Enoka, RM. Rate Coding Is Compressed But Variability Is Unaltered for Motor Units in a Hand Muscle of Old Adults. *Journal of neurophysiology* 2007;97(5):3206-3218.

Baudry, S, Klass, M, Pasquet, B, & Duchateau, J. Age-related fatigability of the ankle dorsiflexor muscles during concentric and eccentric contractions. *European Journal of Applied Physiology* 2007;100(5):515-525.

Bigland-Ritchie, B, Rice, C, Garland, S, & Walsh, M. Task-dependent factors in fatigue of human voluntary contractions. *Advances in experimental medicine and biology* 1995;384:361-380.

Bigland-Ritchie, B, & Woods, JJ. Changes in muscle contractile properties and neural control during human muscular fatigue. *Muscle & Nerve* 1984;7(9):691-699.

Björklund, M, Crenshaw, AG, Djupsjöbacka, M, & Johansson, H. Position sense acuity is diminished following repetitive low-intensity work to fatigue in a simulated occupational setting. *European Journal of Applied Physiology* 2000;81(5):361-367.

Björkstén, M, & Jonsson, B. Endurance limit of force in long-term intermittent static contractions. *Scand. J Work Environ. & Health* 1977;3:23-27.

Brouillette, D, Thivierge, G, Marchand, D, & Charland, J. Preparative study regarding the implementation of a muscular fatigue model in a virtual task simulator. *Work: A Journal of Prevention, Assessment and Rehabilitation* 2012;41:2216-2225.

Bureau of Labor Statistics. (2012). Nonfatal Occupational Injuries and Illnesses Requiring Days Away From Work, 2011. Washington, DC: US Department of Labor.

Burke, RE, Levine, DN, Tsairis, P, & Zajac, FE. Physiological types and histochemical profiles in motor units of the cat gastrocnemius. *The Journal of Physiology* 1973;234(3):723-748.

Carnahan, BJ, Redfern, MS, & Norman, B. Designing safe job rotation schedules using optimization and heuristic search. *Ergonomics* 2000;43(4):543-560.

Cavuoto, LA, & Nussbaum, MA. Obesity-related differences in muscular capacity during sustained isometric exertions. *Applied Ergonomics* 2013;44(2):254-260.

Chaffin, DB. Localized muscle fatigue–definition and measurement. *Journal of Occupational Medicine* 1973;15(4):346-354.

Deeb, JM, Drury, CG, & Pendergast, DR. An exponential model of isometric muscular fatigue as a function of age and muscle groups. Ergonomics 1992;35(7/8):899-918.

De Luca, CJ and Contessa P. Hierarchical control of motor units in voluntary contraction. *Journal of Neurophysiology* 2012;107(1): 178-195.

Dideriksen, JL, Farina, D, Baekgaard, M, & Enoka, RM. An integrative model of motor unit activity during sustained submaximal contractions. *Journal of Applied Physiology* 2010;108(6):1550-1562.

Ding, J, Lee, SCK, Johnston, TE, Wexler, AS, Scott, WB, & Binder-Macleod, SA. Mathematical model that predicts isometric muscle forces for individuals with spinal cord injuries. *Muscle & Nerve* 2005;31(6):702-712.

Ding, J, Wexler, AS, & Binder-Macleod, SA. A predictive model of fatigue in human skeletal muscles. *J Appl Physiol* 2000;89(4):1322-1332.
Edwards, RHT. Human muscle function and fatigue. *Human muscle fatigue: physiological mechanisms* 1981:1-18.
El ahrache, K, Imbeau, D, & Farbos, B. Percentile values for determining maximum endurance times for static muscular work. *International Journal of Industrial Ergonomics* 2006;36(2):99-108.
Enoka, R, & Stuart, D. Neurobiology of muscle fatigue. *J Appl Physiol* 1992;72(5):1631-1648.
Enoka, RM. Mechanisms of muscle fatigue: Central factors and task dependency. *Journal of Electromyography and Kinesiology* 1995;5(3):141-149.
Fitts, RH. Cellular mechanisms of muscle fatigue. *Physiol Rev* 1994;74(1):49-94.
Forde, M, Punnett, L, & Wegman, D. Pathomechanisms of work-related musculoskeletal disorders: conceptual issues. *Ergonomics* 2002;45(9):619-630.
Frey-Law, LA, Looft, JM, & Heitsman, J. A three-compartment muscle fatigue model accurately predicts joint-specific maximum endurance times for sustained isometric tasks. *Journal of Biomechanics* 2012;45(10):1803-1808.
Frey Law, L, & Avin, K. Endurance time is joint-specific: A modelling and meta-analysis investigation. *Ergonomics* 2010;53(1):109-129.
Fuglevand, AJ, Winter, DA, & Patla, AE. Models of recruitment and rate coding organization in motor-unit pools. *Journal of neurophysiology* 1993;70(6):2470-2488.
Gandevia, SC. Spinal and Supraspinal Factors in Human Muscle Fatigue. *Physiol Rev* 2001;81(4):1725-1789.
Gandevia, SC, Allen, GM, & McKenzie, DK. *Central fatigue: critical issues, quantification and practical implications* (Vol. 384: Fatigue). New York: Plenum Press, 1995.
Giat, Y, Mizrahi, J, & Levy, M. A model of fatigue and recovery in paraplegic's quadriceps muscle subjected to intermittent FES. *Journal of Biomechanics* 1996;118(3):357.
Gregory, CM, & Bickel, CS. Recruitment Patterns in Human Skeletal Muscle During Electrical Stimulation. *Physical Therapy* 2005;85(4):358-364.
Häkkinen, K, & Komi, PV. Effects of fatigue and recovery on electromyographic and isometric force- and relaxation-time characteristics of human skeletal muscle. *European Journal of Applied Physiology* 1986;55(6):588-596.
Harrigan, MW. Employment projections to 2012: Concepts and context. *Monthly Lab. Rev.* 2004;127:3.
Hawkins, D, & Hull, ML. Muscle force as affected by fatigue: Mathematical model and experimental verification. *Journal of Biomechanics* 1993;26(9):1117-1128.
He, J, Watkins, S, & Kelley, DE. Skeletal Muscle Lipid Content and Oxidative Enzyme Activity in Relation to Muscle Fiber Type in Type 2 Diabetes and Obesity. *Diabetes* 2001;50(4):817-823.
Henneman, E, Somjen, G, & Carpenter, D. Functional significance of cell size in spinal motoneurons. *Journal of neurophysiology* 1965;28:560-580.
Hicks, AL, Kent-Braun, J, & Ditor, DS. Sex differences in human skeletal muscle fatigue. *Exercise and Sport Sciences Reviews* 2001;29(3):109-112.

Horton, L, Nussbaum, M, & Agnew, M. Effects of rotation frequency and task order on localised muscle fatigue and performance during repetitive static shoulder exertions. *Ergonomics* 2012;55(10):1205-1217.

Hunter, SK. Sex differences and mechanisms of task-specific muscle fatigue. *Exerc Sport Sci Rev* 2009;37(3):113-122.

Hunter, SK, & Enoka, RM. Changes in muscle activation can prolong the endurance time of a submaximal isometric contraction in humans. *Journal of Applied Physiology* 2003;94(1):108-118.

Iridiastadi, H, & Nussbaum, M. Muscular fatigue and endurance during intermittent static efforts: effects of contraction level, duty cycle, and cycle time. *Hum Factors* 2006;48(4):710-720.

Iridiastadi, H, & Nussbaum, MA. Muscle fatigue and endurance during repetitive intermittent static efforts: development of prediction models. *Ergonomics* 2006;49(4):344-360.

James, A, & Green, S. A phenomenological model of muscle fatigue and the power-endurance relationship. *Journal of Applied Physiology* 2012;113(10):1643-1651.

Keenan, KG, & Valero-Cuevas, FJ. Experimentally valid predictions of muscle force and EMG in models of motor-unit function are most sensitive to neural properties. *J Neurophysiol* 2007;98(3):1581-1590.

Kernell, D. *The motoneurone and its muscle fibres*. Oxford; New York: Oxford University Press, 2006.

Kumar, S. Theories of musculoskeletal injury causation. *Ergonomics* 2001;44(1):17-47.

Kupa, E, Roy, S, Kandarian, S, & De Luca, C. Effects of muscle fiber type and size on EMG median frequency and conduction velocity. *Journal of Applied Physiology* 1995;79(1):23-32.

Larivière, C, Arsenault, AB, Gravel, D, Gagnon, D, & Loisel, P. Surface electromyography assessment of back muscle intrinsic properties. *Journal of Electromyography and Kinesiology* 2003;13(4):305-318.

Lexell, J, Henriksson-Larsén, K, Winblad, B, & Sjöström, M. Distribution of different fiber types in human skeletal msucles: effects of aging studied in whole muscle cross sections. *Muscle & Nerve* 1983;6(8):588-595.

Lindström, B, Lexell, J, Gerdle, B, & Downham, D. Skeletal muscle fatigue and endurance in young and old men and women. *The Journals of Gerontology, Series A* 1997;52(1):59-67.

Liu, JZ, Brown, RW, & Yue, GH. A Dynamical Model of Muscle Activation, Fatigue, and Recovery. *Biophysical Journal* 2002;82(5):2344-2359.

Ma, L, Chablat, D, Bennis, F, & Zhang, W. A new simple dynamic muscle fatigue model and its validation. *International Journal of Industrial Ergonomics* 2009;39(1):211-220.

Ma, L, Chablat, D, Bennis, F, Zhang, W, & Guillaume, F. A new muscle fatigue and recovery model and its ergonomics application in human simulation. *Arxiv preprint arXiv:1010.5891* 2010.

Ma, L, Chablat, D, Bennis, F, Zhang, W, Hu, B, & Guillaume, F. A novel approach for determining fatigue resistances of different muscle groups in static cases. *International Journal of Industrial Ergonomics* 2011;41(1):10-18.

Ma, L, Zhang, W, Hu, B, Chablat, D, Bennis, F, & Guillaume, F. Determination of subject-specific muscle fatigue rates under static fatiguing operations. *Ergonomics* 2013;56(12):1889-1900.

Ma, R, Chablat, D, Bennis, F, & Ma, L. Human Muscle Fatigue Model in Dynamic Motions. In J. Lenarcic & M. Husty (Eds.), *Latest Advances in Robot Kinematics* (pp. 349-356): Springer Netherlands, 2012.

Maffiuletti, N, Jubeau, M, Munzinger, U, Bizzini, M, Agosti, F, Col, A, Lafortuna, C, & Sartorio, A. Differences in quadriceps muscle strength and fatigue between lean and obese subjects. *European Journal of Applied Physiology* 2007;101(1):51-59.

Manenica, I. A technique for postural load assessment. *The ergonomics of working postures, Taylor & Francis, London* 1986:270-277.

Marion, MS, Wexler, AS, & Hull, ML. Predicting fatigue during electrically stimulated non-isometric contractions. *Muscle & Nerve* 2010;41(6):857-867.

Mathiassen, S, & Winkel, J. Can occupational guidelines for work-rest schedules be based on endurance time data? *Ergonomics* 1992;35(3):253-259.

Mathiowetz, V, Kashman, N, Volland, G, Weber, K, Dowe, M, & Rogers, S. Grip and pinch strength: normative data for adults. *Arch Phys Med Rehabil* 1985;66(2):69-74.

Miller, AEJ, MacDougall, JD, Tarnopolsky, MA, & Sale, DG. Gender differences in strength and muscle fiber characteristics. *European Journal of Applied Physiology* 1993;66(3):254-262.

Monod, H, & Scherrer, J. The work capacity of a synergic muscular group. *Ergonomics* 1965;8(3):329-338.

Moritz, CT, Barry, BK, Pascoe, MA, & Enoka, RM. Discharge Rate Variability Influences the Variation in Force Fluctuations Across the Working Range of a Hand Muscle. *Journal of neurophysiology* 2005;93(5):2449-2459.

Rohmert, W. Ermittlung von erholungspausen für statische arbeit des menschen. *Internationale Zeitschrift für Angewandte Physiologie enschliesslich Arbeitsphysiologie* 1960;18:123-164.

Roman-Liu, D, Tokarski, T, & Kowalewski, R. Decrease of force capabilities as an index of upper limb fatigue. *Ergonomics* 2005;48(8):930-948.

Sih, B, Ng, L, & Stuhmiller, J. Generalization of a model based on biophysical concepts of muscle activation, fatigue and recovery that explains exercise performance. *International journal of sports medicine* 2012;33(4):258-267.

Sjøgaard, G, & Jensen, B. Muscle pathology with overuse. *Chronic Musculoskeletal Injuries in the Workplace. In: Ranney D, editor. WB Saunders Company: Philadelphia, PA* 1997:17-40.

Sommerich, C, McGlothlin, J, & Marras, W. Occupational risk factors associated with soft tissue disorders of the shoulder: a review of recent investigations in the literature. *Ergonomics* 1993;36(6):697-717.

Sparto, PJ, Parnianpour, M, Reinsel, TE, & Simon, S. The effect of fatigue on multijoint kinematics, coordination, and postural stability during a repetitive lifting test. *J Orthop Sports Phys Ther* 1997;25(1):3-12.

Tanner, CJ, Barakat, HA, Dohm, GL, Pories, WJ, MacDonald, KG, Cunningham, PRG, Swanson, MS, & Houmard, JA. Muscle fiber type is associated with obesity and weight loss. *American Journal of Physiology - Endocrinology And Metabolism* 2002;282(6):E1191-E1196.

Vøllestad, N. Measurement of human muscle fatigue. *Journal of Neuroscience Methods* 1997;74(2):219-227.

Wexler, AS, Jun, D, & Binder-Macleod, SA. A mathematical model that predicts skeletal muscle force. *Biomedical Engineering, IEEE Transactions on* 1997;44(5):337-348.

Wood, DD, Fisher, DL, & Andres, RO. Minimizing fatigue during repetitive jobs: optimal work-rest schedules. *Human Factors* 1997;39(1):83-101.

Wretling, ML, Gerdle, B, & Henriksson-LarsÉN, K. EMG: a non-invasive method for determination of fibre type proportion. *Acta Physiologica Scandinavica* 1987;131(4):627-628.

Xia, T, & Frey Law, LA. A theoretical approach for modeling peripheral muscle fatigue and recovery. *Journal of Biomechanics* 2008;41(14):3046-3052.

Yassierli, & Nussbaum, M. Effects of age, gender, and task parameters on fatigue development during intermittent isokinetic torso extensions. *International Journal of Industrial Ergonomics* 2009;39(1):185-191.

Yassierli, Nussbaum, MA, Iridiastadi, H, & Wojcik, LA. The influence of age on isometric endurance and fatigue is muscle dependent: a study of shoulder abduction and torso extension. *Ergonomics* 2007;50(1):26-45.

Zhang, Z, Li, KW, Zhang, W, Ma, L, & Chen, Z. Muscular fatigue and maximum endurance time assessment for male and female industrial workers. *International Journal of Industrial Ergonomics* 2014;44(2):292-297.

# 3. Mathematical Models of Localized Muscle Fatigue: Sensitivity Analysis and Assessment of two Occupationally-Relevant Models

Ehsan Rashedi and Maury A. Nussbaum

## Abstract

Muscle fatigue models (MFM) have broad potential application if they can accurately predict muscle capacity and/or endurance time during the execution of diverse tasks. As an initial step toward facilitating improved MFMs, we assessed the sensitivity of selected existing models to their inherent parameters, specifically that model the fatigue and recovery processes, and the accuracy of model predictions. These evaluations were completed for both prolonged and intermittent isometric contractions, and were based on model predictions of endurance times. Based on a recent review of the literature, four MFMs were initially chosen, from which a preliminary assessment led to two of these being considered for more comprehensive evaluation. Both models had a higher sensitivity to their fatigue parameter. Predictions of both models were also more sensitive to the alteration of their parameters in conditions involving lower to moderate levels of effort, though such conditions may be of most practical, contemporary interest or relevance. Although both models yielded accurate predictions of endurance times during prolonged contractions, their predictive ability was inferior for more complex (intermittent) conditions. When optimizing model parameters for different loading conditions, the recovery parameter showed considerably larger variability, which might be related to the inability of these MFMs in simulating the recovery process under different loading conditions. It is argued that such models may benefit in future work from improving their representation of recovery process, particularly how this process differs across loading conditions.

**Keywords:** Localized muscle fatigue, fatigue modeling, recovery process, intermittent exertion

* This chapter is now In Press with *PLOS ONE*.

## 3.1. Introduction

Localized muscle fatigue (LMF) is a complex phenomenon that involves reduced muscle force generation capacity and is typically associated with discomfort, pain, and a decline in desired performance. LMF can influence diverse aspects of the neuromuscular system prior to task failure (or, endurance time), and thus has been broadly defined as "any exercise-induced reduction in the ability of a muscle to generate force or power" (Bigland-Ritchie & Woods, 1984; Enoka & Duchateau, 2008). The fatigue-induced reduction in muscle capacity can result from impairments in several central and/or peripheral mechanisms responsible for muscle force generation. These mechanisms are diverse, leading to substantial complexity in the fatigue process, as well as a substantial dependency of LMF on specific loading conditions (Bigland-Ritchie et al., 1995).

LMF development and its consequences (e.g., discomfort and decline in muscle capacity), however, are important concerns in many fields such as rehabilitation, human factors engineering, and occupational health and safety. As examples of the latter, LMF has been argued as a contributing factor to the development of work-related musculoskeletal disorders (Chaffin et al., 2006), suggested to increase the risk for accidents such as falls (Bentley & Haslam, 2001; Hsiao & Simeonov, 2001), and found to compromise performance on precision tasks (Sparto et al., 1997). Again in the occupational domain, it is often of interest to quantify the presence or extent of LMF, as this can be useful for task assessment or redesign, and more generally to determine the extent to which task demands may exceed an individual's capacity. However, it is not practical to measure LMF directly in many situations, particularly during actual task performance. As such, and given the noted dependency of LMF on loading conditions, the use of muscle fatigue models (MFMs) to predict muscle fatigue has broad potential application.

Existing MFMs has been broadly categorized into two types, *empirical* and *theoretical* (Rashedi & Nussbaum, 2015). Empirical MFMs are based on empirical observations and fitting to experimental data. These models are simple and suitable for some purposes (e.g., for a few or small range of task demands), though they suffer from lack of generalizability. Theoretical MFMs, on the other hand, are based on mathematical representations of physiological processes that are either presumed or supported by existing evidence. These models have utilized several approaches for predicting declines in muscle force during diverse fatiguing tasks. Some of these models are particularly relevant to task design or evaluation in occupational settings (see Table 2.1), since they can be easily implemented and their underlying modeling rationale is related to voluntary contractions (and not, for example, muscle activation due to electrical stimulation).

To improve and/or facilitate applicability of these models (such as in existing software and digital human modeling), it is useful to assess and compare the performance of these models under different loading conditions. Identifying conditions in which relatively better or worse model performance exists can serve as a basis for generating and testing formal hypotheses, which may lead to further improving such models in the future. Another useful step toward improving MFMs is to conduct a sensitivity analysis, to determine which input parameters contribute more substantially to output variability or which parameters are more influential in affecting model predictions. Such information can provide a foundation to determine where additional research is needed, for example to better specify model parameters or whether such parameters might be variables vs. constants. However, to our knowledge, outcomes of MFMs have not been compared to evaluate their consistency, nor has the noted sensitivity to model parameters been formally assessed. Such evidence is expected to facilitate more accurate predictions of muscle fatigue and recovery during complex industrial tasks, particularly in a proactive fashion. More effective proactive design, such as by integrating existing or refined MFMs in digital simulations may, in turn, reduce the development of LMF and associated adverse consequences.

## 3.2. Methods

Four contemporary mathematical MFMs were initially considered, which our recent review highlighted as having the most relevance and potential application in occupational settings (i.e., predict responses to voluntary contractions, computationally efficient, and not overly complex) (Rashedi & Nussbaum, 2015). The first model, by Ma et al. (Ma et al., 2009) (MCBZ), consists of two first-order differential equations for fatigue development and recovery with associated constant parameters (F and R).

$$\frac{dQ(t)}{dt} = -F\,\frac{Q(t)}{MVC}\,F_{ext}(t) \tag{3.1a}$$

$$\frac{dQ(t)}{dt} = R\left(MVC - Q(t)\right) \tag{3.1b}$$

where $Q(t)$ is the current muscle capacity. External muscle force ($F_{ext}$) and personal factors, such as maximum voluntary contraction (MVC) and fatigue resistance, are incorporated implicitly (Eq. 3.1a). Fatigue and recovery are modeled separately:

recovery from fatigue can only occur while a muscle is in the resting state. The recovery process is represented as in Eq. 3.1b.

The second MFM, by Xia and Frey-Law (Xia & Frey Law, 2008) (XFL), was based on compartmental theory and divided the pool of motor units (MU) into three compartments: fatigued ($M_F$), activated ($M_A$), and resting MUs ($M_R$). The transfer rate between compartments is proportional to compartment size, leading to three coupled, first-order differential equations. Generated muscle force is proportional to the size of active MU compartment. Similar to the MCBZ MFM, there are two constant parameters in this model (F and R, for fatigue and recovery), and $C(t)$ is the activation-deactivation drive.

$$\begin{cases} \frac{dM_R}{dt} = -C(t) + R.M_F \\ \frac{dM_A}{dt} = C(t) - F.M_A \\ \frac{dM_F}{dt} = F.M_A - R.M_F \end{cases} \tag{3.2}$$

The third MFM is also based on compartmental modeling, though with an additional compartment to the pool of MUs (Sih et al., 2012). Fatigued MUs are considered to have two sub-pools of active and inactive fatigued MUs. An additional motor drive has been simulated accordingly, accounting for the transfer rates between these fatigued MUs. In practice, however, only one motor drive has been considered, which simplifies the model structure to exactly the same structure as the XFL MFM (Xia & Frey Law, 2008). As such, no further assessment of this MFM has been undertaken here. The fourth MFM, by James and Green (James & Green, 2012) (JG), was developed by assuming a continuum of MU twitch speed rather than having two pools of slow and fast twitch MUs. This model does not have any fatigue recovery process, however, so its performance was only evaluated here during prolonged isometric exertions.

These MFMs were implemented in Matlab 13.0 (The Mathworks, Inc. USA) using a numerical approach, which avoided the problem of analytically solving a complex set of differential equations.  For the purpose of simulations, continuous differential equations in the models were transformed into discrete space using the Matlab function "C2D".  In all numerical analyses, a time step of one second was used, yielding a resolution of predicted endurance times of ± 1 sec.

### 3.2.1. Loading conditions

Most previous efforts in fatigue modeling have been devoted to simulating the fatigue process, particularly using prolonged isometric contractions for modeling and validation purposes (see(Rashedi & Nussbaum, 2015) for a review).  However, this type of loading has relatively low occupational relevance, as most work tasks have intermittent resting periods.  Further, prolonged and intermittent fatiguing exertions are fundamentally different, since blood flow is less of a limiting factor during the latter (Vøllestad, 1997), particularly when the ratio of exertion time to rest is not large (Hunter et al., 2009).  As such, a range of intermittent isometric exertions has been considered here, in additional to more simplistic, isometric contractions.

Fatigue/recovery was simulated, as described above, using the XFL and MCBZ MFMs, for 36 different loading conditions.  These conditions include all combinations of: 1) three exertion levels (EL), of 20, 40, and 60% of maximum voluntary contraction (MVC); 2) four duty cycles (DC), of 50, 65, and 80% (intermittent isometric exertions), and 100% (i.e. prolonged isometric exertion); and 3) three cycle times (CT), of 30, 60, and 240 sec (Figure 3.1). These specific task characteristics were chosen to correspond to a range of task parameters likely to occur in occupational settings. Other exertions, such as those involving non-isometric and/or non-isotonic contraction, were not considered due to the increase in loading and modeling complexity involved.

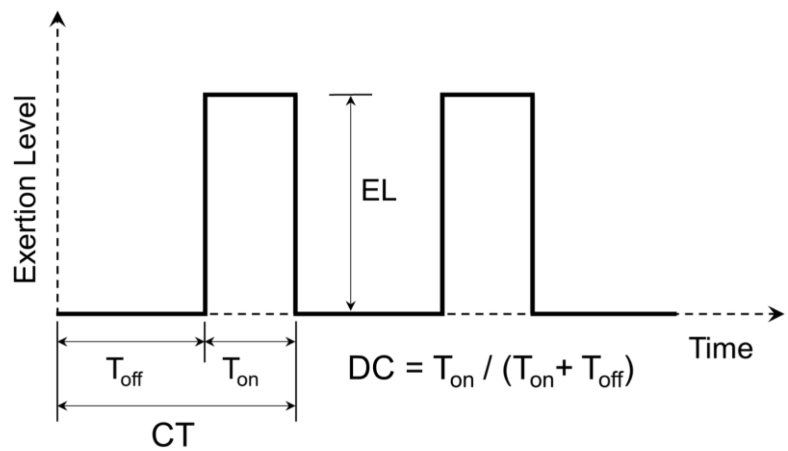



Figure 3.1. Representation of Exertion Level (EL), Duty Cycle (DC), and Cycle Time (CT). $T_{off}$ is the rest time, and $T_{on}$ is the portion of time when an exertion is generated.

### 3.2.2. Sensitivity Analysis

As noted, both the XFL and MCBZ MFMs contain two parameters representing fatigue (F) and recovery (R). The relative contributions of these parameters, or the sensitivity of model estimates to them, was determined by simultaneously varying F and R (21 values of each). Specific parameter ranges investigated, and the baseline values around which they were varied, were chosen based on previously reported values (means ± 1.5 SD) for different muscle groups and joints (Table 3.1) (Frey-Law et al., 2012; Ma et al., 2011). These ranges of parameter values were used to provide a consistent basis for assessing the models, ensuring positive values for the R parameter, and to maximize the potential for these results to be meaningful for several human muscles.

Table 3.1: Parameter baselines, increments, and ranges used for the sensitivity analysis of two muscle fatigue models (MFM).

| MFM | Parameter | Baseline ± Increment | Range |
|---|---|---|---|
| **XFL (Frey-Law et al., 2012)** | F | 0.01090 ± 0.00071 | 0.00390 – 0.01800 |
| | R | 0.00101 ± 0.00007 | 0.00032 – 0.00170 |
| **MCBZ (Ma et al., 2011)** | F | 1.06870 ± 0.04840 | 0.58470 – 1.55280 |
| | R | 0.10250 ± 0.00980 | 0.00500 – 0.20000 |

For continuous functions, partial differential techniques can be used for assessing sensitivity to one or more functional parameters (Hamby, 1994). In this approach, a sensitivity coefficient, $\Phi_i$ can be derived for a particular parameter using:

$$\Phi_i = (\delta Y / \delta X_i) \times (X_i / Y) \quad (3.3)$$

where $X_i$ is a model parameter, Y is the dependent variable, and the quotient $X_i / Y$ is introduced for the purpose of normalization. Here, the parameters F and R are of interest, and endurance time (ET) is used as the dependent variable as a relatively direct approach for assessing predictions generated by a MFM. ET, or the time to task failure, was obtained as the time at which simulated muscle capacity failed to meet or exceed the target exertion level. Model-predicted decrements in force/torque over a fixed period of time, or rates of change over time, are other potential dependent variables. Since our purpose was to assess model sensitivity over a broad range of model parameters, utilizing constant periods of time would have been less informative, and rates of change would be challenging in cases of nonlinear responses that occur during intermittent fatigue/recovery cycles. Of note, 4 hours was used as an upper limit to ET in all analyses, given that longer durations of activity without a rest break were considered rare.

ET, however, is not directly generated by the two MFMs (i.e., there is no analytical formulation or closed-form solution for ET), but rather is only obtained after implementing the models in a task simulation. As such, sensitivity needed to be approximated by evaluating changes in model predictions (i.e., ET) for small variations in parameters (i.e., F and R). For this, $\Phi_i$ was estimated as $\%\Delta ET / \%\Delta X_i$, or the relative change in ET for a relative change in parameter $X_i$. This was determined using (see also Figure 3.2):

$$\Phi_F = [(ET_{m+1}-ET_m)*(F_{m+1}+F_m)] / [(ET_{m+1}+ET_m)*(F_{m+1}-F_m)] \quad (3.4)$$

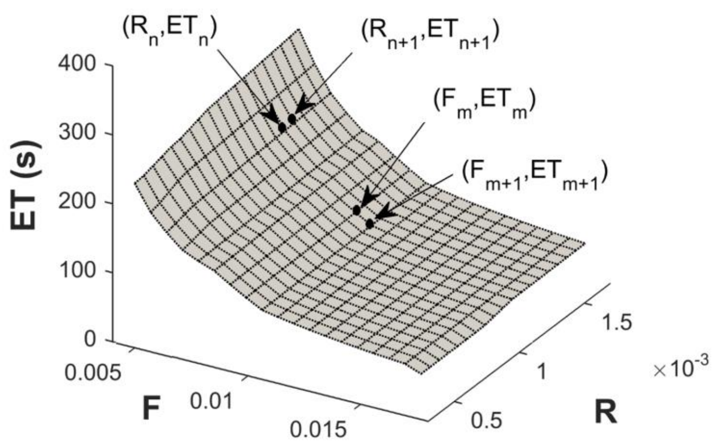



Figure 3.2. Illustration of predicted endurance times (ET) for the XFL model, in a single loading condition (EL = 0.6; DC = 65; CT = 60) and for a range of fatigue (F) and recovery (R) parameters. Methods to derive sensitivity for the F and R parameters, $\Phi_F$ and $\Phi_R$, are also illustrated, and were determined using Eq. (3.4) and (3.5). Sensitivity parameters calculated from the data illustrated in this figure are depicted in Figure 3.6 and 3.7 (3rd row on the left).

$$\Phi_R = [(ET_{n+1}-ET_n)*(R_{n+1}+R_n)] / [(ET_{n+1}+ET_n)*(R_{n+1}-R_n)] \quad (3.5)$$

where *m* and *n* index over the F and R parameters, respectively.

To assess model sensitivities, ET was first determined for each of the 21*21 combinations of F and R parameters; this was done for each of the 36 noted loading conditions (Figure 3.2 shows an example for one loading condition). Next, ET changes with variations of the model parameters were assessed using numerical differentiation. Specifically, for each parameter (i.e., F or R), 20 sensitivity values were calculated for each of the 21 levels of the other parameter, yielding 420 sensitivities (Φ) for each parameter in a given loading condition. This yielded a total of 420 x 36 = 15,150 sensitivity parameters for each of F and R, and in each model.

Preliminary results demonstrated large "spikes" in calculated sensitivity parameters, and which were clearly due to inherent substantial changes in ETs with relatively small changes in parameters during intermittent loading. For example, small alterations in either the F or R parameters can lead to one more or less exertion cycle that can be completed, and in some cases this would represent a considerable relative change in ET

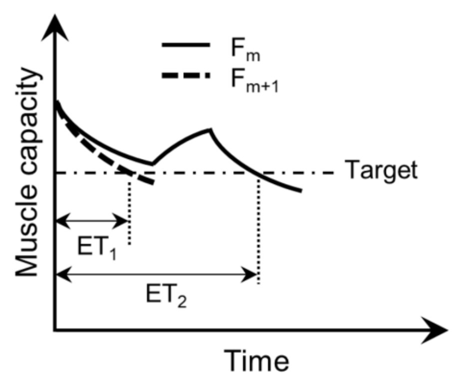



Figure 3.3. Representation of a considerable change in endurance time (ET) that can occur due to relatively small alterations in MFM parameters. Here, $ET_1$ is substantially shorter than $ET_2$ despite a small increment in F (from $F_m$ to $F_{m+1}$).

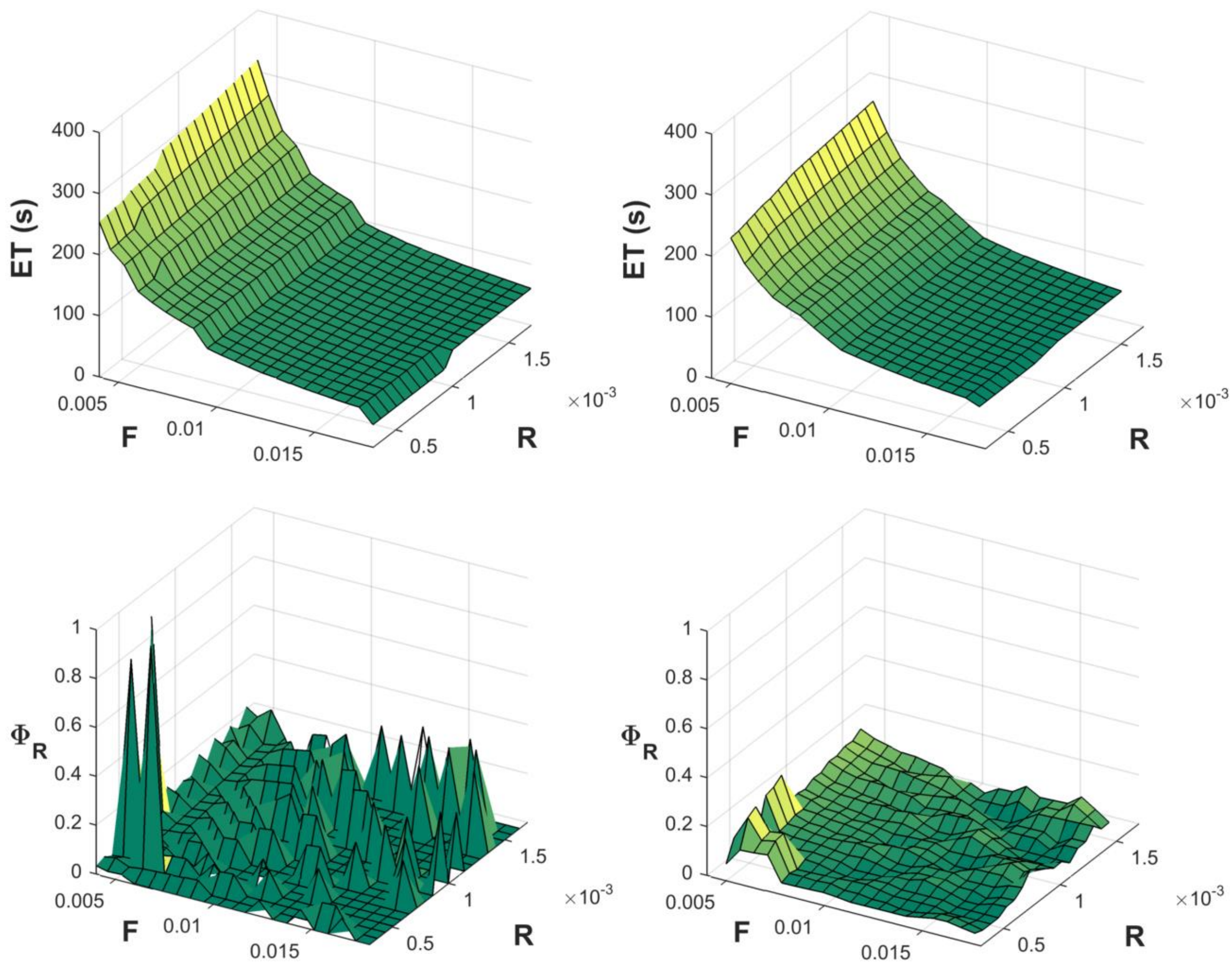



Figure 3.4. Example illustration of the effects of 2D smoothing of endurance time (ET) predictions and the derived recovery sensitivity parameter ($\Phi_R$). Original values of both are depicted in the left figures, and after smoothing on the right. For this illustration, the same model and loading condition was used as in Figure 3.2.

(Figure 3.3). Such substantial ET changes will, in turn, lead to large changes in sensitivity parameters given the differentiation involved. To avoid this, and to better reflect the general patterns of sensitivity, ET values were first "smoothed". This smoothing was done using a mean filter over a rectangle of size 3*3 (i.e., replacing each ET by the mean of itself and all neighboring points), Sensitivity parameters are not reported on the boundaries of the matrix of F and R parameters, both to avoid edge effects and since a full rectangle was not possible. Figure 3.4 illustrates an example of this smoothing procedure and associated effects on the derived recovery sensitivity parameter.

Subsequently, sensitivity parameters were obtained as a function of task parameters (i.e., EL, DC, and CT). For this purpose, the mean sensitivity of one parameter was derived for the midrange value of the other parameter: for each specific combination of task parameters, the mean of eight $\Phi_F$ values was obtained while setting the R parameter to its midrange value. This approach was used to demonstrate the overall trends regarding model sensitivity to the F and R parameters across different loading conditions.

### 3.2.3. MFM Comparisons

To compare the MCBZ (Ma et al., 2009) and XFL (Xia & Frey Law, 2008) MFMs, their ability to predict ETs was assessed for three types of loading conditions, and for which empirical data (ETs) was available from the literature for a consistent muscle group. Specific loading types assessed were: 1) prolonged isometric exertions; 2) intermittent isometric exertions; and, 3) a combination of these two load types. For each loading type, individual model parameters (F and R) were obtained that minimized deviations between model predictions of ETs and existing empirical data, with such error quantified as the root mean square deviation (RMSD). For prolonged isometric loading, the intensity-ET relationship presented by Frey Law and Avin (2010) was used. For intermittent isometric contractions, four empirical studies were available that each involved the hand/grip (Table 3.2). Parameter values for both models were then obtained using an iterative search (over a larger range comparing to Table 3.1), to identify the combination of F and R that produced the least RMSD error. For prolonged isometric contractions, RMSD was minimized across 19 exertion levels (0.1% - 100% MVC, in 5% increments). For intermittent isometric exertions, RMSD was minimized across the four conditions available (Table 3.2). For the combination, RMSD was minimized for the total of 23 conditions (19 isometric + 4 intermittent). Since the iterative search for parameters was done separately for the three loading, three values of RMSD for each model were obtained.

Table 3.2: Empirical studies involving hand/grip exertions that included intermittent isometric loading conditions with reported mean (SD) endurance times (ET).

| **Study** | **EL (% MVC)** | **DC (%)** | **CT (sec)** | **ET (sec)** |
|---|---|---|---|---|
| **(Carpentier et al., 2001)** | 50 | 65 | 20 | 502 (213) |
| **(Fujimoto & Nishizono, 1993)** | 40 | 60 | 10 | 720 (240) |
| **(Fulco et al., 1994)** | 50 | 50 | 10 | 444 (48) |
| **(Pitcher & Miles, 1997)** | 80 | 50 | 10 | 62 16) |

## 3.3. Results

### 3.3.1. Prolonged Isometric Contractions

ET predictions from the XFL, MCBZ, and JC MFMs during prolonged isometric contractions are depicted in Figure 3.5 (see Table 3.3 for optimized parameter values for the former two MFMs). Notably, in MCBZ MFM, recovery process is only active during true rest, thus, R parameter during prolonged isometric contractions is not applicable for this MFM.  Both the XFL and MCBZ MFMs demonstrated substantial consistency with reported ET values (Frey Law & Avin, 2010), and with $R^2$ >0.99 for both.  The JC MFM output was less consistent with these target values ($R^2$ ~ 0.91), and overpredicted and underpredicted ETs for low and moderate effort levels, respectively.  As noted earlier, the JC MFM was not considered further (i.e., for intermittent contractions), since this model does not include a recovery process.

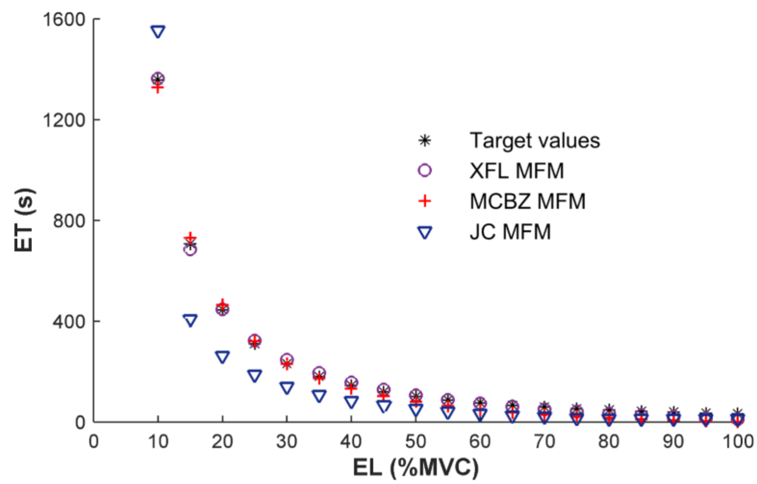



Figure 3.5. Endurance Time (ET) predictions of three MFMs for sustained isometric exertions.  Target values are from data reported by Frey Law and Avin (2010).

Table 3.3: Optimized MFM parameters obtained for prolonged and intermittent isometric exertions, and the combination of both.

| **MFM** | **Prolonged Exertions** | | | **Intermittent Exertions** | | | **Combined Exertions** | | |
|---|---|---|---|---|---|---|---|---|---|
| | F | R | RMSD (s) | F | R | RMSD (s) | F | R | RMSD (s) |
| XFL | 0.0108 | 0.0008 | 14.0 | 0.0162 | 0.0065 | 206.1 | 0.0143 | 0.0014 | 292.3 |
| MCBZ | 1.0400 | NA* | 24.7 | 0.8428 | 0.1930 | 217.3 | 0.9074 | 0.0982 | 265.6 |

* NA: not applicable

### 3.3.2. Sensitivity Analysis

Representative examples of fatigue and recovery sensitivity parameters for both the XFL and MCBZ MFMs are presented in Figure 3.6 and 7, respectively.  Of note, all $\Phi_F$ values are negative, since a larger F parameter leads to smaller ET, and all $\Phi_R$ values are positive, since an increment in R results in a larger ET. In general, and except for some localized steep changes related to the inherent characteristics of intermittent exertions noted above (*cf* Figure 3.3), $\Phi_F$ was relatively "flat" for higher exertion levels. The maximum magnitude of $\Phi_F$ was also larger than $\Phi_R$ in both MFMs. For lower exertion levels, both models were more sensitive to changes of the F parameter (i.e., larger $\Phi_F$) at lower F values and at higher R values. Similarly, both models were more sensitive to R parameter changes (i.e., larger $\Phi_R$) at lower F values and higher R values. Such conditions, with lower F and higher R, are those involving longer ETs. As such, sensitivity to both parameters, in both models, was highest for those tasks with the longer ETs.

Mean sensitivity parameters were obtained as a function of task parameters (i.e., EL, DC, and CT), with representative examples shown in Figure 3.8 (similar results were obtained for the remaining CTs). Both MFMs were more sensitive to alterations of their parameters at lower ELs and DCs, which are conditions involving lower physical demands. Consistent for all CTs, both $\Phi_F$ and $\Phi_R$ were largest for the lower DC (50%) and lowest EL (20% MVC). More generally, $\Phi_F$ and $\Phi_R$ were both larger for lower DCs and ELs.

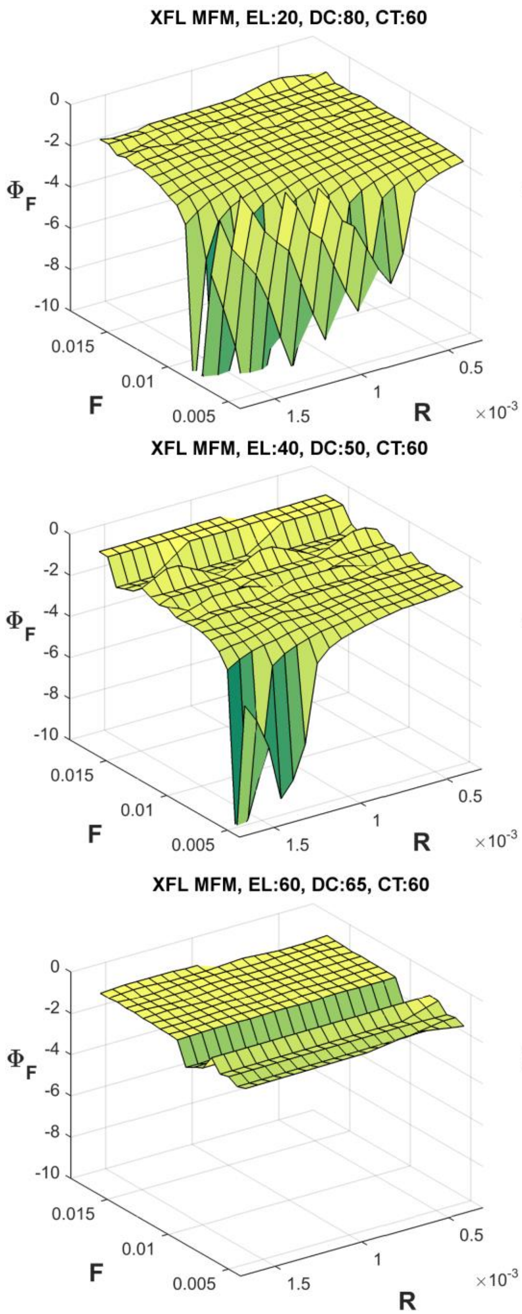



Figure 3.6. Representative examples of fatigue sensitivity parameters (i.e., $\Phi_F$) for the XFL (left) and MCBZ (right) MFMs. $\Phi_F$ values were determined using Eq. (3.4), iterating the F and R parameters over a wide range (Table 3.1). Higher values of $\Phi_F$ indicate larger relative sensitivity to changes in F values. Some $\Phi_F$ values (>10) are not shown, to better illustrate patterns of responses.

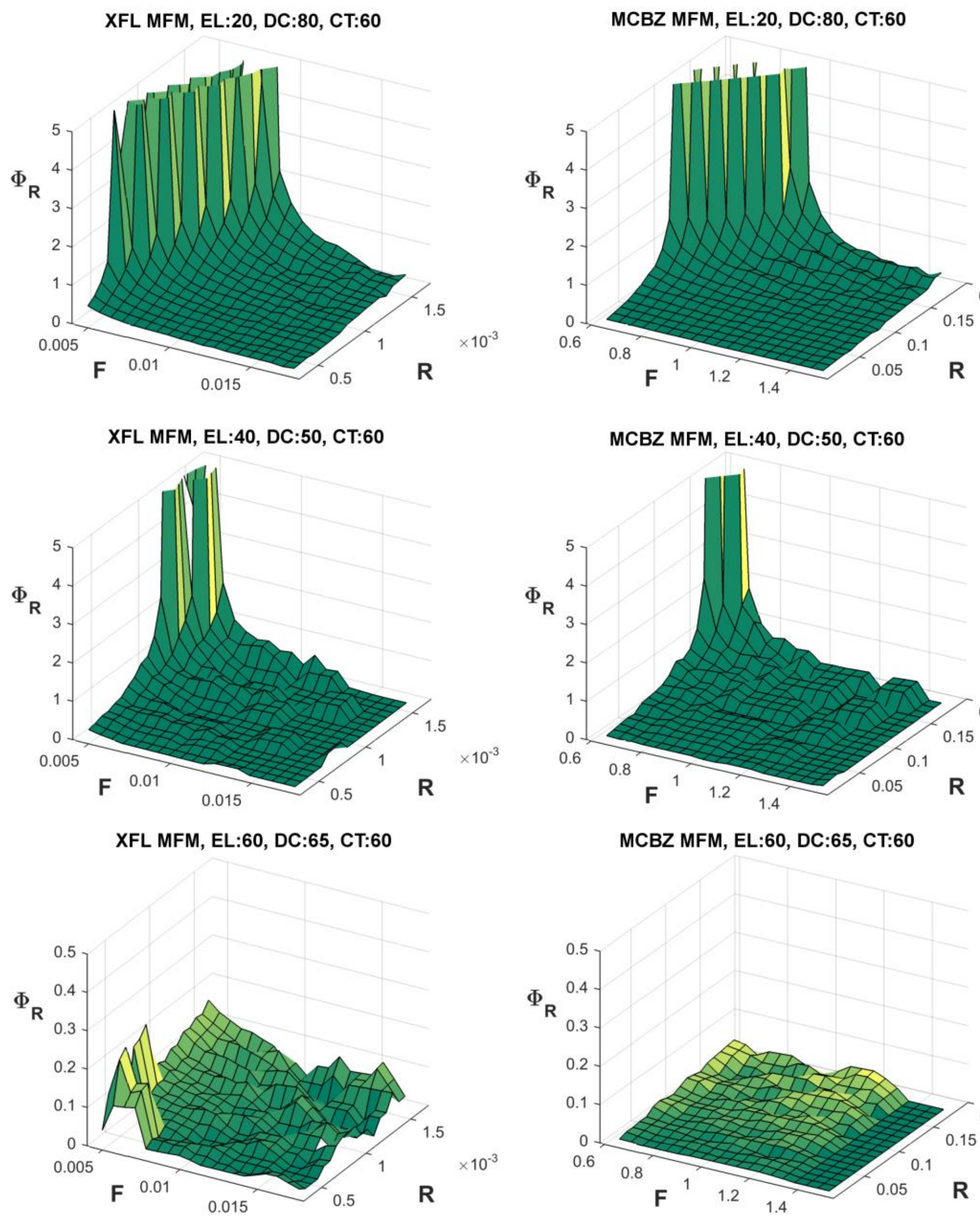



Figure 3.7. Representative examples of recovery sensitivity parameters (i.e., $\Phi_R$) for the XFL (left) and MCBZ (right) MFMs. $\Phi_R$ values were determined using Eq. (3.5), iterating the F and R parameters over a wide range (Table 3.1). Higher values of $\Phi_R$ indicate larger relative sensitivity to changes in R values. Some $\Phi_R$ values (>5) are not shown, to better illustrate patterns of responses.

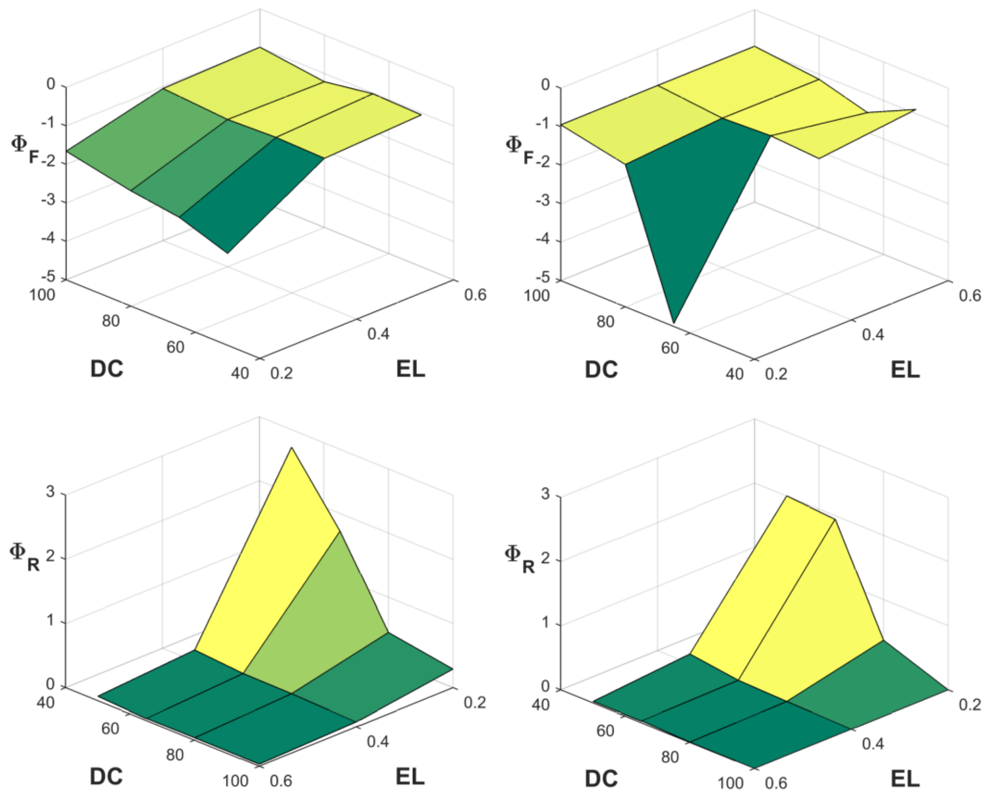



Figure 3.8. Sensitivity values of one model parameter (F and R) at the midrange of the other parameter, for different values of task parameters in the XFL (left) and MCBZ (right) MFMs (CT= 60 s). Note that in this figure the viewpoint is different for $\Phi_F$ and $\Phi_R$, to better visualize the patterns of responses.

### 3.3.3. MFM Comparisons

The XFL and MCBZ MFMs were used to predict ET during three loading types (prolonged, intermittent, and combined), after obtaining optimized parameters for each type. Resulting parameters and associated RMSD values are presented in Table 3.3. The MCBZ MFM yielded a lower correspondence (higher RMSD) between predicted and reported ETs in prolonged and intermittent loading types. For the XFL MFM, the F and

R parameters increased between prolonged and intermittent exertions by ~50 and 700%, respectively. For the MCBZ MFM, there was a ~20% decrease in the F parameter. After fitting the models simultaneously to combined prolonged and intermittent loading conditions, optimized F and R parameter values were between respective values obtained for the two separate loading types. Similarly, RMSD values for both MFMs were largest when fitting to the combined vs. separate loading types.

### 3.4. Discussion

Localized muscle fatigue (LMF) is a complex phenomenon, given the diverse mechanisms that underlie both initiation of and recovery from fatigue. There is substantial value in quantifying LMF, however, since it has potential adverse effects on both performance and injury risk. Given the number of task-related variables (as well as intra- and inter-individual differences) that impact LMF, it is not practical to measure LMF for all possible conditions. Nor is this practical (or often feasible) in the workplace, specifically measuring fatigue for each worker. Therefore, it of interest and potential practical utility to predict LMF development and/or endurance capability using MFMs, given a set of task demands and without the need for direct measurements of LMF (e.g., from an actual worker or using a mock-up). MFMs have been broadly categorized into two types, *empirical* and *theoretical* (Rashedi & Nussbaum, 2015). We focused here on *theoretical* MFMs, because of their broader range of utility, in contract to empirical MFMs that appear mainly useful for a narrower set of specific applications. Based on a previous review of the literature (Rashedi & Nussbaum, 2015), four MFMs were initially chosen, from which a preliminary assessment led to two of these (XFL and MCBZ) being considered for comprehensive analysis. Specifically, this analysis involved assessing the sensitivity of model predictions to alteration of their parameters, and a comparison of the models' ability to predict ET in different loading conditions.

The sensitivity analysis revealed that both models had higher (normalized) sensitivity to the F parameter, suggesting the dominance of the fatigue vs. recovery parameters. Except for some relatively large but localized changes, related to inherent characteristics of intermittent exertions (Figure 3.3), sensitivity parameters were relatively "flat" over the ranges investigated, particularly for higher levels of EL and DC. Both $\Phi_F$ and $\Phi_R$ demonstrated larger values at lower values of F and higher values of R (Figure 3.6 and 3.7). Since lower F and higher R values yield lower rates of fatigue development over time, and thus longer endurance times, both MFMs appear to be more sensitive to their parameters in less demanding loading conditions. This same outcome was observed from averaged values of sensitivity parameters across different task conditions (Figure 3.8), in which both models were more sensitive at lower values of EL and DC. Lower EL and DC again indicate less demanding loading conditions, in which the fatigue process would be relatively less active compared to the more active recovery process. A larger sensitivity of models for "easier" tasks may exacerbate the challenges in predicting LMF for such tasks. In the occupational domain, MFMs are probably most useful at predicting LMF for these lower demanding tasks, since, compared to more physically demanding conditions, such tasks are less often the target of task analyses and redesign to reduce or eliminate the hazardous exposure, for example through automation or use of assistive devices (Rashedi et al., 2014; Rempel et al., 2010).

Regarding their ability to predict ET during prolonged isometric conditions, both the XFL and MCBZ MFMs were able to nearly duplicate empirical intensity-ET values (Frey Law & Avin, 2010). During more complex intermittent contractions, the optimized F parameter in the XFL MFM (Xia & Frey Law, 2008) increased ~50%, while the R increment was substantially larger (i.e., ~700%) (Table 3.3). Relatively smaller alterations of the F parameter between prolonged and intermittent contractions might suggest that a comparable fatigue process is involved in both contraction types, such as due to similar excitation-contraction processes. In fact, Xia (Xia, 2014) assumed such a similarity in developing his recently modified MFM (discussed below). In contrast to the fatigue process, however, fundamental differences exist in the recovery process between prolonged and intermittent contractions. During rest periods in intermittent contractions, blood flow increases (Vedsted et al., 2006), resulting in muscle reperfusion (Zhang et al., 2004). Faster removal of metabolites (e.g., lactic acid from prior muscle fatigue) in a complete rest condition may expedite the recovery process, and can account for the higher values of the R parameter predicted for intermittent vs. prolonged contractions. Furthermore, in prolonged contractions and depending on the intensity of exertion, blood flow occlusion may prevent the removal of fatigue byproducts and replacement of oxygen and glucose in muscle (Rowell, 1993). These evidence justify the observation of substantially different recovery process between pronged and intermittent exertions.

To further explore the performance of these models in different loading conditions, additional assessments were completed. Optimized parameters for both MFMs were first obtained for prolonged contractions (Table 3.3), and these parameters were subsequently used to predict ET in intermittent contractions. From this, both models under-predicted

ETs for intermittent loading conditions (by ~80 – 125%). Such under-prediction may have resulted from overestimating the rate of fatigue and/or underestimating the rate of recovery. Result of optimizing the MFM parameters for intermittent contractions (Table 3.3), showed the latter speculation might be more likely. In other words, a deficiency in simulating the recovery process is more probable, since the R parameter in XFL MFM demonstrated much larger changes between the two conditions (i.e., ~700%). As such, the recovery parameter may need to be distinct between – or specified as a function of – different loading conditions. A fundamentally new approach to simulating the recovery process may also be needed.

To address the limitations of the XFL MFM in accurately predicting changes in muscle capacity during more complex loading conditions (e.g., intermittent contractions), recent studies (Looft, 2014; Sonne & Potvin, 2015; Xia, 2014) have introduced new approaches to simulate the recovery process (Xia & Frey Law, 2008). Rather than assuming a constant R parameter, Xia (2014) proposed having R vary based on the exertion level to reflect changes in blood flow in muscle recovery. However, outcomes with this modification were not substantially different, possibly due to over-simplification of the relationship between blood flow and muscle contraction. Looft (2014) introduced a multiplier to the model, specifically to increase the rate of recovery in rest periods (reflecting post-contraction reperfusion). He fit the XFL MFM with the new rest multiplier to empirical data from the literature and reported improvements in predicting muscle fatigue during intermittent contractions. However, substantially larger errors were reported after assessing the performance of this modified version of the model for predicting ETs (Looft, 2014). As such, it was concluded that using the rest multiplier may not be suitable for predicting ETs during intermittent contractions.

More recently, Sonne and Potvin (2015) sought to increase the biological fidelity of the original XFL MFM (Xia & Frey Law, 2008), by modifying recovery and fatigue rates to represent graded physiological MU characteristics. Outcomes of this new modeling approach were compared with those from the original model in two conditions: 1) with original parameter values, and 2) with optimized parameter values for intermittent contractions. These authors demonstrated that, while the modified model can provide better predictions of muscle fatigue than the model with original parameter values, it has similar performance to the original model with parameter values optimized for new experimental conditions. Meanwhile, the new model did not provide good estimations of ET for prolonged isometric contraction, particularly at lower exertion levels (<40% MVC). While these alternative modeling approaches have demonstrated the potential for improved predictions in more complex intermittent contractions, performance was actually compromised for simpler loading conditions (i.e., prolonged isometric contractions). Such outcomes are similar to what was found here, specifically that model predictions are ineffective when done simultaneously for both prolonged and intermittent contractions (i.e., the largest errors were found for mixed loading conditions, a combination of prolonged and intermittent contractions).

MFM can facilitate the prediction of LMF development and/or endurance capability, providing potential benefits for tasks analysis and pro-active assessment and potentially obviating the need for the direct measurements of LMF. Two MFMs with practical utility for application in occupational settings were assessed here, in the context of prolonged and intermittent contractions. As this work examined only static exertions, future studies would benefit from incorporating dynamic contractions, which are more complex due to inherently larger alterations in MU recruitment and blood flow. Another limitation of the current study was the focus on only one muscle group (i.e., hand/grip muscles). Subsequent work should assess MFM performance for other muscle groups, since fatigue and recovery processes are not only task dependent, but also dependent on the muscle group involved (e.g., related to differences in fiber type distribution). Of note, the four studies from which data were used to assess model performance (Table 3.2) did not target consistent muscles group. Specifically, the studies address the first dorsal interosseous, adductor pollicis, and "grip" muscles. Along with typical inter-individual variability, which can be substantial, this increases the variability in ET, given differences in fiber type distributions of the involved muscles. Future work evaluating and comparing MFMs should, ideally use more comparable data sets. Similarly, MFMs should be evaluated to assess their potential to account for important inter-individual differences, such as related to gender and aging.

In summary, two MFMs were assessed here in terms of their sensitivity to inherent model parameters and their ability to predict ET in both prolonged and intermittent exertions. The two MFMs were, in general, more sensitive to the alterations of the F parameter that represents the rate of muscle fatigue. Both models demonstrated a higher sensitivity to their F and R parameters in conditions involving lower to moderate levels of effort, though such conditions maybe those that are of most practical interest in the occupational domain. The ability of these models ability to predict ET was inferior for mixed loading condition (a combination of prolonged and intermittent contractions). When optimizing model parameters for different loading conditions, the R parameter showed considerably larger variability, which might be related to the inability of these MFMs in simulating the recovery process under different loading conditions. For future application, improved model predictions of fatigue and recovery are needed, especially across diverse loading conditions, and a specific focus on an improved representation of recovery processes is recommended.

## References


Bentley, T, & Haslam, R. Identification of risk factors and countermeasures for slip, trip and fall accidents during the delivery of mail. *Applied Ergonomics* 2001;32(2):127-134.

Bigland-Ritchie, B, Rice, C, Garland, S, & Walsh, M. Task-dependent factors in fatigue of human voluntary contractions. *Advances in experimental medicine and biology* 1995;384:361-380.

Bigland-Ritchie, B, & Woods, JJ. Changes in muscle contractile properties and neural control during human muscular fatigue. *Muscle & Nerve* 1984;7(9):691-699.

Carpentier, A, Duchateau, J, & Hainaut, K. Motor unit behaviour and contractile changes during fatigue in the human first dorsal interosseus. *J. Physiol.* 2001;534(3):903-912.

Chaffin, DB, Andersson, GB, & Martin, BJ. *Occupational biomechanics* (4th ed.). Hoboken, N. J.: Wiley-Interscience, 2006.

Enoka, RM, & Duchateau, J. Muscle fatigue: what, why and how it influences muscle function. *J Physiol* 2008;586(1):11-23.

Frey-Law, LA, Looft, JM, & Heitsman, J. A three-compartment muscle fatigue model accurately predicts joint-specific maximum endurance times for sustained isometric tasks. *Journal of Biomechanics* 2012;45(10):1803-1808.

Frey Law, L, & Avin, K. Endurance time is joint-specific: A modelling and meta-analysis investigation. *Ergonomics* 2010;53(1):109-129.

Fujimoto, T, & Nishizono, H. Involvement of membrane excitation failure in fatigue induced by intermittent submaximal voluntary contraction of the first dorsal interosseous muscle. *The Journal of sports medicine and physical fitness* 1993;33(2):107-117.

Fulco, CS, Cymerman, A, Muza, SR, Rock, PB, Pandolf, KB, & Lewis, SF. Adductor pollicis muscle fatigue during acute and chronic altitude exposure and return to sea level. *Journal of Applied Physiology* 1994;77(1):179-183.

Hamby, D. A review of techniques for parameter sensitivity analysis of environmental models. *Environmental monitoring and assessment* 1994;32(2):135-154.

Hsiao, H, & Simeonov, P. Preventing falls from roofs: a critical review. *Ergonomics* 2001;44(5):537-561.

Hunter, SK, Griffith, EE, Schlachter, KM, & Kufahl, TD. Sex differences in time to task failure and blood flow for an intermittent isometric fatiguing contraction. *Muscle Nerve* 2009;39(1):42-53.

James, A, & Green, S. A phenomenological model of muscle fatigue and the power-endurance relationship. *Journal of Applied Physiology* 2012;113(10):1643-1651.

Looft, JM. *Adaptation and validation of an analytical localized muscle fatigue model for workplace tasks.* (Doctor of Philosophy), The University of Iowa 2014.

Ma, L, Chablat, D, Bennis, F, & Zhang, W. A new simple dynamic muscle fatigue model and its validation. *International Journal of Industrial Ergonomics* 2009;39(1):211-220.

Ma, L, Chablat, D, Bennis, F, Zhang, W, Hu, B, & Guillaume, F. A novel approach for determining fatigue resistances of different muscle groups in static cases. *International Journal of Industrial Ergonomics* 2011;41(1):10-18.

Pitcher, JB, & Miles, TS. INFLUENCE OF MUSCLE BLOOD FLOW ON FATIGUE DURING INTERMITTENT HUMAN HAND-GRIP EXERCISE AND RECOVERY. *Clinical and Experimental Pharmacology and Physiology* 1997;24(7):471-476.

Rashedi, E, Kim, S, Nussbaum, M, & Agnew, M. Ergonomic evaluation of a wearable assistive device for overhead work. *Ergonomics* 2014;57(12):1864-1874.

Rashedi, E, & Nussbaum, M. A Review of Occupationally-Relevant Models of Localized Muscle Fatigue *International Journal of Human Factors Modelling and Simulation* 2015;5(1):61-80.

Rempel, D, Star, D, Barr, A, Blanco, MM, & Janowitz, I. Field Evaluation of a Modified Intervention for Overhead Drilling. *Journal of Occupational and Environmental Hygiene* 2010;7(4):194-202.

Rowell, LB. *Human cardiovascular control*: Oxford University Press, 1993.

Sih, B, Ng, L, & Stuhmiller, J. Generalization of a model based on biophysical concepts of muscle activation, fatigue and recovery that explains exercise performance. *International journal of sports medicine* 2012;33(4):258-267.

Sonne, MW, & Potvin, JR. A modified version of the three-compartment model to predict fatigue during submaximal tasks with complex force-time histories. *Ergonomics* 2015:1-14.

Sparto, PJ, Parnianpour, M, Reinsel, TE, & Simon, S. The effect of fatigue on multijoint kinematics, coordination, and postural stability during a repetitive lifting test. *J Orthop Sports Phys Ther* 1997;25(1):3-12.

Vedsted, P, Blangsted, A, Søgaard, K, Orizio, C, & Sjøgaard, G. Muscle tissue oxygenation, pressure, electrical, and mechanical responses during dynamic and static voluntary contractions. *European Journal of Applied Physiology* 2006;96(2):165-177.

Vøllestad, N. Measurement of human muscle fatigue. *Journal of Neuroscience Methods* 1997;74(2):219-227.
Xia, T. Modeling Peripheral Muscle Fatigue Using a Variable Recovery Rate. *Advances in Physical Ergonomics and Human Factors: Part II* 2014;15:231.
Xia, T, & Frey Law, LA. A theoretical approach for modeling peripheral muscle fatigue and recovery. *Journal of Biomechanics* 2008;41(14):3046-3052.
Zhang, Q, Andersson, G, Lindberg, LG, & Styf, J. Muscle blood flow in response to concentric muscular activity vs. passive venous compression. *Acta Physiologica Scandinavica* 2004;180(1):57-62.

## 4. Cycle time influences the development of muscle fatigue at low to moderate levels of intermittent muscle contraction

Ehsan Rashedi and Maury A. Nussbaum

**Abstract**

Localized muscle fatigue (LMF) during a repetitive task can be influenced by several aspects such as the level and duration of exertions. Among these aspects, though, the influence of cycle time remains unclear. Here, the effect of cycle time on LMF and performance was examined for a simple biomechanical system during repetitive static efforts. Participants performed 1-hour trials of intermittent isometric index finger abduction in all combinations of two cycle times (30 and 60 sec) and two exertion levels (15 and 25% of maximum voluntary capacity). Measures of discomfort, performance (force fluctuations), and muscle capacity (voluntary strength and low-frequency twitch responses) were obtained, all of which demonstrated a beneficial effect of the 30 sec cycle time. Specifically, the shorter cycle time led to lower rates of increase in perceived discomfort, lower rates of increase in force fluctuations, lower rates of decrease in voluntary capacity, and smaller changes in twitch responses. These benefits, reflecting less LMF development in the shorter duty cycle, were quite consistent between genders and the two levels of effort. Results of this study can be used to modify current models predicting work-rest allowance and/or LMF, helping to enhance performance and reduce the risk of adverse musculoskeletal outcomes.

**Keywords**: cycle time; localized muscle fatigue; task variation; repetitive exertion; WMSD

## 4.1. Introduction

Localized muscle fatigue (LMF) has received increasing attention in many disciplines, and has been recognized as an important measure in research and interventions aimed at reducing musculoskeletal disorder risks in general, and work-related musculoskeletal disorder (WMSD) risk in the occupational environment specifically. The specific role of LMF in the development of WMSDs is not clear yet, however, and the existence of LMF does not necessarily imply an increased risk of WMSDs (Mathiassen & Winkel, 1992). Within the occupational domain, however, factors such as working posture and sustained muscle contractions that are closely related with muscle fatigue (Lin et al., 2009) have been shown to contribute to soft tissue injuries (Sommerich et al., 1993; Veiersted, 1994). Several theories and conceptual model have also been proposed describing potential links between LMF and WMSD-related mechanisms (Armstrong et al., 1993; Forde et al., 2002; Sejersted & Vøllestad, 1993). For example, fatigue caused by prolonged static postures may lead to muscular imbalances, including the potential for both underused and overused muscles (Forde et al., 2002).

Diverse sources of variability can influence the development of LMF during task execution. Important individual differences include anthropometry, age, gender, fiber type distribution, and fitness status. Specific tasks demands can also be highly influential, on both LMF and LMF-induced changes in task performance. Extensive existing evidence has assessed the influence of task parameters such as the intensity and duration of work, as examples of factors affecting physical exposure (Horton et al., 2012; Iridiastadi & Nussbaum, 2006; Nussbaum et al., 2001; Yassierli & Nussbaum, 2009). A majority of previous studies on muscle fatigue have involved prolonged static contractions, with well-described relationships between effort level and endurance times (El ahrache et al., 2006; Frey Law & Avin, 2010). However, this type of loading has relatively low occupational relevance, as most work tasks have intermittent resting periods (Adamo et al., 2009; Iridiastadi & Nussbaum, 2006). Broadly, intermittent contractions can be characterized based on three task parameters: the exertion level (EL); duty cycle (DC: the ratio between the exertion period and the CT); and cycle time (CT). Consistently, higher ELs and DCs have been found to cause higher rates of LMF development and/or decreased endurance times (Björkstén & Jonsson, 1977; Jørgensen et al., 1988), likely a direct result of the increased total effort generated over a given time interval.

In contrast to the clear effects of EL and DC during intermittent efforts, the influence of CT on LMF development is less clear. However, the extremes of cycle time, specifically very short and long values, are not favorable. Tasks with high levels of repetition are reported to increase the pressure around peripheral nerves, initiate chronic nerve compression, and cause subsequent swelling and impairment of the vascular supply (Byl et al., 1997; Forde et al., 2002; Stauber, 2004). In contrast, sustained, low-level, isometric contractions recruit the same low threshold motor units (MUs) according to the size principle (Henneman et al., 1965). Notably, each time that a MU is activated it generates at least 30% of its maximal capacity (Sjøgaard & Jensen, 1997), even when the overall task demand is less than 30% of capacity. Considering contractions that are sustained for a long period, it is expected to have some overloaded muscle fibers that experience a loss of calcium homeostasis, or the “Cinderella” response of MUs (Sjøgaard & Jensen, 1997). This phenomenon can occur with or following fatigue, and has been linked to soft tissue inflammation (Lovlin et al., 1987). As such, while prolonged static exertions can be harmful, high repetitions with very short cycle times may also be unfavorable (Silverstein et al., 1986; Sommerich et al., 1993). In between these two extremes, a range of CTs can be considered, and is a potential design parameter for occupational tasks. A practical question remains: given a fixed “amount” of workload that has to be generated intermittently for a specific duration (aka the “tension-time product”), how the work and rest should be distributed? In other words, should frequent short rest breaks be given or longer infrequent ones?

Some studies have linked shorter CTs (i.e., more task variation) with a slower development of LMF (Dickerson et al., 2015; Petrofsky et al., 2000; Yassierli & Nussbaum, 2007), while others provided evidence in favor of longer CTs (Byström et al., 1991; Westgaard, 1988). Still other studies (Engström et al., 1999; Moore & Wells, 2005) have indicated that the effect of CT was not significant at controlled levels of EL and DC. As such, the specific effects of cycle time on muscular fatigue development remain unclear. Previous work on intermittent exertions has investigated CT effects for several intermittent tasks, but using biomechanically complex systems such as the shoulder joint (Dickerson et al., 2015; Iridiastadi & Nussbaum, 2006; Mathiassen, 1993). This complexity has potential effects on LMF monitoring, which may lead to unclear or inconsistent outcomes. For example, there may be changes in the synergistic or antagonistic contraction of multiple muscles. To our knowledge, the influence of CT during low to moderate level of exertions has not yet been reported for a biomechanically "simple" joint, for which the specific effects of CT might be more easily identifiable. This study thus investigated the effect of CT for such a simple system, using both subjective and objective measures to assess the development and consequences of LMF. Based on the weight of existing evidence, and the noted adverse physiological effects of sustained contractions, it was hypothesized that under conditions of similar workload a longer CT will negatively influence perceived discomfort, performance, and muscle capacity.

## 4.2. Methods

### 4.2.1. Participants

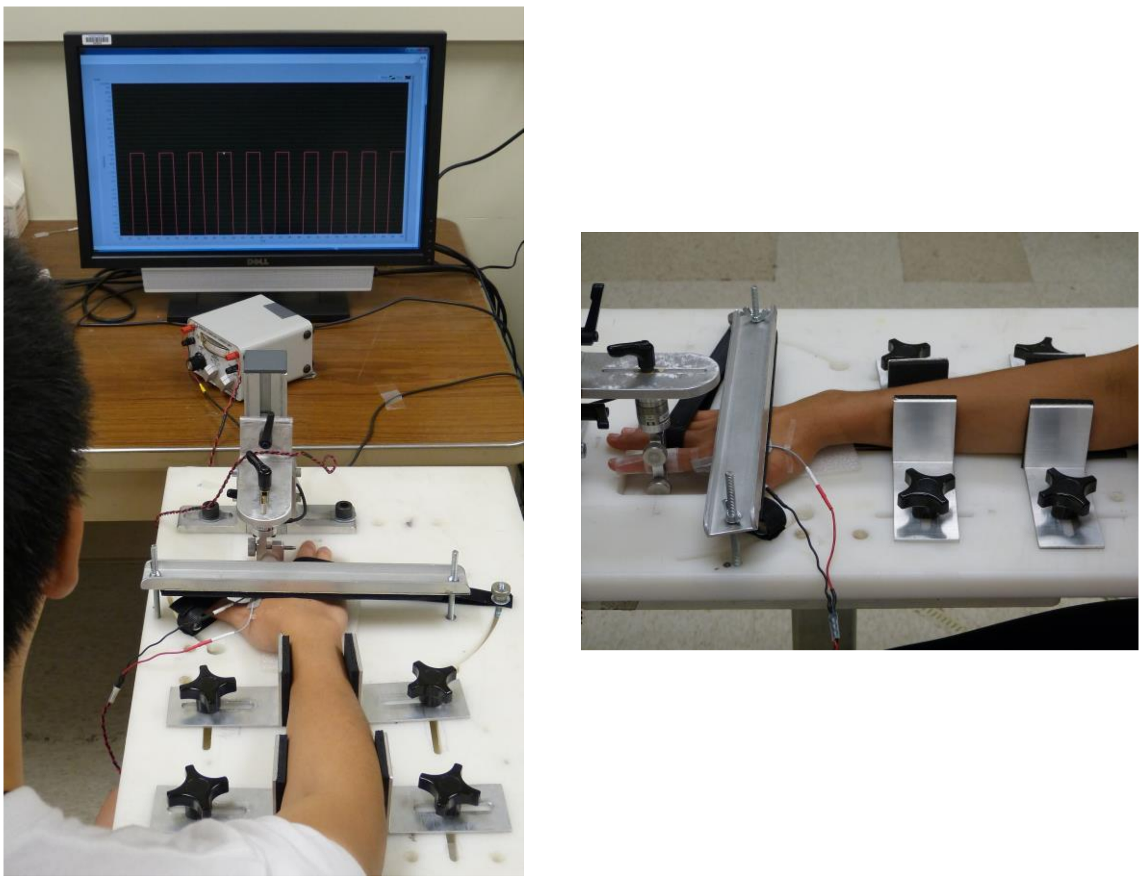
Twelve participants (gender balanced) complete the study, and were recruited from the university and local community. Means (SD) of age, stature, and body mass were 25 (3.3) yrs, 175 (9.6) cm, and 72 (10.2) kg, respectively. All participants reported being at least moderately physically active and having no musculoskeletal disorders or injuries currently or in the preceding 12 months. Prior to any data collection, participants gave informed consent using procedures approved by the Virginia Tech Institutional Review Board.

### 4.2.2. Experimental Design and Procedures

A repeated measures design was used, in which participants completed five sessions including an initial 1-hr practice session, and four subsequent data collection sessions (~ 3 hrs each). Each session was separated by at least two days, to minimize carryover effects due to residual fatigue. Participants performed intermittent isometric exertions in four conditions, involving all combinations of two cycle times (CT = 30 and 60 sec) and two exertion levels (EL = 15 and 25% of maximum voluntary contraction = MVC). For each, a single duty cycle (DC = 50%) was used, and the order of exposure to the four conditions was counterbalanced across participants using 4 × 4 balanced Latin Squares.

These specific task parameters were chosen for two reasons. First, these parameters led to tasks that could all be continued for a reasonable duration (i.e., 60 min). Second, they were intended to represent low to medium levels of occupational exertions for diverse work conditions. A working duration of 1 hour was considered as a shortened version of an actual work-shift, to facilitate implementation in a laboratory setting. All tasks involved index finger abduction, due to the simplicity of movement biomechanics and since the first dorsal interosseous (FDI) muscle is solely responsible for this functional effort.

Upon arrival for both the practice and data collection sessions, participants were calibrated to the 10-point Borg CR-10 scale (Borg, 1990) that was used to provide ratings of perceived discomfort (RPDs). Specifically, participants practiced providing ratings using this scale, while leaning against a wall with their knees bent 90º, as in earlier work (Rashedi et al., 2014; Sood et al., 2007); they provided repeated RPDs for the thighs until ratings reached ≥8. During data collection, participants were seated comfortably in a chair with the examined forearm (dominant arm) resting on a table. The elbow was flexed to 135º, the shoulder abducted, and the forearm pronated with the palm on the table surface (see Figure 4.1a).   Both the hand and forearm rested on the table and were constrained with several restraints and with the thumb fully abducted and blocked (Figure 4.1b).  Adjacent fingers were stabilized using Velcro™ strapping and/or surgical tape (Eichelberger & Bilodeau, 2007; Zijdewind & Kernell, 1994).  Several brackets were used to minimize motions of the hand, wrist, and lower arm, and configured so that the lateral surface of the proximal inter-phalangeal joint of the index finger was centered on a load cell.



(a) (b)

Figure 4.1. (a) Participant performing index finger abduction at 15% MVC with CT = 60 sec.  Participants were instructed to follow the target (red line) with their real-time level of force displayed (white dot); (b) Forces were generated by index finger abduction against a load cell, while the hand and forearm rested on a table with several restraints.

During experimental sessions, LMF was induced by performing index finger abduction in 6 bouts of 10-min intermittent isometric contractions (see Figure 4.2 for an overview). Baseline (pre-fatigue) measures included a minimum of three MVCs (4–5 s duration separated by 1–2 min rest) and low frequency twitches (LFTs). In each MVC trial, participants were asked to maximally activate their FDI muscle in an isometric exertion that involved index finger abduction. The highest value among the completed trials was recorded. Participants were given visual feedback during performance of the intermittent task regarding the level of exertion and the target force (Figure 4.1a); this target force was set to 15 or 25% of MVC. Moreover, the level of exerted force was visually monitored through the course of experiment, and participants were verbally encouraged to correct the level of force in case of deviation from the desired level. Single MVC trials were completed between each of the 10-min bouts of exercise, along with RPDs. Participants were asked about their level of perceived discomfort (emphasizing discomfort in the FDI muscle) right before completing the task. After completing the intermittent task, a single MVC was again completed, followed by LFTs. All participants completed the intermittent contractions, in all four conditions, for the full duration (1 hour).

Low frequency twitches (LFTs) were used, similar to earlier work (Johnson, 1998; Johnson et al., 1995), as an objective method to quantify LMF development during the intermittent tasks. In contrast to MVCs and high-frequency (or tetanic) twitches, this method is less likely to cause muscle fatigue and discomfort (Edwards et al., 1977).

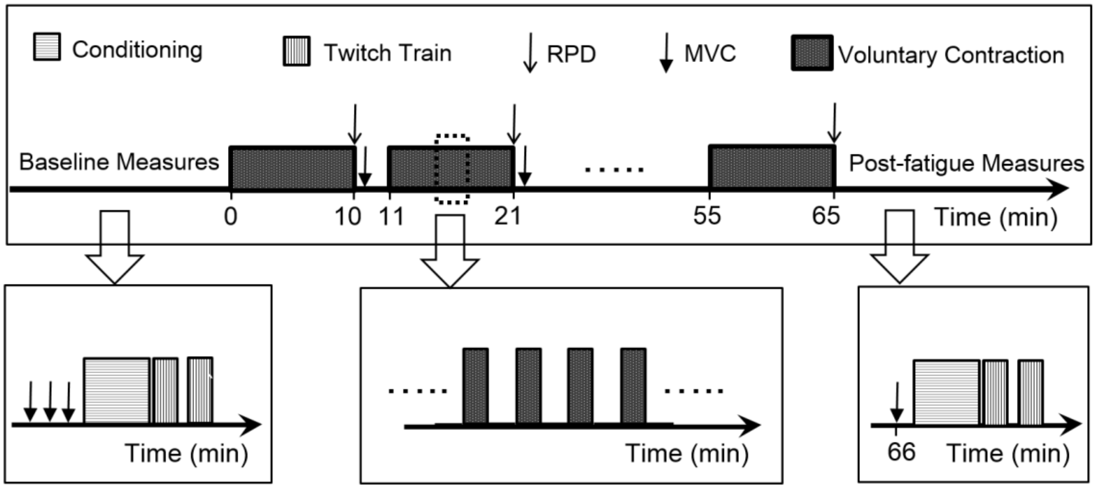



Figure 4.2. Schematic indication of the experimental procedures. The overall procedure is shown at the top, while the lower graphics indicate procedures completed at baseline (left), during the intermittent task (intermittent voluntary contractions, center), and after the task (right). Conditioning involved continuous stimulation at 2 Hz to reach a plateau in twitch force, while twitch trains involved stimulation trains at 2 Hz with rest between trains.

Compared to other methods such as tetanic force, loss of force generation capacity can also be assessed earlier using LFTs (Edwards et al., 1977), and the LFT method may be more sensitive for detecting physiological alterations of a muscle during relatively low intensity activities (Mellor & Stokes, 1992), such as those used here and during long working shifts. In the preliminary session, and after skin preparation involving shaving and cleaning the skin with alcohol, the optimal stimulation location and maximum current were identified using standard procedures as described earlier (Johnson, 1998; Johnson et al., 1995). Stimulation was done using two Ag-AgCl surface electrodes (PALS® Platinum Model J10R00, Axelgaard Manufacturing Co.) that were placed over the FDI muscle. Muscle twitches were evoked using a dual-channel, current-controlled muscle stimulator (Grass S88, AstroMed, Inc., West Warwick, RI) in series with a stimulus isolation unit (SIU5, AstroMed, Inc., West Warwick, RI) and a constant current unit (CCU1, AstroMed, Inc., West Warwick, RI). Current level was set to a level that was perceived as the maximum tolerable for about 3 min, which corresponded to the required time for conditioning the FDI muscle (Adamo et al., 2009; Desmedt & Hainaut, 1968; Rankin et al., 1988). For each participant, the same stimulation site and current were used in all data collection sessions.

On experimental days, skin preparation and electrode placement were done as in the preliminary session. During electrical stimulation, the FDI muscle was first conditioned with 2 Hz twitches for 2-3 min to reach a stable state of twitch force after potentiation (with stimulation characteristics the same as in the preliminary session). Subsequently, two 16-twitch trains were applied to the FDI muscle with a 10 sec between trains. During MVCs and muscle electrical stimulation, exerted forces were sampled at 1000 Hz using a 6 DOF load cell (Nano25-E, ATI Inc., Apex, NC), and low pass filtered at 50 Hz. Of note, electromyographic measures of the FDI were not feasible using surface electrodes since twitches were induced using stimulation of the FDI muscle belly (to avoid any antagonistic stimulation).

### 4.2.3. Dependent measures (DV) and statistical analyses

Both subjective and objective measures were obtained to quantify LMF and performance. The former consisted of the ratings of perceived discomfort (RPD), while the latter included force fluctuations, MVCs, and measures derived from LFT responses. The latter were obtained at baseline and the end of the working period, while RPDs and MVCs were obtained after every 10 min block of intermittent work (see Figure 4.2). To account for substantial individual variability in strength, all MVC values obtained during the intermitted task were normalized as a percentage of pre-task values. Sample force and LFT data are presented in Figure 4.3, with specific procedures for COV and LFT responses provided subsequently.

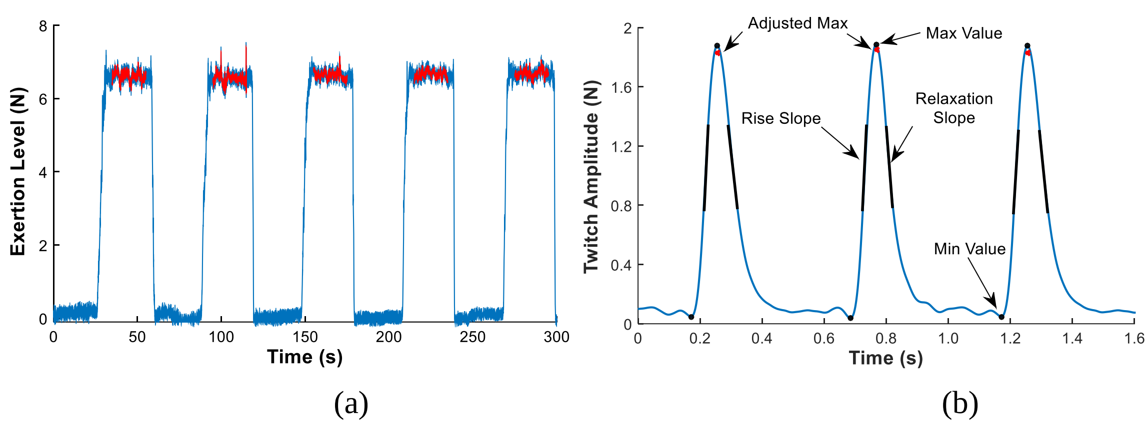



Figure 4.3. (a) A 5-min bout of intermittent isometric exertions (CT = 60 sec); the portion of the signal (highlighted in red) was used for calculating the coefficient of variation (COV); (b) A sample of LFT responses, with LFT magnitude (adjusted max) = max value – min value.

Force fluctuations (or the steadiness of force production) have been used to evaluate the effects of fatigue (Enoka & Stuart, 1992; Hunter et al., 2004), and specifically as a measure of performance (Christou & Carlton, 2002; Enoka et al., 2003; Singh et al., 2010; Tracy & Enoka, 2002). Consistent with these earlier studies, force fluctuations were quantified here using the coefficient of variation (COV), or the standard deviation of the generated force divided by the mean force over a specified time window. COVs were first calculated over segments of force data taken from within the working portion of each cycle. The first and the last 5 sec of each working portion were excluded, to reduce the potential effects of transitional exertions between the rest and work periods (see Figure 4.3a). Then, means of these COVs were obtained over each 10-min block of intermittent work.

Three LFT measures were derived. First, LFT magnitude was calculated from the mean of peak values in the two twitch trains (each included 16 twitches). Of note, pre-twitch forces were not zero in some LFT trials (see Figure 4.3b), due to slight postural deviations. To address this, the maximum twitch force was adjusted by subtracting the minimal pre-twitch value obtained within a 300 msec window prior to the time of maximum twitch force (Figure 4.3b). Both the rise and relaxation rates of twitch forces were respectively determined, as the slopes from linear fits over the rising and lowering phases of individual twitch responses (Blacker et al., 2013; Wilder & Cannon, 2009). The largest slope magnitudes were determined, by obtaining the highest value of the first derivative of the force time series (Blacker et al., 2013). Mean values of LFT responses (magnitude and slopes) were obtained across the two twitch trains completed before and after the intermittent task; DVs were derived as the ratio of these two means (after/before).

Separate repeated measures analyses of variance (RANOVAs) were used to evaluate the effects of CT and EL on RPDs, COVs, and normalized MVCs. Time was included as a continuous covariate in these analyses, supported by qualitative assessments of temporal patterns in all trials. Gender and the order of condition presentation were included as blocking variables in these analyses. Similar RANOVAs were used for LFT measures, though without a Time effect. Significant effects were followed by *post hoc* paired comparisons using Tukey's HSD where relevant. Partial eta-squared ($\eta_p^2$) was used to assess effect sizes, and was qualitatively interpreted as: small (<0.01), moderate (<0.06), large (<0.14) (Cohen, 1988). All statistical analyses were conducted using JMP Pro 11 (SAS Institute Inc., Cary, NC), and statistical significance was determined when $p < 0.05$. Summary results are presented as means (standard deviations).

## 4.3. Results

A summary of statistical results is provided in Table 4.1, for RPD, COV, and MVC. Additional details are provided separately below, for each of these DVs.

### 4.3.1. RPDs

RPDs increased significantly over time, with an overall increase rate of 0.059/min. These changes over time, however, differed between genders, ELs, and CTs (Figure 4.4). RPDs increased at a 40% higher rate for males (0.071/min) versus females (0.047/min), and a 43.5% higher rate in the EL = 25% MVC condition (0.072/min) compared to EL = 15% MVC (0.046/min). There was also a significant difference in the rate of increase between

CT conditions, being 17.3% higher for CT = 60 sec (0.064/min) versus CT = 30 sec (0.054/min).

Table 4.1. Summary of ANOVA results for the main and first-order interaction effects of Gender (G), Time (T), Exertion Level (EL), and Cycle Time (CT) on RPD, COV, and normalized MVC values. No higher-order interaction effects were significant ($p > 0.39$). Significant effects are highlighted in bold, and effect sizes ($\eta_p^2$) are interpreted as small (S), moderate (M), or large (L).

| | **RPD** | | **COV** | | **MVC** | |
|---|---|---|---|---|---|---|
| | ***p* value** | $\eta_p^2$ | ***p* value** | $\eta_p^2$ | ***p* value** | $\eta_p^2$ |
| **G** | 0.45 | 0.050 (M) | 0.95 | 0.001 (S) | 0.94 | 0.001 (S) |
| **EL** | **<0.0001** | 0.790 (L) | 0.22 | 0.021 (M) | **<0.0001** | 0.724 (L) |
| **CT** | **<0.0001** | 0.652 (L) | **0.0081** | 0.243 (L) | **<0.0001** | 0.606 (L) |
| **T** | **<0.0001** | 0.865 (L) | **<0.0001** | 0.659 (L) | **<0.0001** | 0.872 (L) |
| **G×EL** | 0.07 | 0.028 (M) | 0.10 | 0.053 (M) | 0.10 | 0.029 (M) |
| **G×CT** | 0.98 | 0.038 (M) | 0.18 | 0.122 (L) | **0.0015** | 0.204 (L) |
| **EL×CT** | 0.33 | 0.142 (L) | **0.0196** | 0.098 (L) | 0.30 | 0.024 (M) |
| **G×T** | **<0.0001** | 0.208 (L) | 0.21 | 0.124 (L) | 0.88 | 0.144 (L) |
| **EL×T** | **<0.0001** | 0.623 (L) | **0.0005** | 0.266 (L) | **<0.0001** | 0.436 (L) |
| **CT×T** | **0.0433** | 0.441 (L) | **0.0002** | 0.517 (L) | **0.0263** | 0.213 (L) |

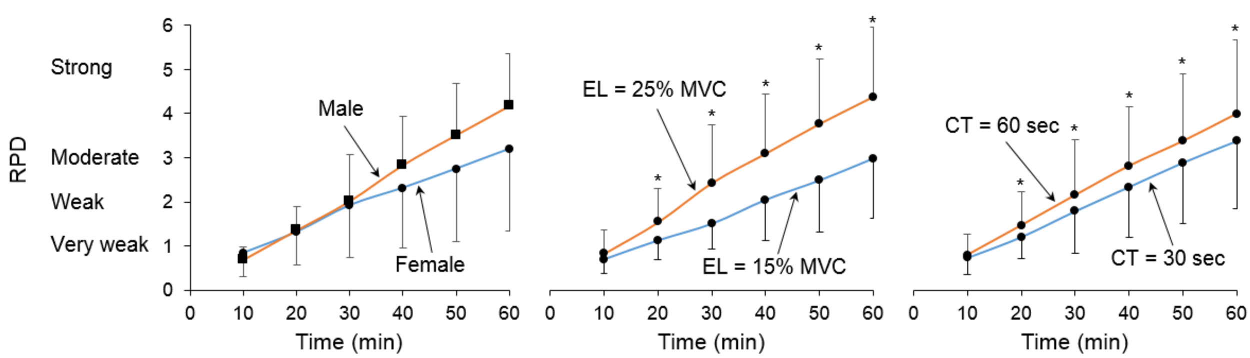



Figure 4.4. Ratings of perceived discomfort (RPD) over time for the two levels of Gender, Exertion Level (EL), and Cycle Time (CT). The symbol * indicates a significant pairwise difference at a given time. Error bars indicate standard deviations.

### 4.3.2. Performance

COV increased significantly over time (mean rate = 0.0001/min), though differed between EL and CT conditions (Figure 4.5). The rate of COV increase was 84.4% higher in the EL = 25% MVC condition (0.000153/min) compared to EL = 15% MVC (0.000062/min). CT also influenced the rate of increase, being 92.9% higher with CT = 60 sec (0.000158/min) versus CT = 30 sec (0.000058/min). There was also a significant CT × EL interaction effect on COV (Figure 4.6); at the higher EL (25% MVC) COV was ~9% lower in the shorter CT, whereas COV was consistent between CT conditions at the lower EL (15% MVC).

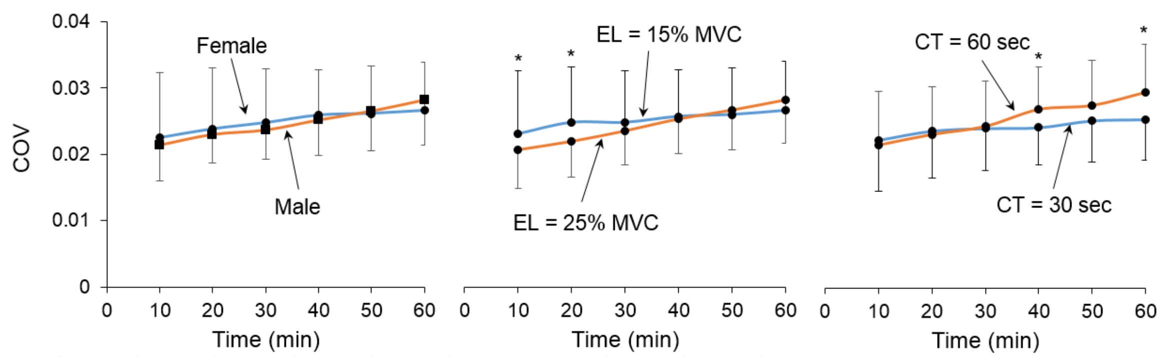



Figure 4.5. Coefficients of variation (COV) over time for the two levels of Gender, Exertion Level (EL), and Cycle Time (CT). The symbol * indicates a significant pairwise difference at a given time. Error bars indicate standard deviations.

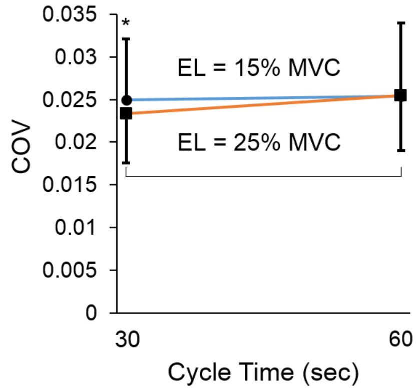



Figure 4.6. Coefficients of variation (COV) for the two Exertion Levels (EL) and Cycle Times (CT). The symbol * indicates a significant pairwise difference between ELs, while the bracket indicates a significant difference between CTs for the EL=25% MVC condition. Error bars indicate standard deviations, and lines are only used to aid visualization.

#### 4.3.3. MVC

Normalized MVCs decreased over time, at an overall rate of 0.165 %MVC/min. This decline of strength over time was similar between genders, but was significantly different between ELs and CTs (Figure 4.7). The rate of decrease was 75.0% higher for the EL = 25% MVC (-0.226 %MVC/min) comparing to EL = 15% MVC (-0.103 %MVC/min). And, the rate of decrease was 39.3% higher for the CT = 60 sec (-0.197 %MVC/min) versus CT = 30 sec (-0.132/min) condition. There was also a significant G × CT interaction effect on normalized MVC (Table 4.1); females demonstrated ~5% lower mean of normalized MVC in the longer CT, whereas males showed ~2% lower mean of normalized MVC for CT = 60 sec.

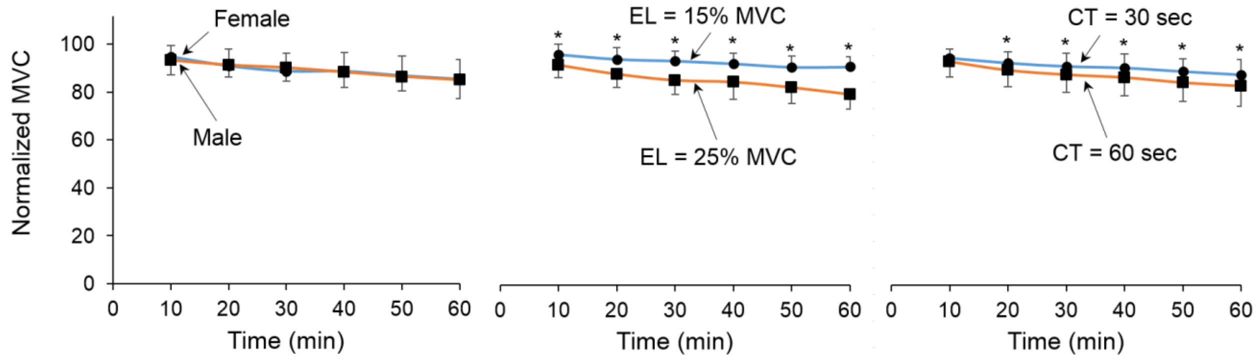



Figure 4.7. Changes in normalized MVC (percentage of pre-task values) over time for the two levels of Gender, Exertion Level (EL), and Cycle Time (CT). The symbol * indicates a significant pairwise difference at a given time. Error bars indicate standard deviations.

#### 4.3.4. Low frequency Twitch (LFT)

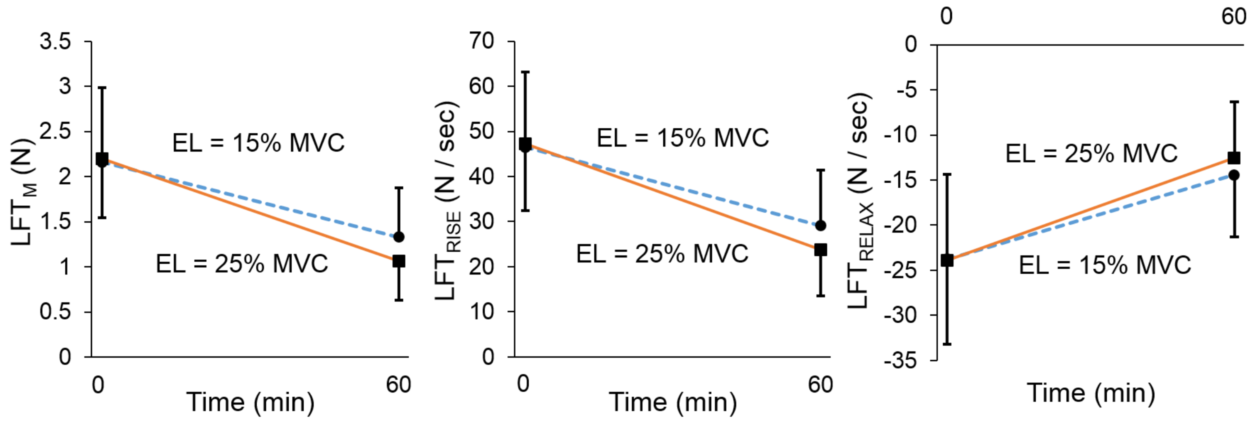
ANOVA results for the effects of Gender, EL, and CT, and their interaction effects on low frequency twitch magnitude ($LFT_{M}$), LFT rise slope ($LFT_{RISE}$), and LFT relaxation slope ($LFT_{RELAX}$) are summarized in Table 4.2. Significant main effects of both EL and CT were found for all three measures (Figure 4.8). Baseline values were nearly identical for the two levels of EL and CT, while all post fatigue values were significantly different between the two levels of each independent variable. More specifically, mean values of $LFT_{M}$, $LFT_{RISE}$, and $LFT_{RELAX}$ were respectively 22.4, 20.3, and 13.9% lower for the EL = 25% MVC condition. Similarly these values were respectively 11.8, 11.7 and 7.6% lower for the CT = 60 sec condition.

Table 4.2. Summary of ANOVA results for the main and interaction effects of Gender (G), Exertion Level (EL), and Cycle Time (CT) on low frequency twitch magnitude ($LFT_M$), LFT rise slope ($LFT_{RISE}$), and LFT relaxation slope ($LFT_{RELAX}$) values. Significant effects are highlighted in bold, and effect sizes ($\eta_p^2$) are interpreted as small (S), moderate (M), or large (L).

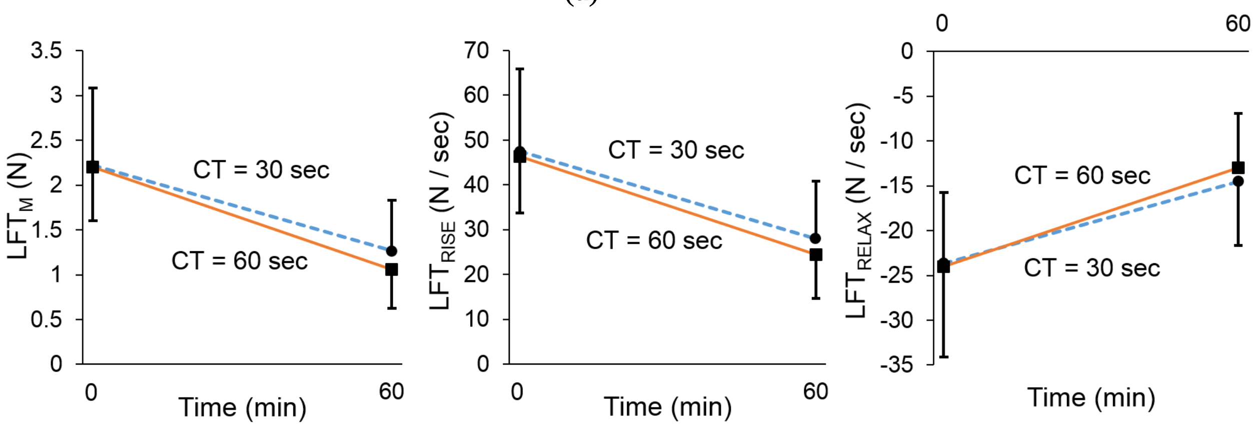
| | **$LFT_M$** | | **$LFT_{RISE}$** | | **$LFT_{RELAX}$** | |
|---|---|---|---|---|---|---|
| | ***p* value** | **$\eta_p^2$** | ***p* value** | **$\eta_p^2$** | ***p* value** | **$\eta_p^2$** |
| **G** | 0.77 | 0.008 (S) | 0.66 | 0.020 (M) | 0.32 | 0.100 (L) |
| **EL** | **<0.0001** | 0.851 (L) | **<0.0001** | 0.808 (L) | **0.0088** | 0.476 (L) |
| **CT** | **0.0024** | 0.455 (L) | **0.0399** | 0.249 (L) | **0.0415** | 0.216 (L) |
| **G×EL** | 0.99 | 0.000 (S) | 0.85 | 0.005 (S) | 0.84 | 0.004 (S) |
| **G×CT** | 0.6 | 0.019 (M) | 0.58 | 0.009 (S) | 0.48 | 0.004 (S) |
| **EL×CT** | 0.52 | 0.094 (S) | 0.49 | 0.071 (M) | 0.88 | 0.005 (S) |
| **G×EL×CT** | 0.99 | 0.000 (S) | 0.56 | 0.051 (M) | 0.68 | 0.039 (M) |

## 4.4. Discussion

The primary goal of this study was to investigate the effect CT for a simple biomechanical system, and specifically to assess the development and consequences of LMF during intermittent isometric exertions. The results provide consistent evidence of an influence of CT on discomfort, changes in muscle contractile status and performance. Before additional discussion of these results, however, it is important to assess whether the current experimental protocols actually induced LMF. Considering perceived discomfort measures, development of LMF is indicated by mean values of 3.7 for post-fatigue RPDs, indicating a higher than “moderate” level of perceived discomfort. MVC is a common measure utilized, and often considered as the “gold standard” in LMF assessment (Vøllestad, 1997). Overall, the mean decline in MVC after the 1-hour exercise was ~15%, greater than a value of 10% suggested as the “normal” variation of muscle strength (Rainoldi et al., 2001). LFTs were also used to assess contractile status and as an objective measure that is independent of voluntary activations, which can be confounded by factors such as the level of motivation. Peak twitch responses ($LFT_M$) decreased substantially (~47%) at the end of the exercise, again supporting the development of LMF.



(a)



(b)

Figure 4.8. LFT measures obtained pre-task (time = 0) and post-task (time = 60 min), for the two levels of: (a) Exertion Level (15% MVC: dashed line and 25% MVC: solid line), and (b) Cycle Time (30 sec: dashed line and 60 sec: solid line). All pairwise differences between pre- and post-fatigue measures were significant, as were all post-fatigue pairwise comparisons. Error bars indicate standard deviations, and lines are used only to aid visualization.

Generating muscle force at higher level of exertion (here, as a % of MVC) is equivalent to a larger physical demand and can be expected to lead to higher rates of LMF development and fatigue-induced consequences. Here, there were consistent effects of EL; these are comparable to extensive earlier evidence and as such are not discussed further. However, for a certain level of effort (i.e., specific EL), the effects of CT were mixed in earlier studies, with some supporting the beneficial effect of shorter CTs (Dickerson et al., 2015; Iridiastadi & Nussbaum, 2006), or equivalently shorter/more frequent rest breaks, while others did not find similar benefits of a shorter CT (Byström et al., 1991; Westgaard, 1988). It is difficult to provide a specific reason for the mixed effects of CT reported earlier, though differences in experimental protocols are likely. To our knowledge, all prior reports of the effect of CT on LMF have involved functional tasks involving biomechanically complex systems such as the shoulder joint (Dickerson et al., 2015; Iridiastadi & Nussbaum, 2006; Mathiassen, 1993). As mentioned earlier, use of such complex systems may make it more difficult to identify specific CT effects, for example due to changes in load sharing between multiple muscles (e.g., changes in agonistic and antagonistic recruitment) and/or within muscles (e.g., changes in recruitment of muscle compartments). These same reasons may explain the inconsistency across measures found in some studies. For example, Iridiastadi and Nussbaum (2006) generally found that a shorter CT resulted in less LMF during intermittent arm abductions, based on EMG mean power frequency, but also that that EMG magnitude (RMS) was an insensitive measure. Consistency of outcomes in current study is thus probably a result of the use of a simple biomechanical system (i.e., metacarpophalangeal joint of the index finger) and secondarily related to the difference in outcome measures used.

All objective and subjective measures here indicated positive effects of a shorter CT. Specifically, the shorter CT resulted in lower rates of LMF development and rates of change in performance (Figures 4, 5, 7, and 8). More specifically, the longer CT (60 sec) yielded higher rates of increase for RPD and COV, and higher rate of MVC decline. Similarly, post-exercise decreases in $LFT_{M}$, $LFT_{RISE}$, and $LFT_{RELAX}$ were larger in the CT = 60 sec condition. Further, the effects of CT on these temporal patterns were generally consistent between genders and the two ELs (Table 4.1).

As mentioned earlier, results of earlier studies regarding the effect of CT are mixed. Effects of CT on subjective measures of discomfort were non-significant in several studies (Dickerson et al., 2015; Iridiastadi & Nussbaum, 2006; Mathiassen, 1993), with only one indicating a significantly larger decline in RPD values for a shorter CT (Yassierli & Nussbaum, 2007). Consistent with our results, though, Yassierli and Nussbaum (2007) found a lower rate of MVC decline with a shorter CT. However, Iridiastadi and Nussbaum (2006) and Dickerson et al. (2015) found no significant differences in MVC declines between different CT conditions. Moreover, three earlier studies reported an increase in endurance time with shorter CTs (Dickerson et al., 2015; Mathiassen, 1993; Yassierli & Nussbaum, 2007), while another found no significant effect (Iridiastadi & Nussbaum, 2006). Importantly, none of the noted studies controlled the “amount” of physical effort, since the focus was on endurance time and thus variable task durations were involved.

Physiological aspects accounting for the observed benefits of a shorter CT are likely related to the larger magnitude of task variation (i.e., change in exposure to workload) and more frequent alterations in recruited motor units (MUs). Task variation can be categorized as involving “temporal” (e.g., change of CT) and “activity” (e.g., job rotation) variations (Luger et al., 2014). More variation in biomechanical exposure is considered as an effective intervention in movements with high levels of repetition (Srinivasan & Mathiassen, 2012), with an objective of decreasing movement similarity. Moreover, there is some evidence that exposure variability is inversely associated with muscle fatigue, with more variation in movement patterns leading to slower LMF development (Madeleine, 2010; Srinivasan & Mathiassen, 2012). In the presence of LMF, the central nervous system (CNS) may “explore” alternative movement solutions, with the goal of varying MU recruitment for a given task (Fuller et al., 2011). This is related to the concept of motor flexibility, or equivalently the redundancy in the musculoskeletal system (Scholz & Schöner, 1999), where the CNS may alter the motor solution for specific task over time (Latash et al., 2002; Rashedi et al., 2010). While variation can decrease the rate of LMF development (Srinivasan & Mathiassen, 2012), it may also be advantageous in avoiding the so-called Cinderella effect, preventing a small pool of MUs from constant activation (Hägg, 1991). Vedsted et al. (2006) also showed that more variation results in larger fluctuations in intramuscular pressure, potentially beneficial for enhancing blood flow and reducing ischemia. Another potential mechanism is related to the rate of $K^+$ release, which is expected to be lower with shorter CTs due to the overall $K^+$ flux into muscle cells during the more frequent rest periods (Byström & Kilbom, 1990). Finally, anaerobic metabolism may be less active with shorter CTs, because of the additional variability (Byström & Kilbom, 1990).

A few limitations in the present work should be noted. Only relatively young and healthy participants were included, and as such future work is needed to determine if the current results are consistent more generally. Only a narrow range of CTs (and ELs) was included here, so any conclusions about the effects of CT must necessarily be somewhat limited. Subsequent work is again needed to evaluate CT effects over a larger range of task conditions, including both shorter and longer CTs, varying duty cycles, higher and lower ELs, different muscles, and more complex force-time patterns. Further, muscle activity (EMG) was not monitored, though its use in future work would help assessing the influence of CT on muscular recruitment. Finally, high-frequency stimulation, commonly used to assess peak contractile capacity, was not used (preliminary testing indicated that it was not tolerable by many participants). As such, our current measures of contractile capacity (i.e., MVCs) may have not been fully representative, and we were unable to assess the contribution of central vs. peripheral sources of LMF.

The present study examined the effect of two cycle times during low to moderate levels of intermittent static exertions. Diverse measures were used, all of which demonstrated a beneficial effect of a shorter cycle time. Specifically, the shorter cycle time led to changes in the rates of change in perceived discomfort, performance, and muscle capacity, as well as task-related changes in contractile status. These differences indicate less fatigue development and fatigue-induced consequences with the shorter cycle time, and were quite consistent between genders and the two exertion levels investigated. Current rest-allowance prediction models (e.g., Rohmert (1960)) do not consider the effect of cycle time, and the present results suggest that these may thus be oversimplified for some applications. The current results may also help in enhancing existing models used to simulate or predict muscle fatigue (e.g., see review in Rashedi and Nussbaum (2015)). Accounting for the effects of cycle time may help in reducing LMF development and thereby help to enhance performance and reduce the risk of WMSDs.

## References


Adamo, D, Khodaee, M, Barringer, S, Johnson, P, & Martin, B. Low mean level sustained and intermittent grip exertions: Influence of age on fatigue and recovery. *Ergonomics* 2009;52(10):1287-1297.

Armstrong, T, Buckle, P, Fine, L, Hagberg, M, Jonsson, B, Kilbom, A, Kuorinka, I, Silverstein, B, Sjogaard, G, & Viikari-Juntura, E. A conceptual model for work-related neck and upper-limb musculoskeletal disorders. *Scand J Work Environ Health* 1993;19:73-84.

Björkstén, M, & Jonsson, B. Endurance limit of force in long-term intermittent static contractions. *Scand. J Work Environ. & Health* 1977;3:23-27.

Blacker, S, Fallowfield, J, & Willems, M. Intra- and interday reliability of voluntary and electrically stimulated isometric contractions of the quadriceps femoris. *Journal of Electromyography and Kinesiology* 2013;23(4):886-891.

Borg, G. Psychophysical scaling with applications in physical work and the perception of exertion. *Scand J Work Environ Health* 1990;16(Suppl 1):55-58.

Byl, N, Merzenich, M, Cheung, S, Bedenbaugh, P, Nagarajan, S, & Jenkins, W. A primate model for studying focal dystonia and repetitive strain injury: effects on the primary somatosensory cortex. *Physical Therapy* 1997;77(3):269-284.

Byström, S, & Kilbom, Å. Physiological response in the forearm during and after isometric intermittent handgrip. *European Journal of Applied Physiology and Occupational Physiology* 1990;60(6):457-466.

Byström, S, Mathiassen, S, & Fransson-Hall, C. Physiological effects of micropauses in isometric handgrip exercise. *European Journal of Applied Physiology and Occupational Physiology* 1991;63(6):405-411.

Christou, E, & Carlton, L. Age and contraction type influence motor output variability in rapid discrete tasks. *Journal of Applied Physiology* 2002;93:489-498.

Cohen, J. *Statistical power analysis for the behavioral sciences*: Lawrence Erlbaum Associates, Inc, 1988.

Desmedt, J, & Hainaut, K. Kinetics of myofilament activation in potentiated contraction: staircase phenomenon in human skeletal muscle. *Nature (London)* 1968;217(5128):529-532.

Dickerson, C, Meszaros, K, Cudlip, A, Chopp-Hurley, J, & Langenderfer, J. The influence of cycle time on shoulder fatigue responses for a fixed total overhead workload. *Journal of Biomechanics* 2015;48(11):2911–2918.

Edwards, R, Hill, D, Jones, D, & Merton, P. Fatigue of long duration in human skeletal muscle after exercise. *J Physiology* 1977;272:769-778.

Eichelberger, T, & Bilodeau, M. Central fatigue of the first dorsal interosseous muscle during low-force and high-force sustained submaximal contractions. *Clinical Physiology and Functional Imaging* 2007;27(5):298-304.

El ahrache, K, Imbeau, D, & Farbos, B. Percentile values for determining maximum endurance times for static muscular work. *International Journal of Industrial Ergonomics* 2006;36(2):99-108.

Engström, T, Hanse, J, & Kadefors, R. Musculoskeletal symptoms due to technical preconditions in long cycle time work in an automobile assembly plant: a study of prevalence and relation to psychosocial factors and physical exposure. *Applied Ergonomics* 1999;30(5):443-453.

Enoka, R, Christou, E, Hunter, S, Kornatz, K, Semmler, J, Taylor, A, & Tracy, B. Mechanisms that contribute to differences in motor performance between young and old adults. *Journal of Electromyography and Kinesiology* 2003;13(1):1-12.

Enoka, R, & Stuart, D. Neurobiology of muscle fatigue. *J Appl Physiol* 1992;72(5):1631-1648.

Forde, M, Punnett, L, & Wegman, D. Pathomechanisms of work-related musculoskeletal disorders: conceptual issues. *Ergonomics* 2002;45(9):619-630.

Frey Law, L, & Avin, K. Endurance time is joint-specific: A modelling and meta-analysis investigation. *Ergonomics* 2010;53(1):109-129.

Fuller, J, Fung, J, & Côté, J. Time-dependent adaptations to posture and movement characteristics during the development of repetitive reaching induced fatigue. *Experimental Brain Research* 2011;211(1):133-143.

Hägg, G. Static work loads and occupational myalgia—a new explanation model. *Electromyographical kinesiology* 1991:141-144.

Henneman, E, Somjen, G, & Carpenter, D. Functional significance of cell size in spinal motoneurons. *Journal of neurophysiology* 1965;28:560-580.

Horton, L, Nussbaum, M, & Agnew, M. Effects of rotation frequency and task order on localised muscle fatigue and performance during repetitive static shoulder exertions. *Ergonomics* 2012;55(10):1205-1217.

Hunter, S, Duchateau, J, & Enoka, R. Muscle fatigue and the mechanisms of task failure. *Exercise and Sport Sciences Reviews* 2004;32(2):44-49.

Iridiastadi, H, & Nussbaum, M. Muscular fatigue and endurance during intermittent static efforts: effects of contraction level, duty cycle, and cycle time. *Hum Factors* 2006;48(4):710-720.

Johnson, P. *The development, characterization and implementation of a technique to measure muscle fatigue during computer use.* (Ph.D.), University of California, Berkeley with the University of California, San Francisco, 1998.

Johnson, P, Lehman, S, & Rempel, D. *Measuring low frequency fatigue with 2 Hertz stimulation. I. Stimulus-related potentiation effects.* Paper presented at the Engineering in Medicine and Biology Society, IEEE 17th Annual Conference,1995.

Jørgensen, K, Fallentin, N, Krogh-Lund, C, & Jensen, B. Electromyography and fatigue during prolonged, low-level static contractions. *European Journal of Applied Physiology* 1988;57:316-321.

Latash, M, Scholz, J, & Schöner, G. Motor control strategies revealed in the structure of motor variability. *Exercise and Sport Sciences Reviews* 2002;30(1):26-31.

Lin, D, Nussbaum, M, Seol, H, Singh, N, Madigan, M, & Wojcik, L. Acute effects of localized muscle fatigue on postural control and patterns of recovery during upright stance: influence of fatigue location and age. *Eur J Appl Physiol* 2009;106(3):425-434.

Lovlin, R, Cottle, W, Pyke, I, Kavanagh, M, & Belcastro, A. Are indices of free radical damage related to exercise intensity. *European Journal of Applied Physiology and Occupational Physiology* 1987;56(3):313-316.

Luger, T, Bosch, T, Veeger, D, & de Looze, M. The influence of task variation on manifestation of fatigue is ambiguous – a literature review. *Ergonomics* 2014;57(2):162-174.

Madeleine, P. On functional motor adaptations: from the quantification of motor strategies to the prevention of musculoskeletal disorders in the neck-shoulder region. *Acta Physiol (Oxf)* 2010;199 Suppl 679:1-46.

Mathiassen, S. The influence of exercise/rest schedule on the physiological and psychophysical response to isometric shoulder-neck exercise. *European Journal of Applied Physiology* 1993;67:528-539.

Mathiassen, S, & Winkel, J. Can occupational guidelines for work-rest schedules be based on endurance time data? *Ergonomics* 1992;35(3):253-259.

Mellor, R, & Stokes, M. Detection and severity of low frequency fatigue in the human adductor pollicis muscle. *Journal of the Neurological Sciences* 1992;108(2):196-201.

Moore, A, & Wells, R. Effect of cycle time and duty cycle on psychophysically determined acceptable levels in a highly repetitive task. *Ergonomics* 2005;48(7):859-873.

Nussbaum, M, Clark, L, Lanza, M, & Rice, K. Fatigue and endurance limits during intermittent overhead work. *AIHAJ-American Industrial Hygiene Association* 2001;62(4):446-456.

Petrofsky, J, Stacy, R, & Laymon, M. The relationship between exercise work intervals and duration of exercise on lower extremity training induced by electrical stimulation in humans with spinal cord injuries. *European Journal of Applied Physiology* 2000;82(5-6):504-509.

Rainoldi, A, Bullock-Saxton, J, Cavarretta, F, & Hogan, N. Repeatability of maximal voluntary force and of surface EMG variables during voluntary isometric contraction of quadriceps muscles in healthy subjects. *Journal of Electromyography and Kinesiology* 2001;11(6):425-438.

Rankin, L, Enoka, R, Volz, K, & Stuart, D. Coexistence of twitch potentiation and tetanic force decline in rat hindlimb muscle. *Journal of Applied Physiology* 1988;65(6):2687-2695.

Rashedi, E, Khalaf, K, Nassajian, MR, Nasseroleslami, B, & Parnianpour, M. How does the central nervous system address the kinetic redundancy in the lumbar spine? Three-dimensional isometric exertions with 18 Hill-model-based muscle fascicles at the L4—L5 level. *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine* 2010;224(3):487-501.

Rashedi, E, Kim, S, Nussbaum, M, & Agnew, M. Ergonomic evaluation of a wearable assistive device for overhead work. *Ergonomics* 2014;57(12):1864-1874.

Rashedi, E, & Nussbaum, M. A Review of Occupationally-Relevant Models of Localized Muscle Fatigue *International Journal of Human Factors Modelling and Simulation* 2015;5(1):61-80.

Rohmert, W. Ermittlung von erholungspausen für statische arbeit des menschen. *Internationale Zeitschrift für Angewandte Physiologie enschliesslich Arbeitsphysiologie* 1960;18:123-164.

Scholz, J, & Schöner, G. The uncontrolled manifold concept: identifying control variables for a functional task. *Experimental Brain Research* 1999;126(3):289-306.

Sejersted, O, & Vøllestad, N. Physiology of muscle fatigue and associated pain. In H. Værøy & H. Merskey (Eds.), *Progress in Fibromyalgia and Myofascial Pain* (pp. 41-51). Amsterdam: Elsevier Science Publishers B.V., 1993.

Silverstein, B, Fine, L, & Armstrong, T. Hand wrist cumulative trauma disorders in industry. *British Journal of Industrial Medicine* 1986;43(11):779-784.

Singh, N, Arampatzis, A, Duda, G, Heller, M, & Taylor, W. Effect of fatigue on force fluctuations in knee extensors in young adults. *Philosophical Transactions of the Royal Society A: Mathematical, Physical and Engineering Sciences* 2010;368(1920):2783-2798.

Sjøgaard, G, & Jensen, B. Muscle pathology with overuse. *Chronic Musculoskeletal Injuries in the Workplace. In: Ranney D, editor. WB Saunders Company: Philadelphia, PA* 1997:17-40.

Sommerich, C, McGlothlin, J, & Marras, W. Occupational risk factors associated with soft tissue disorders of the shoulder: a review of recent investigations in the literature. *Ergonomics* 1993;36(6):697-717.

Sood, D, Nussbaum, M, & Hager, K. Fatigue during prolonged intermittent overhead work: reliability of measures and effects of working height. *Ergonomics* 2007;50(4):497-513.

Srinivasan, D, & Mathiassen, S. Motor variability in occupational health and performance. *Clinical Biomechanics* 2012;27(10):979-993.

Stauber, W. Factors involved in strain-induced injury in skeletal muscles and outcomes of prolonged exposures. *Journal of Electromyography and Kinesiology* 2004;14(1):61-70.

Tracy, B, & Enoka, R. Older adults are less steady during submaximal isometric contractions with the knee extensor muscles. *Journal of Applied Physiology* 2002;92:1004-1012.

Vedsted, P, Blangsted, A, Søgaard, K, Orizio, C, & Sjøgaard, G. Muscle tissue oxygenation, pressure, electrical, and mechanical responses during dynamic and static voluntary contractions. *European Journal of Applied Physiology* 2006;96(2):165-177.

Veiersted, K. Sustained muscle tension as a risk factor for trapezius myalgia. *International Journal of Industrial Ergonomics* 1994;14:333-339.

Vøllestad, N. Measurement of human muscle fatigue. *Journal of Neuroscience Methods* 1997;74(2):219-227.

Westgaard, R. Measurement and evaluation of postural load in occupational work situations. *European Journal of Applied Physiology* 1988;57:291-304.

Wilder, M, & Cannon, J. Effect of age on muscle activation and twitch properties during static and dynamic actions. *Muscle Nerve* 2009;39(5):683-691.

Yassierli, & Nussbaum, M. Effects of age, gender, and task parameters on fatigue development during intermittent isokinetic torso extensions. *International Journal of Industrial Ergonomics* 2009;39(1):185-191.

Yassierli, & Nussbaum, M. Muscle fatigue during intermittent isokinetic shoulder abduction: age effects and utility of electromyographic measures. *Ergonomics* 2007;50(7):1110-1126.

Zijdewind, I, & Kernell, D. Index finger position and force of the human first dorsal interosseus and its ulnar nerve antagonist. *Journal of Applied Physiology* 1994;77(2):987.

# 5. Quantifying the task (or history) dependency of recovery from localized muscle fatigue induced by intermittent index finger abduction

Ehsan Rashedi and Maury A. Nussbaum

**Abstract**

Muscle fatigue and recovery are complex processes influencing muscle force generation capacity. While fatigue reduce this capacity, muscle recovery act to restore the unfatigued muscle state. Many factors can potentially affect muscle recovery, among these may be a task dependency of recovery following an exercise. However, little has been reported regarding the history dependency of recovery after fatiguing contractions. We examined the dependency of muscle recovery subsequent to four different histories of fatiguing muscle contractions, imposed using two cycle times (30 and 60 sec) during low to moderate levels (15% and 25% of maximum voluntary contraction (MVC)) of intermittent static exertions involving index finger abduction. MVC and low frequency twitch measures of muscle capacity were used, all of which indicated a dependency of muscle recovery on muscle capacity state existing immediately after fatiguing exercise. This dependency did not appear to be modified by either cycle time or exertion level. Overall, these results imply that the post-exercise rate of recovery is primarily influenced by the post-exercise muscle state. These results may help improving existing models of muscle recovery, facilitating more accurate predictions of LMF development and thereby help to enhance muscle performance and reduce the risk of injury.

## 5.1. Introduction

Human muscle fatigue and recovery are complex and multifactorial phenomena, stemming from several underlying mechanisms responsible for muscle force generation. During task execution, both fatigue and recovery take place concurrently. Recovery involves processes that act in opposition to fatigue-related mechanisms, with the aim of restoring the unfatigued state of the muscle. These recovery processes – essential for optimal muscle performance – depend on many factors such as blood flow and the availability of the cellular enzymes. Extensive prior research has investigated muscle fatigue, and the task-dependency of fatigue is well documented (Enoka & Duchateau, 2008). For example, different tasks can involve different muscles with diverse contractile properties and will have associated differences in fatigue development. The task dependency of muscle fatigue has also be related to specific task parameters, such as the contraction level, duty cycle, and cycle time in intermittent contractions, and which can have a substantial influence on fatigue development as well (Iridiastadi & Nussbaum, 2006).

In contrast to fatigue, relatively less evidence is available regarding the recovery process. Similar to muscle fatigue, diverse measures can quantify muscle recovery, with each typically indicating different temporal patterns or durations required for full recovery (Westgaard & Winkel, 1996). For example, muscle capacity (strength) appears to recover more rapidly after localized muscle fatigue than either endurance time or contractile properties identified using low-frequency twitches (LFT). Specifically, recovery of pre-fatigue maximum voluntary contraction (MVC) after fatiguing isometric exertions can require up to few hours (Kroon & Naeije, 1991), whereas recovery of endurance time is more prolonged and can last up to few days (Kroon & Naeije, 1991; Sahlin & Ren, 1989). Recovery of LFTs – a measure independent of voluntary activation – also requires longer durations, with full recovery again up to a few days (Edwards et al., 1977; Jones, 1996). Post-fatigue recovery as reflected in these measures typically follows an exponential pattern (Elfving et al., 2002; Reid et al., 1993; Wood et al., 1997), likely reflecting that physiological processes such as heart rate, oxygen uptake, and blood lactate elimination also occur exponentially over time. In turn, differences in the process or timing of recovery measures probably reflect differences in the recovery rates of relevant underlying physiological processes. For example, restoration of MVC after fatiguing contractions is likely faster than restoration of endurance due to a slower restoration of $H^+$ concentration, which limits endurance due to impaired capacity to rephosphorylate ADP (Sahlin & Ren, 1989).

Compared to the task dependency of muscle fatigue, the task dependency of muscle recovery has not been well explored. Blood flow depends on the task characteristics, especially the level of exertion (Fitts, 1994). In fact, blood flow, which is a main factor in the muscle recovery process, is proportionally coupled to muscle metabolic rate (Murrant & Sarelius, 2015). Meanwhile, diverse measures such as contraction strength (Anrep & von Saalfeld, 1935; Hamann et al., 2004), muscular work (Andersen & Saltin, 1985), and oxygen consumption (Bockman, 1983) have found to have linear relationships with metabolic rate. Thus, changes in task characteristics such as contraction level can influence metabolic rate, and thereby blood flow and recovery during an exercise, as long as that exercise involves periods of time during which recovery can occur (e.g., during intermittent exertions). In contrast, high-level muscle contractions increase intramuscular pressure, restricting blood flow (Järvholm et al., 1988; Sejersted et al., 1984). More directly, the dependency of recovery on task characteristics can be inferred from the results of the previous chapter. Specifically, the level of LMF differed after one hour of intermittent exertion completed using two CTs, each at two levels of exertion. Since the overall physical demand for the two CTs (at each exertion level) was constant, differences in final values of LMF can be attributed to the overall recovery process during exercise and in intermittent rest breaks. This evidence thus suggests a clear dependency of recovery processes on the specific task characteristics involved during exercise.

In addition, relatively little has been reported regarding the history dependency of the recovery process, specifically after an exercise is completed, and only a few studies have compared the recovery process between different fatiguing protocols. In one example, Fuglevand et al. (1993) reported that a longer recovery time was required after maintaining 20% vs. 65% MVC efforts until task failure (inability to maintain the target force). A longer recovery period following low-intensity + long duration versus high-intensity + short duration tasks was also in studies of human muscle (Baker et al., 1993) and mouse muscle fibers (Lännergren & Westerblad, 1991). However, these latter studies did not control the "level" of fatigue, and a question remains as whether recovery might depend on the exertion history leading to a given level of muscle fatigue. To our knowledge, only the study of Iguchi et al. (2008) compared recovery between two different exertion histories while fatiguing muscles to the same level. Specifically, they utilized quadriceps contractions sustained at 35% and 65% of MVC until force generation capacity decreased to 45% MVC. The authors reported no significant difference in the recovery process between the exertion levels, though recovery was only monitored for 5 minutes. Since, as noted, recovery may require more prolonged durations, additional investigation seems warranted.

In the previous chapter, we investigated the dependency of the fatigue process on cycle time during low to moderate exertion levels of intermittent muscle contraction. A dependency of localized muscle fatigue on cycle time was shown, even though there was a consistent level of overall physical demand. It was concluded that the difference in fatigue development might be related to recovery processes occurring during execution of the intermittent task. In the present study, we have focused on the potential effect of contraction history on post-fatigue recovery. Based on the expected dependency of recovery to the task during exercise, it was hypothesized that post-fatigue recovery will also be affected by the history of exercise-induced muscle fatigue.

## 5.2. Methods

### 5.2.1. Participants

Twelve participants (gender balanced) complete the study, and were recruited from the university and local community. Means (SD) of age, stature, and body mass were 25 (3.3) yrs, 175 (9.6) cm, and 72 (10.2) kg, respectively. All participants reported being at least moderately physically active and having no musculoskeletal disorders or injuries currently or in the preceding 12 months. Prior to any data collection, participants gave informed consent using procedures approved by the Virginia Tech Institutional Review Board.

### 5.2.2. Experimental Design and Procedures

In this study, data from the experiment reported in the prior chapter are used, in which the effects EL and CT on fatigue development was explored. Here, post-exercise (post-fatigue) data were obtained during a one-hour recovery period. In a repeated measures design, participants completed five sessions including an initial 1-hr practice session and four subsequent data collection sessions (~ 3 hrs each). Each session was separated by at least two days, to minimize carryover effects due to residual fatigue. Participants performed intermittent isometric exertions in four conditions, involving all combinations of two cycle times (CT = 30 and 60 sec) and two exertion levels (EL = 15 and 25% of maximum voluntary contraction = MVC). For each, a single duty cycle (DC = 50%) was used, and the order of exposure to the four conditions was counterbalanced across participants using 4 × 4 balanced Latin Squares. The rationale for using these specific task parameters and the specific measures of fatigue is presented in Chapter 4. Note this experimental design yielded two different histories of muscle contraction each at two levels of physical demand (i.e., two ELs). Data collection sessions included 1 hour of intermittent exercise, followed by 1 hour of post-exercise recovery. All participants completed the intermittent contractions, in all four conditions, for the full duration (1 hour). During the recovery period, participant sat in a relaxed position and several measures of muscle capacity were obtained as described below.

All tasks involved index finger abduction, due to the simplicity of movement biomechanics and since the first dorsal interosseous (FDI) muscle is solely responsible for this functional effort. During data collection, participants were seated comfortably in a chair with the examined forearm (dominant arm) resting on a table. Details about the experimental setup and participant hand postures were described in the previous chapter.

LMF was induced by performing index finger abduction in 6 bouts of 10-min intermittent isometric contractions (see Figure 5.1 for an overview). Baseline (pre-fatigue) measures included a minimum of three MVCs (4–5 s duration separated by 1–2 min rest) and low frequency twitches (LFTs). In each MVC trial, participants were asked to maximally activate their FDI muscle in an isometric exertion that involved index finger abduction. The highest value among the completed trials was recorded. Participants were given visual feedback during performance of the intermittent task regarding the level of exertion and the target force. Single MVC trials were completed between each of the 10-min bouts of exercise.

In the 1-hour recovery phase, MVC and LFT measures were obtained shortly after completing the intermittent task, and again at 10, 30, and 60 minutes after task completion. Specifically, the first MVC and LFT measures were begun at 0.2 and 2 minutes after the exercise completion, respectively. One additional MVC was completed 5 minutes after completing the exercise.

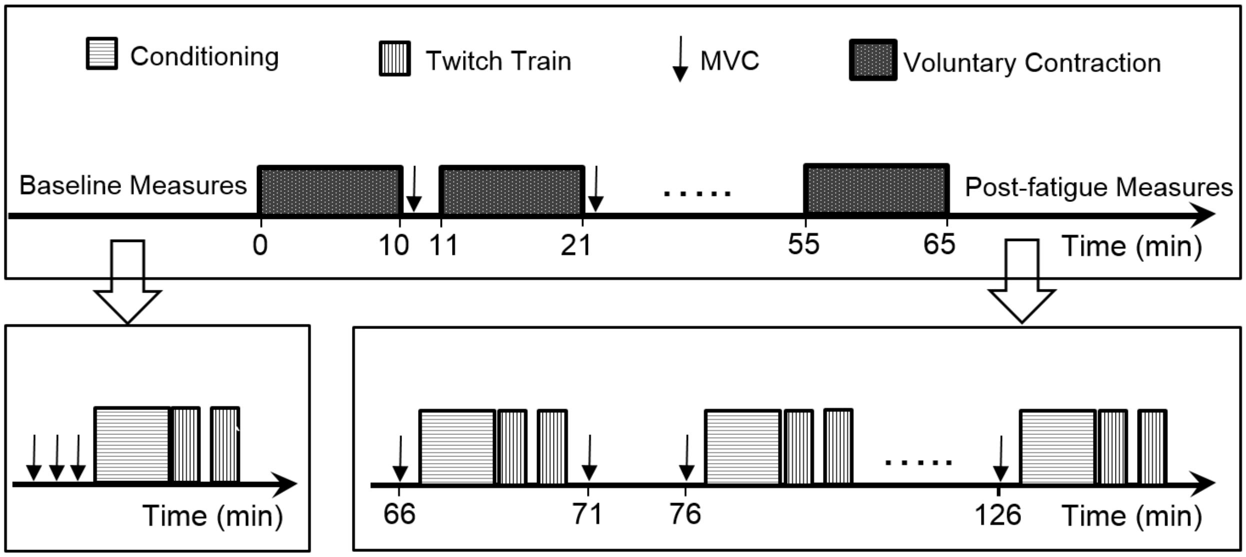



Figure 5.1. Schematic indication of the experimental procedures. The overall procedure is shown at the top, while the lower graphics indicate procedures completed at baseline (left) and after task completion (right). Conditioning involved continuous stimulation at 2 Hz to reach a plateau in twitch force, while twitch trains involved stimulation trains at 2 Hz with rest between trains.

Procedures and instrumentation utilized for electrical stimulation of the muscle belly were described in last chapter. In brief, the FDI muscle was first conditioned with 2 Hz twitches for 2-3 min to reach a stable state of twitch force after potentiation. Subsequently, two 16-twitch trains were applied to the FDI muscle with a 10 sec break between trains. During MVCs and muscle electrical stimulation, exerted forces were sampled at 1000 Hz using a 6 DOF load cell (Nano25-E, ATI Inc., Apex, NC), and low pass filtered at 50 Hz.

### 5.2.3. Dependent measures (DV) and statistical analyses

Objective measures (MVC and LFT responses) were obtained to quantify recovery after LMF. These were obtained at baseline and during the recovery phase as mentioned earlier. To account for substantial individual variability in strength, all MVC and LFT values obtained during the recovery phase were normalized as a percentage of pre-task values. Sample force and LFT data were presented in the last chapter.

Three LFT measures were derived. First, LFT magnitude ($LFT_M$) was calculated from the mean of peak values in the two twitch trains (each included 16 twitches). Both the rise ($LFT_{RISE}$) and relaxation ($LFT_{RELAX}$) rates of twitch forces were respectively determined, as the slopes from linear fits over the rising and lowering phases of individual twitch responses (Blacker et al., 2013; Wilder & Cannon, 2009). The largest slope magnitudes were determined, by obtaining the highest value of the first derivative of the force time series (Blacker et al., 2013). Mean values of LFT responses (magnitude and slopes) were obtained across the two twitch trains completed before and after the intermittent task; DVs were derived as the ratio of these two means (after/before).

Post-fatigue recovery was modeled using separate regressions of normalized DVs (i.e., MVC, $LFT_M$, $LFT_{RISE}$, and $LFT_{RELAX}$) as a function of time (*t)*, specifically representing the DVs as a linear function of log(*t*). While this is equivalent to a nonlinear regression of the DVs vs. time, the earlier approach was for simplicity and to aid identification of "bad" trials (see below). The regression function can be represented as:

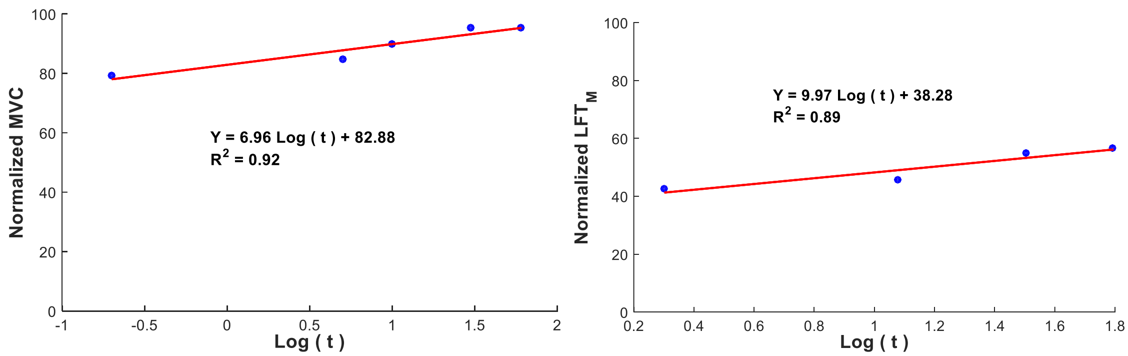
$$DV = A \times \log(t) + B \qquad (5.1)$$

where A and B are the slope and intercept, respectively. As noted, MVCs were collected at $t$ = 0.2, 5, 10, 30, and 60 min after the cessation of exercise, while LFT measures were from twitch trains at $t$ = 2, 12, 32, and 62 min. Matlab 13.0 (The Mathworks, Inc. USA) was used for all regressions, which yielded a pair of A and B parameters for each participant, in each condition, and for each DV. Of note, every DV in every trial was inspected visually. From this, several trials were excluded because an exponential trend in recovery was not evident. Of the total of 192 trials (12 subjects × 4 task conditions × 4 DVs), 4 trials of MVC and 6 trials from each of the LFT measures were excluded, from a total of 10 different participants. For the remaining trials, respective means (SD) of $R^2$ for the regression models of MVC, $LFT_M$, $LFT_{RISE}$, and $LFT_{RELAX}$ were 0.90 (0.11), 0.91 (0.10), 0.90 (0.08), and 0.89 (0.11).

Separate repeated measures analyses of variance (RANOVAs) were used to evaluate the effects of exertion history (i.e., different CTs and ELs) on the rates of recovery (A parameter) for each of the DVs (MVC, $LFT_M$, $LFT_{RISE}$, and $LFT_{RELAX}$). There was substantial within- and between-participant variability in the DV magnitudes at the start of the recovery phase, as expected given differences in the task demands and individual fatigue development. To account for this variability, and to evaluate the dependency of recovery rates on initial conditions, intercepts (A parameters) were included as a continuous covariate in each RANOVA model. Gender and the order of condition presentation were included as blocking variables in these analyses. Partial eta-squared ($\eta_p^2$) was used to assess effect sizes, and was qualitatively interpreted as: small (>0.01), moderate (>0.06), large (>0.14) (Cohen, 1988). All statistical analyses were conducted using JMP Pro 11 (SAS Institute Inc., Cary, NC), and statistical significance was determined when $p < 0.05$. Summary results are presented as means (standard deviations).

## 5.3. Results

Representative samples of obtained data and regression model fits are presented in Figure 5.2, and a summary of recovery rates is provided in Table 5.1.




(a) (b)

Figure 5.2. A sample of linear fit process for (a) normalized MVC, and (b) normalized $LFT_M$.

Table 5.1: Mean (SD) recovery rates (A parameters) for MVC and LFT measures.

| EL (%MVC) | CT (sec) | MVC | $LFT_M$ | $LFT_{RISE}$ | $LFT_{RELAX}$ |
|---|---|---|---|---|---|
| **15** | **30** | 3.05 (1.00) | 7.61 (4.25) | 8.06 (4.55) | 11.30 (6.77) |
| **15** | **60** | 4.43 (1.26) | 7.72 (3.01) | 8.91 (4.36) | 11.22 (5.82) |
| **25** | **30** | 6.39 (2.54) | 7.32 (3.14) | 9.22 (5.46) | 10.90 (5.88) |
| **25** | **60** | 7.15 (1.62) | 8.82 (4.35) | 10.85 (4.50) | 13.60 (6.14) |

A summary of statistical results is provided in Table 5.2. Note that gender did not have a significant effect for any of the measures, and no interaction effects higher than first-order were significant ($p > 0.24$). The effect of Intercept (B parameter in Eq. 5.1) was significant for all DVs ($p < 0.0016$), and in each case the rate of recovery (A parameter) was inversely related to the intercept (Figure 5.3). Of note, one of the participants demonstrated substantially larger LMF, though this data point was not influential on the relationship between recovery rate and intercept for any of the DVs (i.e., did not alter the statistical analysis outcome). The effect of EL on recovery rate of $LFT_M$ approached significance ($p = 0.07$), with mean (SD) values of 7.66 (3.65) for EL = 15 and 8.07 (3.78) for 25% MVC.

Table 5.2. Summary of ANOVA results for the main and interaction effects of Exertion Level (EL), Cycle Time (CT), and Intercept (B parameter in Eq. 5.1) on recovery rates (A parameters in Eq. 5.1) for MVC, and low frequency twitch magnitude ($LFT_M$), LFT rise slope ($LFT_{RISE}$), and LFT relaxation slope ($LFT_{RELAX}$) values. Significant effects are highlighted in bold, and effect sizes ($\eta_p^2$) are interpreted as small (S), moderate (M), or large (L).

| | **MVC** | | **$LFT_M$** | | **$LFT_{RISE}$** | | **$LFT_{RELAX}$** | |
|---|---|---|---|---|---|---|---|---|
| | ***p* value** | **$\eta_p^2$** | ***p* value** | **$\eta_p^2$** | ***p* value** | **$\eta_p^2$** | ***p* value** | **$\eta_p^2$** |
| **EL** | 0.21 | 0.074 (M) | 0.07 | 0.046 (S) | 0.19 | 0.040 (S) | 0.52 | 0.024 (S) |
| **CT** | 0.14 | 0.014 (S) | 0.83 | 0.004 (S) | 0.58 | 0.007 (S) | 0.46 | 0.010 (S) |
| **B** | **0.0008** | 0.274 (L) | **0.0016** | 0.219 (L) | **<0.0001** | 0.437 (L) | **<0.0001** | 0.465 (L) |
| **EL×CT** | 0.38 | 0.045 (S) | 0.28 | 0.014 (S) | 0.65 | 0.002 (S) | 0.45 | 0.015 (S) |
| **EL×B** | 0.17 | 0.011 (S) | 0.25 | 0.015 (S) | 0.91 | 0.004 (S) | 0.61 | 0.006 (S) |
| **CT×B** | 0.92 | 0.008 (S) | 0.54 | 0.013 (S) | 0.93 | 0.012 (S) | 0.44 | 0.002 (S) |

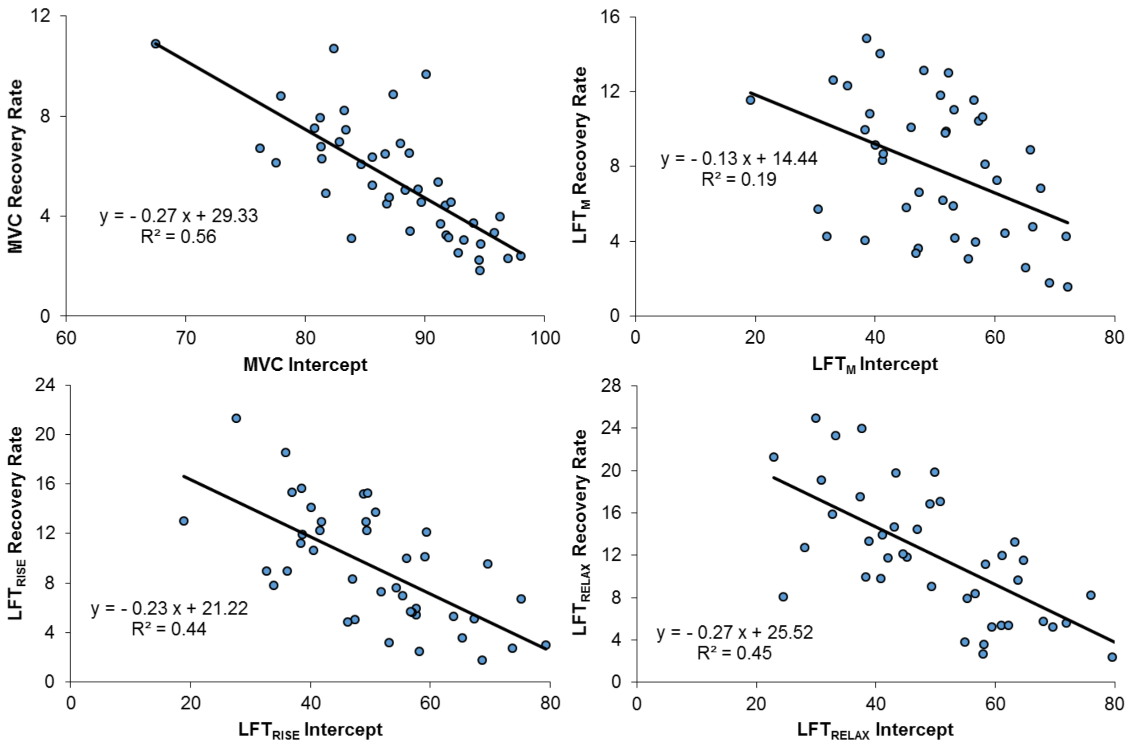


Figure 5.3. Relationships between rates of recovery (A parameter) and initial conditions (B parameter) for: MVC (top left), $LFT_M$ (top right), $LFT_{RISE}$ (bottom left), and $LFT_{RELAX}$ (bottom right). Regression lines and function are presented in each graph.

## 5.4. Discussion

The purpose of this study was to investigate the effect of contraction history on muscle recovery after fatiguing exertions. One approach to this would be imposing a consistent level of muscle fatigue using different exertion histories (or patterns of exertion). An alternative approach was used here, for two main reasons. First, tracking the level of muscle fatigue over time is challenging, and would require frequent testing (e.g., of MVC and LFT responses) that could confound the actual effects of the exercises themselves. Also, there is inherent variability in the responses to such tests, making it difficult to achieve the desired target fatigue level with precision. Second, interest here was in the effects of contraction history over consistent exercise durations with similar overall physical demands, which was considered more practically applicable (e.g., to the design of occupational task).

The results provide consistent evidence that post-exercise recovery rate depends on initial conditions, specifically the level of fatigue-induced reduction in muscle contractile status (Table 5.2). This relationship was fairly linear for recovery of both MVC and LFT measures (Figure 5.3), with a higher rate of recovery (on a logarithmic time scale) occurring when the initial state involved a lower contractile level. From the previous chapter, it was shown that a different history of muscle contraction resulted in different levels of muscle fatigue, even for similar levels of overall physical demand. There was substantial within- and between-participant variability in the DV magnitudes at the start of the recovery phase, as expected given differences in the task demands (using two exertion levels) and individual fatigue development. To account for this variability, and to assess the dependency of recovery rates on initial conditions, post-fatigue capacity of muscle (initial condition) was included as a covariate in our analysis. This factor turned out to be the only significant parameter influencing the rate of recovery (Table 5.2).
To our knowledge, there is only one prior reported study that assessed the history dependency of muscle recovery, and which involving fatiguing a muscle to a similar level of fatigue under different loading conditions (Iguchi et al., 2008). In this study, the quadriceps muscle was fatigued to 45% MVC, using sustained contractions at 35% and 65% of MVC. These authors reported no significant difference in the recovery process between the two contraction histories. Here, four different history of contraction were

utilized, focusing on low to moderate exertion levels (i.e., 15 and 25% MVC), and similarly no influence of contraction history was evident. Of note, though, the noted study only assessed the recovery process over a short period of time (5 min), while recovery here was evaluated over a 1-hour period.

Based on results of the previous chapter, it was concluded that differences in post-fatigue muscle capacity between tasks involving different cycle times are likely related to the recovery process during exertions and intermittent breaks, particularly since the overall physical demand was similar between the different cycle times (at a specific exertion level). However, the overall effect of fatigue and recovery (i.e., during both exercise and intermittent rests) was examined in that study. Considering that there was a similar physical demand between cycle times for each EL, the differences in LMF could be attributed to the overall recovery process, though it was not possible to associate the post-exercise fatigue level to either of the recovery types. Here, examination of the effect of different contraction histories on post-fatigue recovery indicated no difference in the recovery process. Therefore, it seems reasonable to associate the difference in fatigue development to differences in the recovery processes during exercise. Particularly, recovery processes during exercise could be considerable, since blood flow occlusion was likely not substantial due to the incorporation of low to moderate levels of exertion.

Specific mechanisms influencing recovery during an exercise are not clear. In fact, there are two mechanisms that can work in opposition to influence blood flow during exercise, thereby affecting the recovery rate. As mentioned earlier, blood flow is proportionally coupled to muscle metabolic rate (Murrant & Sarelius, 2015). Higher levels of metabolic rate may occur with larger physical demand, such as a higher contraction level (Anrep & von Saalfeld, 1935; Hamann et al., 2004), leading to higher levels of blood flow and eventually faster recovery. A second mechanism that may affect the blood flow is intramuscular pressure. Higher levels of intramuscular pressure, which can be a result of higher level of muscle contraction, leads to locally reduced levels of blood flow (Järvholm et al., 1988; Sejersted et al., 1984). As such, while there is a tendency to increase the blood flow at higher demands, higher intramuscular pressure acts against this mechanism.

Results of the current study have implications in terms of modeling the recovery process, particularly during muscle rest (Figure 5.3). Based on the current results, post-fatigue rates of recovery appear to depend primarily on the initial muscle status. This is conceptually similar to the assumption made in the recovery model of Ma et al. (2010), which defined the rate of recovery to be proportional to the difference between maximum and current muscle capacity. While any changes in metabolic rate and intramuscular pressure are unlikely during complete rest, it is likely that at a given time recovery should depend on the current muscle capacity, yielding the well-known exponential recovery process during rest. A question still remains, though, as to why the fatigue and recovery model (Ma et al., 2010) demonstrated unsatisfactory predictions for intermittent isometric contractions (see Chapter 3). One reason might be related to the fact that this model separates the fatigue and recovery processes, and does not consider any recovery occurring during an exercise. The model of Xia and Frey Law (2008) also accounts for the proportionality between the rate of recovery and current force muscle capacity, but does this both during and after exercise. However, this modeling employs one constant proportionality factor for all exercise conditions. This model could potentially be enhanced, therefore, by employing a variable proportionality factor, for example one that was larger during complete rest (more blood flow) and lower with increasing levels of muscle contraction (less blood flow).

A few limitations in the present work should be noted. Only relatively young and healthy participants were included, and as such future work is needed to determine if the current results are consistent more generally. Only a narrow range of task parameters was included here, so any conclusions about the task-dependency of recovery must necessarily be somewhat limited. Subsequent work is again needed to assess the influence of exertion history over a larger range of task conditions, including both shorter and longer CTs, varying duty cycles, higher and lower ELs, different muscles, and more complex force-time patterns. Of note, electromyographic (EMG) measures of the FDI were not feasible here using surface electrodes, since twitches were induced using stimulation of the FDI muscle belly (to avoid any antagonistic stimulation). EMG use in future work may also help in assessing the task-dependency of muscle recovery. Finally, high-frequency stimulation, commonly used to assess peak contractile capacity, was not used, since preliminary testing indicated that it was not tolerable by many participants. As such, our current measures of contractile capacity (i.e., MVCs and LFT) may have not been fully representative, and we were unable to assess the contribution of central vs. peripheral recovery.

The present study examined the dependency of muscle recovery on four different history of fatiguing muscle contractions, imposed using two cycle times during low to moderate levels of intermittent static exertions. MVC and LFT measures were used, all of which indicated a dependency of muscle recovery on current muscle capacity existing after fatiguing exercise. Modifying influences of the task parameters (cycle time and exertion level) appeared small. Overall, these results imply that the post-exercise rate of recovery is primarily influenced by the post-exercise muscle state. Or, simply, how the muscle recovers depends on the state from which it starts to recovery but not on how it got to that state. The current results may also help in enhancing existing models of muscle recovery, facilitating more accurate predictions of LMF development and thereby help to enhance muscle performance and reduce the risk of injury.

## References


Andersen, P, & Saltin, B. Maximal perfusion of skeletal muscle in man. *The Journal of Physiology* 1985;366(1):233-249.

Anrep, GV, & von Saalfeld, E. The blood flow through the skeletal muscle in relation to its contraction. *The Journal of Physiology* 1935;85(3):375-399.

Baker, AJ, Kostov, KG, Miller, RG, & Weiner, MW. Slow force recovery after long-duration exercise: metabolic and activation factors in muscle fatigue. *Journal of Applied Physiology* 1993;74(5):2294-2300.

Blacker, S, Fallowfield, J, & Willems, M. Intra- and interday reliability of voluntary and electrically stimulated isometric contractions of the quadriceps femoris. *Journal of Electromyography and Kinesiology* 2013;23(4):886-891.

Bockman, EL. Blood flow and oxygen consumption in active soleus and gracilis muscles in cats. *American Journal of Physiology - Heart and Circulatory Physiology* 1983;244(4):H546-H551.

Cohen, J. *Statistical power analysis for the behavioral sciences*: Lawrence Erlbaum Associates, Inc, 1988.

Edwards, R, Hill, D, Jones, D, & Merton, P. Fatigue of long duration in human skeletal muscle after exercise. *J Physiology* 1977;272:769-778.

Elfving, B, Liljequist, D, Dedering, Å, & Németh, G. Recovery of electromyograph median frequency after lumbar muscle fatigue analysed using an exponential time dependence model. *European Journal of Applied Physiology* 2002;88(1):85-93.

Enoka, RM, & Duchateau, J. Muscle fatigue: what, why and how it influences muscle function. *J Physiol* 2008;586(1):11-23.

Fuglevand, AJ, Zackowski, KM, Huey, KA, & Enoka, RM. Impairment of neuromuscular propagation during human fatiguing contractions at submaximal forces. *J Physiology* 1993;460:549-572.

Hamann, JJ, Buckwalter, JB, Clifford, PS, & Shoemaker, JK. Is the blood flow response to a single contraction determined by work performed? *Journal of Applied Physiology* 2004;96(6):2146-2152.

Iguchi, M, Baldwin, K, Boeyink, C, Engle, C, Kehoe, M, Ganju, A, Messaros, AJ, & Shields, RK. Low frequency fatigue in human quadriceps is fatigue dependent and not task dependent. *J Electromyogr Kinesiol* 2008;18(2):308-316.

Iridiastadi, H, & Nussbaum, MA. Muscle fatigue and endurance during repetitive intermittent static efforts: development of prediction models. *Ergonomics* 2006;49(4):344-360.

Järvholm, U, Styf, J, Suurkula, M, & Herberts, P. Intramuscular pressure and muscle blood flow in supraspinatus. *European Journal of Applied Physiology* 1988;58:219-224.

Jones, DA. High- and low-frequency fatigue revisited. *Acta Physiologica Scandinavica* 1996;156(3):265-270.

Kroon, GW, & Naeije, M. Recovery of the human biceps electromyogram after heavy eccentric, concentric or isometric exercise. *European Journal of Applied Physiology and Occupational Physiology* 1991;63(6):444-448.

Lännergren, J, & Westerblad, H. Force decline due to fatigue and intracellular acidification in isolated fibres from mouse skeletal muscle. *The Journal of Physiology* 1991;434(1):307-322.

Ma, L, Chablat, D, Bennis, F, Zhang, W, & Guillaume, F. A new muscle fatigue and recovery model and its ergonomics application in human simulation. *Arxiv preprint arXiv:1010.5891* 2010.

Murrant, CL, & Sarelius, IH. Local control of blood flow during active hyperaemia: what kinds of integration are important? *The Journal of Physiology* 2015;593(21):4699-4711.

Reid, MB, Grubwieser, GJ, Stokic, DS, Koch, SM, & Leis, AA. Development and reversal of fatigue in human tibialis anterior. *Muscle & Nerve* 1993;16:1239-1245.

Sahlin, K, & Ren, J. Relationship of contraction capacity to metabolic changes during recovery from a fatiguing contraction. *J Appl Physiol* 1989;67(2):648-654.

Sejersted, OM, Hargens, AR, Kardel, KR, Blom, P, Jensen, Ø, & Hermansen, L. Intramuscular fluid pressure during isometric contraction of human skeletal muscle. *Journal of Applied Physiology* 1984;56(2):287-295.

Westgaard, RH, & Winkel, J. Guidelines for occupational musculoskeletal load as a basis for intervention: a critical review. *Applied Ergonomics* 1996;27(2):79-88.

Wilder, M, & Cannon, J. Effect of age on muscle activation and twitch properties during static and dynamic actions. *Muscle Nerve* 2009;39(5):683-691.

Wood, DD, Fisher, DL, & Andres, RO. Minimizing fatigue during repetitive jobs: optimal work-rest schedules. *Human Factors* 1997;39(1):83-101.
Xia, T, & Frey Law, LA. A theoretical approach for modeling peripheral muscle fatigue and recovery. *Journal of Biomechanics* 2008;41(14):3046-3052.

## 6. Conclusions and future directions

Localized muscle fatigue (LMF) – a complex, multifactorial phenomenon – has been defined as “any exercise-induced reduction in the ability of a muscle to generate force or power, regardless of the ability to sustain the task” (Bigland-Ritchie & Woods, 1984). LMF is widely used as an indicator of underlying physiological processes, since such fatigue leads to a decline in desired performance and muscle force capacity during diverse activities involving voluntary muscle force generation (Vøllestad, 1997). LMF leads to subjective and objective changes, such as an increased perceived exertion and discomfort, reduced strength, diminished neuromuscular control, muscle tremors, altered electromyographic (EMG) signals, and may increase the risk of WMSDs.

### 6.1. Review of localized muscle fatigue models

LMF is a complex phenomenon that can differ between individuals, tasks, and muscles. Several muscle-fatigue models (MFMs) have been developed in prior research. These models have potential practical value in ergonomics, given that LMF can impair performance, serve as a surrogate measure of injury risk, and may act as a causal factor for WMSDs. Based on the review of existing MFMs, two specific MFMs were identified as most ergonomically relevant. These models were directly compared and some important differences in predictions were found. Identifying such differences was suggested as a useful approach, both for developing testable hypotheses and in guiding subsequent model development or refinement. Other potential methods for improving future MFMs were also suggested, including expansion of the model structure using factors related to individual differences and task-related parameters, and utilizing different types of MUs in the model structure. Predicting LMF using a MFM, while accounting for effects of important individual differences, can contribute to future design/evaluation of work tasks, with the end goal of controlling the development of LMF and associated adverse consequences on performance and injury risk.

## 6.2. Sensitivity analysis and assessment of two muscle fatigue models

MFMs have broad potential application if they can accurately predict muscle capacity and/or endurance time (ET) during the execution of diverse tasks. As an initial step toward facilitating improved MFMs, two MFMs were assessed here in terms of their sensitivity to inherent model parameters and their ability to predict ET in both prolonged and intermittent exertions. The two MFMs were, in general, more sensitive to the alterations of the parameter that represents the rate of muscle fatigue. Both models demonstrated a higher sensitivity to their fatigue and recovery parameters in conditions involving lower to moderate levels of effort, though such conditions maybe those that are of most practical interest in the occupational domain. The ability of these models to predict ET was inferior for mixed loading conditions (a combination of prolonged and intermittent contractions). When optimizing model parameters for different loading conditions, the recovery parameter showed considerably larger variability, which might be related to the inability of these MFMs in simulating the recovery process under different loading conditions. For future application, improved model predictions of fatigue and recovery are needed, especially across diverse loading conditions, and a specific focus on an improved representation of the recovery processes is recommended.

## 6.3. Effect of cycle time on development of localized muscle fatigue

LMF during a repetitive task can be influenced by several aspects, such as the level and duration of exertions. Among these aspects, though, the influence of cycle time (CT) remains unclear. The primary goal of this study was to investigate the effect CT for a simple biomechanical system, and specifically to assess the development and consequences of LMF during intermittent isometric exertions. Diverse measures were used, all of which demonstrated a beneficial effect of a shorter CT. Specifically, the shorter CT led to changes in the rates of change in perceived discomfort, performance, and muscle capacity, as well as task-related changes in contractile status. These differences indicate less fatigue development and fatigue-induced consequences with the shorter CT, and were quite consistent between genders and the two exertion levels (EL) investigated. Current rest-allowance prediction models do not consider the effect of CT, and the present results suggest that these may thus be oversimplified for some applications. The current results may also help in enhancing existing models used to simulate or predict muscle fatigue. Accounting for the effects of CT may help in reducing LMF development and thereby help to enhance performance and reduce the risk of the risk of adverse musculoskeletal outcomes.

## 6.4. History-dependency of muscle recovery

Muscle fatigue and recovery are complex processes influencing muscle force generation capacity. While fatigue reduce this capacity, muscle recovery act to restore the unfatigued muscle state. Many factors can potentially affect muscle recovery, and among these may be a task dependency of recovery following an exercise. However, little has been reported regarding the history dependency of recovery after fatiguing contractions. The present study examined the dependency of muscle recovery on four different history of fatiguing muscle contractions, imposed using two CTs during low to moderate levels of intermittent static exertions. Maximum voluntary contractions (MVCs) and low frequency twitch (LFT) measures were used, all of which indicated a dependency of muscle recovery on current muscle capacity existing after fatiguing exercise. Modifying influences of the task parameters (CT and EL) appeared small. Overall, these results imply that the post-exercise rate of recovery is primarily influenced by the post-exercise muscle state. Or, simply, how the muscle recovers depends on the state from which it starts to recovery but not on how it got to that state. The current results may also help in enhancing existing models of muscle recovery, facilitating more accurate predictions of LMF development and thereby help to enhance muscle performance and reduce the risk of injury.

## 6.5. Limitations and future directions

There are several limitations of this work that should be addressed in the future research. Two MFMs with practical utility for application in occupational settings were assessed here, in the context of prolonged and intermittent contractions. As this work examined only static exertions, future studies would benefit from incorporating dynamic contractions, which are more complex due to inherently larger alterations in MU recruitment and blood flow. With a longer-term goal of supporting the developing of MFMs applicable to dynamic conditions, experimental work needs to be completed to determine the effects of motion on fatigue and recovery under different loading conditions with comparable demands.  Another limitation was the focus on only one muscle group (i.e., hand/grip muscles).  Subsequent work should assess MFM performance for other muscle groups, since fatigue and recovery processes are not only task dependent, but also dependent on the muscle group involved (e.g., related to differences in fiber type distribution).  Future work evaluating and comparing MFMs should ideally use more comparable data sets.  Similarly, MFMs should be evaluated to assess their potential to account for important inter-individual differences, such as related to gender and aging.

A few limitations regarding to the effect of CT on LMF and recovery should also be noted.  Only relatively young and healthy participants were included, and as such future work is needed to determine if the current results are consistent more generally.  Only a narrow range of CTs (and ELs) was included here, so any conclusions about the effects of CT must necessarily be somewhat limited.  Subsequent work is again needed to evaluate CT effects over a larger range of task conditions, including both shorter and longer CTs, varying duty cycles, higher and lower ELs, different muscles, and more complex force-time patterns.  Of note, EMG measures of the FDI were not feasible here using surface electrodes, since twitches were induced using stimulation of the FDI muscle belly (to avoid any antagonistic stimulation).  EMG use in future work may also help in assessing the task-dependency of muscle recovery.  Finally, high-frequency stimulation, commonly used to assess peak contractile capacity, was not used, since preliminary testing indicated that it was not tolerable by many participants.  As such, our current measures of contractile capacity (i.e., MVCs and LFT) may have not been fully representative, and we were unable to assess the contribution of central vs. peripheral recovery.

### 6.6. Summary

Overall, this dissertation contributes to theoretical and practical aspects of human muscle fatigue and recovery. Regarding the theoretical aspects, two specific MFMs were directly compared and some important differences in their predictions were reported. These differences were used as a base for developing testable hypotheses and in designing the experiments of this dissertation study. Further theoretical evaluations were conducted to explore the sensitivity of these models to the model parameters and their ability to predict ET in both prolonged and intermittent exertions. Higher sensitivity was observed for both model parameters and in conditions involving lower to moderate levels of effort. When optimizing model parameters for different loading conditions, variability was substantially larger for the recovery parameter, which might be related to the inability of these MFMs in simulating the recovery process under different loading conditions.

From a practical viewpoint, the effect of CT was explored on development and consequences of LMF during intermittent isometric exertions. Shorter CT led to beneficial changes in the rates of change in perceived discomfort, performance, and muscle capacity, as well as task-related changes in contractile status, all of which indicated less fatigue development and fatigue-induced consequences. Lastly, the dependency of muscle recovery on these different histories of fatiguing muscle contractions was explored. Overall, results imply that the how the muscle recovers depends on the state from which it starts to recover, though not the exertion history that led to that state.

In summary, results of these studies may help in enhancing our understanding of fatigue and recovery processes, and in improving existing models of muscle fatigue and recovery. More accurate predictions of LMF development and thereby enhancing muscle performance may reduce the risk of musculoskeletal injuries and their associated healthcare costs.

## References

Bigland-Ritchie, B, & Woods, JJ. Changes in muscle contractile properties and neural control during human muscular fatigue. *Muscle & Nerve* 1984;7(9):691-699.

Vøllestad, N. Measurement of human muscle fatigue. *Journal of Neuroscience Methods* 1997;74(2):219-227.

# Appendix A: Informed Consent Form

VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY

Informed Consent for Participants
in Research Projects Involving Human Subjects

<u>Title of Project:</u> Localized Muscle Fatigue: Theoretical and Practical Aspects in Occupational Environments
<u>Investigators:</u> Ehsan Rashedi and Dr. Maury Nussbaum

## I. Purpose

The purpose of this project is to measure the development of muscle fatigue in response to different types of physical demands. These measurements will be used to determine how different types of such physical demands influence fatigue, and to evaluate existing models of muscle fatigue.

## II. Procedures

It is important for you to understand that we are not evaluating you or your performance in any way. You are helping us to collect data that will be used to estimate the muscle fatigue in response to different types of physical demands, and in the future to better predict muscle fatigue. Any tasks you perform, or opinions you have will only help us do a better job in this evaluation and prediction. Therefore, we will ask that you perform tasks following several instructions and to the best of your abilities. The information and feedback that you provide is very important to this project. You will be asked to complete exercises in up to 7 sessions, and the total experiment time will be approximately 4 hours in each session.

During the course of this experiment you will be asked to perform the following tasks:

1) Read and sign an Informed Consent Form (this form).
2) Complete paperwork to allow for compensation.
3) Allow experimenters to measures your height, body weight, and several dimensions from your hands and/or arms.
4) Be instructed in the experimental tasks and techniques.
5) Allow experimenters to place electrodes over muscles in your arm and bands to limit the movement of your arm. A same gender or opposite gender (with your permission) experimenter will be applying the electrodes. These electrodes will be used to measure the electrical activity generated by your muscles when they contract (EMG, or electromyography). Such measures are completely "passive", and do not involve applying any electricity to your body. Electrodes will also be placed that will be used to electrically stimulate your muscles, causing them to contract. This stimulation is done in a controlled way, and there is no risk of harming you. It is somewhat uncomfortable, though, especially at the beginning. We will demonstrate this process to you shortly, so you know what it involves. In fact, we will demonstrate it on ourselves first, so you can see that it is a safe procedure.

6) Perform several physical tasks that involve generating forceful efforts with different muscles in your hand and/or arm. You will be asked to do these at different levels of effort and with different patterns of effort vs. rest. Before, during, and after these tasks, we will obtain several measures (EMG, force, and your level of perceived effort and discomfort). We will also be stimulating your muscles occasionally, as will be demonstrated.

## III. Risks and Benefits

We believe that there are minimal risks to you as a participant in this study, as follows.

1) You may experience minor muscle strain resulting from the experimental tasks.
2) You may experience delayed onset muscle soreness, in the 24-48 hours following the experiment.
3) You may experience minor muscle or skin discomfort from the muscle electrical stimulation, and also may experience temporary skin irritation due to the electrodes.

Participants in a study are considered volunteers, regardless of whether they receive payment for their participation. Under Commonwealth of Virginia law, workers compensation does not apply to volunteers. Thus, appropriate health insurance is strongly recommended to cover these types of expenses.

The results of this study will enhance our knowledge regarding the development of muscle fatigue, and particularly the ability to predict muscle fatigue under a range of physical demand conditions. A successful model will contribute to future design/evaluation of work tasks, particularly in a proactive fashion. More effective proactive design may, in turn, reduce the development of muscle fatigue and associated decreases in performance and increases in the risk of work-related musculoskeletal disorders. This work can have additional applications, such as in designing training or rehabilitative procedures. While this research may yield such benefits, no promise or guarantee of benefits is made to you as a participant. You may contact the investigators listed at the end of this form to inquire about the results and conclusions of this research.

## IV. Extent of Anonymity and Confidentiality

Your personal information and identity will be kept in the strictest of confidence. The list associating names with answers will be destroyed one month after completion of data collection. Photographing might occur for assisting in publications. However, any images used in documentation will have your face blacked out to maintain confidentiality. All information will be collected in a file and locked when not being used, and only the investigators have access to the data. It is possible that the Institutional Review Board (IRB) may view this study’s collected data for auditing purposes. The IRB is responsible for the oversight of the protection of human subjects involved in research.

## V. Compensation

You will be compensated for your participation at a rate of $10 per hour for the experimental time completed, plus an additional $10 bonus for completing all experimental sessions. You will be paid at the end of each session in cash.

Freedom to Withdraw

You are free to withdraw from this study at any time without penalty or reason stated, and no penalty or withholding of compensation will occur for doing so. If you choose to withdraw, you will be compensated for the portion of study time of the study completed. Furthermore, you are free not to answer any question or respond to experimental situations without penalty. There may be circumstances under which the investigator may determine that the experiment should not be continued. In this case, you will be compensated for the portion of the project completed.

## VI. Approval of Research

This study has been approved by the Institutional Review Board (IRB) for Research Involving Human Participants at Virginia Tech.

## VII. Participant's Permission

I have read and understand the Informed Consent and conditions of this project. I have had all my questions answered. I hereby acknowledge the above and give my voluntary consent:

________________________________________________________

Participant Signature Date

________________________________________________________

Experimenter Signature Date

**Signature Page**

I have read the description of this study and understand the nature of the research and my rights as a participant. I hereby consent to participate with the understanding that I may discontinue participation at any time if I choose to do so.

__________________________________________________________

Participant's Signature                                             Date

__________________________________________________________

Printed Name

The research team for this experiment is led by Dr. Maury Nussbaum. He may be contacted at the following address and phone number:

Dr. Maury A. Nussbaum
Professor
Department of Industrial and Systems Engineering
521 Whittemore Hall (0118)
Blacksburg, VA 24061
Phone: (540) 231-6053
Email: nussbaum@vt.edu

In addition, if you have any detailed questions regarding your rights as participant in University Research, you may contact the following individual:

Dr. David Moore
Chair, Virginia Tech Institutional Review Board
for the Protection of Human Subjects
Office of Research Compliance
Phone: (540) 231-4991
Email: moored@vt.edu