A review of occupationally-relevant models of localised muscle fatigue

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Abstract: Localised 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 occupationally-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: localised muscle fatigue; LMF; models; performance; ergonomics; recovery; endurance; review.


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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 localised muscle fatigue (LMF), summarise 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 and 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 characterised 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 and Komi, 1986), increased muscle tremors (Hunter and Enoka, 2003), and decrements in proprioceptive ability (Björklund et al., 2000) and sense of effort (Enoka and Stuart, 1992).

WMSDs are prevalent in diverse occupational environments, accounting for about 33% of all non-fatal 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 recognised 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 and Winkel, 2003).
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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 (Sjogaard and 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 Mechanisms of LMF

Mammalian muscle is comprised of muscle fibres 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 fibres. Muscle fibres 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 fibres (Gandevia et al., 1995). Each of these processes can act as a limiting factor and contributor to fatigue (Vøllestad, 1997). These processes are categorised 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.

3 Overview of muscle fatigue modelling

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 summarised below.

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 (Yassierli et al., 2007; Horton et al., 2012; Iridiastadi and Nussbaum, 2006b). 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 and Jonsson, 1977; Iridiastadi and Nussbaum, 2006b). Underlying these effects, greater effort levels/durations result in higher/longer demands on each recruited muscle fibre and more recruitment of MUs, both leading to a faster rate of muscular fatigue.

Quantitatively, a curvilinear relationship has been long recognised 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 and Jonsson, 1977). Cycle time may also impact the development of LMF, but apparently with less substantial effects compared with effort level (Yassierli and Nussbaum, 2009; Iridiastadi and Nussbaum, 2006b). 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 unfavourable (Sommerich et al., 1993). In contrast, one study indicated that a shorter cycle time yields slower development of LMF (Yassierli and Nussbaum, 2009).

3.2 Muscle fatigue modelling 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 MFM to predict muscle fatigue has broad potential application. These models can be categorised into two types, empirical and theoretical, and existing work on each is summarised below.

3.2.1 Empirical MFM

Empirical MFM are based on empirical observations and fitting experimental data, rather than mathematical relationships between underlying system parameters. The origin of this type of MFM 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 generalising 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 and Nussbaum, 2006b; 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 (2006a) 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).

3.2.2 Theoretical MFMs

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 utilised several approaches for predicting declines in muscle force during fatiguing tasks (summarised in Table 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 Ca2+ 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.

### Table 1 Summary of theoretical MFMs

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

Notes: 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.

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

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 fibre level, and have calculated overall muscle force as the sum of individual fibre forces (Deeb et al., 1992; Hawkins and Hull, 1993). Parameters in Deeb et al. (1992) model were obtained using curve fitting to experimental data, however this model was not experimentally validated for other conditions. Regarding to 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 Law (2008) exploited the same MU-based approach in their model, in which they sought to optimise computational effort using a simplified approach in comparison to other models that utilise explicit muscle length-tension and force-velocity relationships. They characterised 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 and 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 (MVC) 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 modelling (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 fatiguability 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.

4 A direct comparison of two theoretical MFMs

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 and 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 modelling 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 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).

4.1 Loading conditions and model outputs

Two distinct isometric loading conditions were examined, including both isotonic and non-isotonic efforts, and with ET 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 Ma et al. (2009) and Xia and Law (2008) models are presented below, for the selected loading conditions (all involving the hand-grip muscles).

4.1.1 Sustained isometric exertions and ET

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 1). Both models predicted similar ETs for each level of exertion (Figure 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 1  Comparison of results from two theoretical (Ma et al., 2009; Xia et al., 2008) and one empirical model (Manenica 1986), for endurance time during sustained isometric exertions involving hand grip

4.1.2 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): ‘BBAABBA’. Of note, all three sequences had the same cumulative physical demands (Figure 2). Simulation results for both models regarding muscle force capacity over time are depicted in Figures 3 and 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 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** 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)

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 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 (Figures 3 and 4). Further, 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.

**Figure 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

![Figure 3](image1)

Note: Task sequences are shown in Figure 2.

**Figure 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

![Figure 4](image2)

Note: Task sequences are shown in Figure 2.
5 Discussion and conclusions

A variety of approaches have been reported for modelling human muscle fatigue, and which were categorised 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 generalisability), 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 modelling 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 modelling 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., ETs 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 modelling 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 modelling 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 utilising 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 normalised effort level (Allman and 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 and Law, 2011).

Loss of muscle fibres with age, especially type II (fast twitch) fibres, likely contributes substantially to age-related differences in LMF development (Lexell et al., 1983). Alterations in muscle fibre composition may also account for age-related differences between different types of muscle contractions (i.e., isometric vs. dynamic contractions), since fast-twitch muscle fibres 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 and 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 (Hunter, 2009; Avin et al., 2010), females typically have a lower proportion of type II muscle fibres (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 which can result in increased fatigue resistance (Hicks et al., 2001). Notably, some of these gender differences (i.e., in muscle mass and fibres 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 fibre 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 (Maffioletti et al., 2007), whereas another recent study found no significance difference (Cavuoto and 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 fibres in obese individuals (Tanner et al., 2002). Incorporating different muscle fibres in the process of developing MFMs could therefore account for differences in LMF related to obesity.

Existing evidence (briefly summarised above) indicates important differences in fatigue between individuals, as well as different muscle groups within an individual. Alterations and/or differences in muscle fibre 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 fibres in some detail (Hawkins and Hull, 1993), yet is considered not occupationally-relevant given the
resultant complexity in model structure. However, MFM s 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 Law (2008). Doing so would implicitly account for differences between muscle fibre types, since:

1 muscle fibres innervated by an α-motor neuron (within a MU) appear to have identical characteristics

2 each muscle fibre is innervated by only one MU (Burke et al., 1973).

Thus, instead of using different types of muscle fibres in a MFM, using different type of MUs could be simpler (i.e., modelling 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 modelling 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; Keenan and Valero-Cuevas, 2007; Dideriksen et al., 2010). 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 and 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 previous studies have shown that EMG can help in determining muscle fibre type composition. In particular, EMG can classify muscle fibres into distinct categories of ‘fast’ and ‘slow’ fibres (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 fibres 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 and Avin, 2010) to identify model parameter values for use in a MFM (Xia and 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 modelling/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 modelled. Use of MFMs including a recovery model, such as in Ma et al. (2009) and Xia and 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, 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). Utilising 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 and Green, 2012; Marion et al., 2010; Sih et al., 2012; Ma 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 utilising 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


A review of occupationally-relevant models of localised muscle fatigue


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**List of acronyms**

<table>
<thead>
<tr>
<th>Acronym</th>
<th>Definition</th>
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<tbody>
<tr>
<td>CNS</td>
<td>Central nervous system</td>
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<tr>
<td>EMG</td>
<td>Electromyography</td>
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<td>ET</td>
<td>Endurance time</td>
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<td>LMF</td>
<td>Localised muscle fatigue</td>
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<td>MEF</td>
<td>Maximum evocable force</td>
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<td>MFM</td>
<td>Muscle fatigue models</td>
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<td>MSD</td>
<td>Musculoskeletal disorder</td>
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<td>MU</td>
<td>Motor unit</td>
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<tr>
<td>MVC</td>
<td>Maximum voluntary contraction</td>
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<tr>
<td>WMSD</td>
<td>Work-related musculoskeletal disorders</td>
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