 2022 Aug 11;16:893275. doi: 10.3389/fnsys.2022.893275Exercise fatigue is a common physiological phenomenon in human activities. The occurrence of exercise fatigue can reduce human power output and exercise performance, and increased the risk of sports injuries. As physiological signals that are closely related to human activities, surface electromyography (sEMG) signals have been widely used in exercise fatigue assessment. Great advances have been made in the measurement and interpretation of electromyographic signals recorded on surfaces. It is a practical way to assess exercise fatigue with the use of electromyographic features. With the development of machine learning, the application of sEMG signals in human evaluation has been developed. In this article, we focused on sEMG signal processing, feature extraction, and classification in exercise fatigue. sEMG based multisource information fusion for exercise fatigue was also introduced. Finally, the development trend of exercise fatigue detection is prospected.Keywords: exercise fatigue, sEMG, machine learning, feature extraction, classificationExercise fatigue is a physiological phenomenon that originates from human activities and results in decreased physical performance (Nijs et al., 2011). The main manifestation of exercise fatigue is muscle fatigue which was defined as the failure to maintain the force output, leading to a reduced performance (Asmussen, 1993). Muscle strength declines progressively during exercise, so fatigue occurs before the task failure (Gandevia, 2001). There are many examples of application fields involving muscle fatigue, including sports (Simsek, 2017; Kuniszyk-Jozkowiak et al., 2018; Liu et al., 2019; Hedayatpour et al., 2021; Liu and Li, 2021; Rodriguez-Rosell et al., 2021), rehabilitation (Gaudet et al., 2018; Kim et al., 2018; Meng et al., 2019; Na et al., 2020; Fundaro et al., 2021; Sato et al., 2021), and occupation (Alberto et al., 2018; Fang et al., 2021; Ji and Huang, 2021; Yu et al., 2021; Gonzalez-Zamora et al., 2022). Physiological signals, biochemical assessments, and questionnaires are used to monitor exercise fatigue (Halson and Shona, 2014). Questionnaires are subjective. Invasive biochemical tests cause discomfort to the subjects. Noninvasive physiological testing with scientific and statistical approaches can provide confidence and certainty.Muscle fatigue is divided into central and peripheral components. Peripheral fatigue is caused by changes in the neuromuscular junction. Central fatigue originates in the central nervous system (CNS). The production of skeletal muscle force depends on contractile mechanisms, and failure of nervous, ion, vascular, and energy systems can contribute to the development of muscle fatigue (Wan et al., 2017). During sustained contractions, metabolic changes in the muscle affect the propagation of action potential. These changes result in a progressive reduction of muscle fiber conduction velocity (CV). This is one of the main causes of the changes in amplitude and spectral EMG variables during fatigue. This physiological variable provides a relevant means to describe and quantify the muscle fatigue (Marco et al., 2017).Exercise fatigue can be detected through surface electromyography (sEMG) (Chang et al., 2012). Constant exercise load results in the rise of EMG activity during fatigue (Chlif et al., 2018). Electrical currents generated by muscle contractions can be monitored in the form of EMG signals and displayed on a computer (Barszap et al., 2016). Bioelectricity is detectable when muscle contracts (Tang et al., 2020). Electromyographic techniques are used in sports and rehabilitation accepted by researchers (Cortez et al., 2017; Pakosz and Konieczny, 2020; Quittmann et al., 2020). EMG can be recorded by needles inserted into the muscle (Kwon et al., 2018) or electrodes placed over the skin surface (Zeng et al., 2021). Compared with invasive EMG, sEMG is more popular among researchers (Khan et al., 2019; Yamagishi et al., 2019; Silva et al., 2020).This review offers an overall summary of the assessment of human exercise fatigue based on sEMG signals and highlights signal processing, feature extraction, and classification. The possibilities for future work on exercise fatigue with sEMG will also be discussed in this study.The sEMG is a biological electric signal generated by muscle contraction that can be harvested by electrodes. The sEMG signals are the most intuitive physiological signals of muscle activity and the best means to detect muscle fatigue. The sEMG signal is a kind of pseudorandom physiological signal that is very weak. The voltage of the sEMG signal range from 50 V to 100 mV and the frequency is varied from 10 to 500 Hz (Pancholi and Joshi, 2018). The electrode skin impedance which is one of the noises that affects the quality of EMG signals must be as low as possible to obtain effective signals (Sae-Lim et al., 2019). Pancholi and Joshi (2018) obtained sEMG signals from five different arm muscles using hardware based on ADS1298 IC (Texas Instruments) and ARM cortex M4 series processor with 4,000 Hz sampling frequency. De la Pena et al. (2019) recorded sEMG-related muscle fatigue in sports training using a portable prototype with a 5,000 Hz sampling rate. Zhao et al. (2020) acquired sEMG data based on a software platform for visualizing sEMG information on muscle fatigue during upper limb rehabilitation training. In Makaram et al. (2021), Biopac MP 36 (Biopac Systems Inc. CA, United States) was used to acquire brachii muscle of 52 healthy participants sEMG during dynamic contractions at an acquisition rate of 10,000 Hz. Wang L. et al. (2021) used wearable sampling electrodes that the sampling frequency is 200 Hz to collect sEMG signals in real-time during the driving tasks. Chen et al. (2021b) used the MP160 physiological record analysis system produced by the American company BIOPAC to analyze the fatigue of miners.Exact electrode positioning is vital for obtaining reliable EMG signals. A study showed high correlations between all electrode sites and clavicular movements. The traditional electrode site record more informative signals in subjects (Zanca et al., 2014). All trunk muscles were affected by electrode position changes, but the abdominal muscles were more affected than the back muscles (Huebner et al., 2015). Ghapanchizadeh et al. (2016) found that the optimal signal from the flexor carpi radialis muscles was presented at 90% and for the extensor carpi radialis muscles was shown at 90% of the electrode position over the forearm length. The muscle moves away from its uncontracted position directly under the EMG electrode when contracts. Elsais et al. (2020) used ultrasound to track the relative motion between skin and muscle to quantify the magnitude of the movement between them and inform protocols for surface EMG placement.The raw sEMG data collected will inevitably be mixed with power line interference and motion artifacts (Zheng and Hu, 2019). Therefore, effective preprocessing is required before feature extraction of these signals (Ahmadizadeh et al., 2021). Common methods of sEMG preprocessing include filtering (Zhao et al., 2020), normalization, and windowing (Fang et al., 2020). Tapia et al. (2017) used independent component analysis (ICA) and empirical mode decomposition (EMD) to process sEMG signals. (Wu et al., 2017) achieved better accuracy for the diagnosis of muscular fatigue through ensemble empirical mode decomposition (EEMD) by Hilbert transform (HT). Zhang et al. (2019) decomposed sEMG signals by principal component analysis (PCA) into principal components and weight vectors that improve the validity of parameters. In Avian et al. (2022), Discrete Wavelet Transform (DWT) is used to process the sEMG signal to increase model performance.The increase from baseline represents the onset of muscle activity. Muscle activity onset can be estimated from EMG and ultrasound (Dieterich et al., 2017). Gupta et al. (2014) determined the onset of medial gastrocnemius muscle activity using visual and automated methods during a stretch-shorten-cycle muscle action. Zhang and Zhou (2012) proposed a novel method of muscle activity onset detection based primarily on the sample entropy (SampEn) analysis of the surface EMG. Liu et al. (2015) presented an unsupervised EMG learning framework based on a sequential Gaussian mixture model (GMM) to detect muscle activity onsets. Appropriate signal preprocessing method is helpful to improve the effectiveness of feature extraction.Feature extraction from the sEMG signal plays an important role in the accuracy of fatigue detection. Time domain, frequency domain, time-frequency domain, and nonlinear parameters are four major types in sEMG-based signal processing (Too et al., 2018b; Yousif et al., 2019; Bukhari et al., 2020).The calculation of time-domain features is the most popular approach for sEMG feature extraction that is directly calculated with time as the independent variable. In the time domain, the sEMG signal is usually regarded as a random signal whose mean value is zero and variance varies with signal intensity. As the calculation of time-domain features is simple and intuitive, it is a widely used feature extraction method of sEMG signal in human movement (Harmon et al., 2021). The typical time-domain features of sEMG signals mainly include the root mean square (RMS) (Cui et al., 2021), integrated EMG (iEMG) (Alam et al., 2020), zero-crossing rate (ZCR) (Kim et al., 2018), waveform length (WL), variance of electromyography (VAR) (Whittaker et al., 2019), and mean absolute value (MAV) (Chapman et al., 2019).With the occurrence of muscle fatigue, the time domain features of sEMG generally show an upward trend over time (Goubault et al., 2022). RMS and iEMG not only reflect the amplitude changes of the sEMG signal in the time domain but also clearly reflect the biomechanical properties and muscle energy changes in the exercise process (Silvetti et al., 2017; Wu et al., 2017). Therefore, RMS and iEMG are often used to indicate muscle activation intensity and human motion state (Triwiyanto et al., 2018).The frequency-domain features are spectrum or power spectrum (PS) features that are obtained by fast Fourier transform (FFT) to the original sEMG signals, thus the frequency band distribution of the signals can be observed directly (Chandra et al., 2020). Researchers believe that frequency domain analysis is more meaningful than time domain analysis in both static and dynamic motion. To quantitatively describe the spectrum and PS features of sEMG signals, the mean frequency (MF) (Hou et al., 2021) and median frequency (MDF) (Park and Park, 2021) are generally used that decrease linearly over time. The deeper the muscle fatigue, the faster the MF decreases. Both MF and MDF can represent the frequency of measured muscle CV, but in practical application, MDF is more sensitive than MF in reflecting muscle activity and functional state. MF can also get good results in muscle fatigue detection (Chai et al., 2019). Significant changes in the PS indicate muscle fatigue, and the PS drifts from high frequency to low frequency. For example, after maximum weight training, the peak value of the sEMG signal will increase and drift to the lower frequency band. During extreme contraction, there is a decrease in muscle fiber action potential and possible emission frequency.To make up for the deficiency of Fourier transform analysis of non-stationary signals, time-frequency features of sEMG signals are introduced. The analysis of time-frequency features is important to the estimation of muscle fatigue state, and it is usually necessary to analyze both the time-domain and frequency-domain features of sEMG signals to obtain comprehensive information of muscle physiological changes. Traditional sEMG time domain and frequency domain analysis methods only describe the time domain or frequency domain features, but the time-frequency domain feature analysis method can overcome this limitation. At present, the available time-frequency analysis methods for sEMG signal mainly include Short-time Fourier Transform (STFT), Wavelet Transform (WT), Choi William distribution (CWD), and Wigner-Ville distribution (WVD), used for visual observation of signal frequency content evolution over time. WVD is a common bilinear time-frequency distribution, which is the most common type of Cohens time-frequency distribution. Instantaneous frequency parameters commonly used are instantaneous mean frequency (IMNF) (Triwiyanto et al., 2017) and instantaneous median frequency (IMDF) (Yousif et al., 2019), which show a downward trend with the deepening of fatigue degree. Average instantaneous MF has higher stability and sensitivity than frequency-domain features. Yousif et al. (2019) applied IMNF and IMDF to assess the muscles fatigue of the male runner during 400 m running with three types of running strategies.The sEMG complexity and entropy decrease linearly with the increase of fatigue degree. The complexity of Lempel-Ziv [C (n)] (Jo et al., 2018) is the speed at which new patterns appear with the increase of the length of time series, indicating the degree of randomness of the series, which decreases linearly in the process of dynamic motion. In dynamic motion, approximate entropy rises first and then decreases for most people. Marginal spectrum entropy (MSE) (Jero and Ramakrishnan, 2019) is a useful real-time muscle fatigue assessment method with the advantages of fast, reliable assessment of muscle fatigue and anti-noise. Compared with approximate entropy and MDF, MSE can be calculated quickly, the data length robustness is better, and muscle fatigue can be assessed reliably. It has high stability for different individuals and good noise resistance. SampEn (Cui et al., 2017), proposed by Richman and Moorman in 2000, measures the probability of generating new patterns in signals by measuring the complexity of time series. SampEn has a strong anti-noise ability and can reduce data deviation. The higher the complexity of the sequence, the higher the entropy. The multi-scale entropy (Fan et al., 2018) of sEMG signals decreases as the load increases, which more accurately represents the complexity of the muscle system. Compared with the traditional SampEn, it can more objectively reflect the working status and fatigue grade of the muscle. Multi-scale entropy analysis has a small amount of calculation and can adapt to the complexity of dynamic muscle contraction under time-varying loads. In practical work, normalized average multi-scale entropy can be used as a quantitative index to measure the dynamic fatigue degree of muscles. Recurrence quantification analysis (RQA) (Chen et al., 2018) is used to determine the percentage of line segments (%DET) reflecting the periodicity of signals. The RQA software is used to calculate and determine the percentage of line segments. Under dynamic and static loads, %DET increases linearly with the occurrence of muscle fatigue. Lejun et al. (2018) used C(n) to evaluate fatigue in all-out cycling exercises. Cui et al. (2017) used SampEn to investigate the fatiguing features of muscle-tendon units (MTUs) within skeletal muscles during static isometric contraction tasks. Hernandez and Camic (2019) investigated the effect of fatigue status and contraction type on the complexity of the sEMG signal, using SampEn and Detrended Fluctuation Analysis (DFA). Kahl and Hofmann (2016) compared the performance of different fatigue detection algorithms quantifying muscle fatigue based on sEMG signals. Fatigue detection algorithms including spectral moments ratio (SMR), SampEn, fuzzy approximate entropy (fApEn), and RQA (%DET) were calculated. After identifying the extracted features from the sEMG signal, the next important step is classification to detect the fatigue state.With the development of machine learning, machine learning algorithms were widely applied to exercise fatigue classification. Classification in this article normally refers to supervised learning where individuals are classified based on their features. Some classification algorithms based on sEMG are listed in Table 1, mainly including fuzzy logic (FL) (Li, 2017), hidden Markov model (HMM) (Shahmoradi et al., 2017), k-nearest neighbor (KNN) (Bukhari et al., 2020), support vector machine (SVM) (Chen et al., 2021b), linear discriminant analysis (LDA) (Ahmed et al., 2020), and artificial neural network (ANN) (Subasi and Kiymik, 2010).Machine learning algorithms for fatigue classification. Support vector machine is a popular machine learning classification method because it is simple, fast, and stable, and shows better accuracy than other methods (Karthick et al., 2018; Wang et al., 2018; Greco et al., 2019; Wang, 2021). Since the features extracted from the sEMG signal are the input of the classifier, the accuracy of classification results is closely related to the feature extraction method. Dimension reduction plays a crucial role in index extraction, which can reduce the calculation time. Common dimension reduction methods include PCA (Qi et al., 2020) and ICA (Too et al., 2018a).Khan et al. (2019) used the random forest trained by distributive power frequency of the sEMG signal of the vastus-lateralis muscle to predict muscle fatigue. They obtained a high accuracy in fatigue classification. Wang et al. (2020) proposed a muscle fatigue classification method based on sEMG signals to detect muscle fatigue by the Convolutional Neural NetworkSupport Vector Machine (CNN-SVM), Support Vector Machine, Convolutional Neural Network, and Particle Swarm OptimizationSupport Vector Machine algorithms. CNN-SVM algorithm achieves the highest accuracy rate in muscle fatigue classification.Zhang et al. (2021) detected muscle fatigue based on the Multidimensional Feature Fusion Network (MFFNet), which is composed of Attention Frequency domain Network (AFNet) and Attention Time-domain Network (ATNet). The result shows 77.37% higher than other classifiers. Bharathi et al. (2022) developed an automated muscle fatigue detection system and acquired 58 healthy volunteers signals under dynamic muscle fatiguing contractions. The extreme learning machine (ELM) model performs well with a 94.09% result. Wang J. H. et al. (2021) proposed a new muscle fatigue recognition model based on the long short-term memory (LSTM) network. Rejith et al. (2016) estimated the elbow kinematics under fatigue using sEMG by Multi-layered Perceptron Neural Network (MLPNN) which gave a classification accuracy of 60.12%. Chen et al. (2021a) studied fatigue of miners with physiological signals by extreme gradient boosting (XG-Boost).Multi-sensor fusion based on sEMG can collect more dimensions of human activities from multiple dimensions. Thus, it is more comprehensive than fusion methods that use data from a single sensor. Qi et al. (2018) proposed a method for driving fatigue assessment based on the electroencephalogram (EEG) and EMG. Zhao et al. (2020) proposed a wearable monitoring device by integrating electrocardiogram and electromyogram (ECG/EMG) sensors to acquire data for monitoring fatigue during rehabilitation training. Martinez-Aguilar and Gutierrez (2019) analyzed cortico-muscular and cortico-cardiac coupling to study the development of muscular fatigue by electromyography (EMG), electrocardiography (ECG), and electroencephalography (EEG). Bilgin et al. (2015) presented a muscle fatigue detection method based on the frequency spectrum of EMG and mechanomyogram (MMG). Scano et al. (2020) studied the combined use of NIRS and sEMG in muscle fatigue assessment.Fatigue detection based on surface EMG has important application value in sports training, rehabilitation treatment, and movement recognition. This article aimed to provide an overview of sEMG signal processing, feature extraction, and classification in exercise fatigue. In real-time detection, portability of the device, removal of artifacts, feature extraction, and classification techniques should be properly investigated. Using appropriate methods can remove noise to improve EMG signal quality. With the increase in the number of EMG channels and features, it is necessary to choose a reasonable dimensionality reduction method. The methods should greatly reduce the computational complexity of the classifier and preserve maximum information of the signal. Classification techniques have been extensively studied, but their timeliness and generalization still have research significance. A combination of processing methods and pattern recognition techniques may be helpful to increase the classification speed and accuracy. And the adaptability of good algorithms to fresh samples needs further study. Finally, we propose that the current review can be used as a guide for further improving exercise fatigue assessment based on sEMG for various applications of the human body.JS mainly wrote the manuscript under the guidance of GL and JC. YS, KL, and ZZ participated in the analysis. All authors contributed to the article and approved the submitted version.The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.This work described in this manuscript was supported by the National Key Research and Development Program of China (2018YFF0300806-1) and the Science and Technology Development Plan Project of Jilin Province (20200404205YY).Ahmadizadeh C., Khoshnam M., Menon C. (2021). Human machine interfaces in upper-limb prosthesis control: a survey of techniques for preprocessing and processing of biosignals.IEEE Signal Proc. Magaz.381222. 10.1109/msp.2021.3057042 [DOI] [Google Scholar]Ahmed S. F., Joyo M. K., Mahdi H. F., Kiwarkis I. J. (2020). 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The development of muscle fatigue is typically quantified as a decline in the maximal force or power capacity of muscle, which means that submaximal contractions can be sustained after the onset of muscle fatigue. There is even evidence that the duration of some sustained tasks is not limited by fatigue of the principal muscles. Here we review experimental approaches that focus on identifying the mechanisms that limit task failure rather than those that cause muscle fatigue. Selected comparisons of tasks, groups of individuals and interventions with the task-failure approach can provide insight into the rate-limiting adjustments that constrain muscle function during fatiguing contractions.Although it is not difficult to know when one is fatigued, it is entirely another matter to be able to identify the physiological mechanisms responsible for this condition. Despite the accumulation of a substantial literature on the topic since the seminal work of Angelo Mosso in the late 1800s (Di Giulio et al. 2006), few principles have emerged to characterize the phenomenon of muscle fatigue. Although progress has been made in the study of muscle fatigue (Nybo & Secher, 2004; Cairns, 2006; Nordstrom et al. 2007; Nybo & Rasmussen, 2007), we are largely unable to state with certainty why an individual becomes fatigued under various conditions. The purpose of this topical review is to examine three issues that constrain a more complete understanding of muscle fatigue and its impact on muscle function: the diversity of measures that have been used to quantify fatigue, the specificity of the impairments that cause fatigue and the lack of knowledge on the mechanisms that limit performance. The discussion will focus on what is muscle fatigue, why does it happen and how does it influence muscle function.What is muscle fatigue? In contrast to the fatigue encountered in clinical settings (Bailey et al. 2007; Friedman et al. 2007; Hacker & Ferrans, 2007), the term muscle fatigue is used to denote a transient decrease in the capacity to perform physical actions. The following excerpts characterize the range of effects ascribed to muscle fatigue:Intensive activity of muscles causes a decline in performance, known as fatigue (Allen & Westerblad, 2001).Performing a motor task for long periods of time induces motor fatigue, which is generally defined as a decline in a person's ability to exert force. (Lorist et al. 2002).CNS administration of caffeine increased treadmill run time to fatigue (Davis et al. 2003).a fatiguing task was performed with the muscles of the left hand until the muscles were exhausted. (Edgley & Winter, 2004).Fatigue is known to be reflected in the EMG signal as an increase of its amplitude and a decrease of its characteristic spectral frequencies. (Kallenberg et al. 2007).the sensation of fatigue is the conscious awareness of changes in subconscious homeostatic control systems (St Clair Gibson et al. 2003).The primary purpose of the study was to use functional magnetic resonance imaging (fMRI) to determine the association between feelings of mental fatigue and blood oxygen level dependent (BOLD) brain responses during a mentally fatiguing cognitive task. (Cook et al. 2007).Muscle fatigue, it seems, can refer to a motor deficit, a perception or a decline in mental function, it can describe the gradual decrease in the force capacity of muscle or the endpoint of a sustained activity, and it can be measured as a reduction in muscle force, a change in electromyographic activity or an exhaustion of contractile function. Such broad usage is problematic, however, because fatigue in this context can encompass several phenomena that are each the consequence of different physiological mechanisms, which reduces the likelihood that the cause of muscle fatigue can be identified. To circumvent this limitation, most investigators invoke a more focused definition of muscle fatigue as an exercise-induced reduction in the ability of muscle to produce force or power whether or not the task can be sustained (Bigland-Ritchie & Woods, 1984; Sgaard et al. 2006).A critical feature of this definition is the distinction between muscle fatigue and the ability to continue the task. Accordingly, muscle fatigue is not the point of task failure or the moment when the muscles become exhausted. Rather, muscle fatigue is a decrease in the maximal force or power that the involved muscles can produce, and it develops gradually soon after the onset of the sustained physical activity. A common protocol used to quantify the development of muscle fatigue is to interrupt the fatiguing exercise with brief maximal contractions (voluntary or electrically evoked) to estimate the decline in the maximal force capacity (Merton, 1954; Bigland-Ritchie et al. 1986b; Hunter et al. 2004b; Sgaard et al. 2006). Similarly, the amount of muscle fatigue caused by an intervention can be quantified as the decline in the maximal force or power measured immediately after the fatiguing contraction (Taylor et al. 1996; Griffin et al. 2001; Hunter et al. 2004a; Lvnez et al. 2005; McNeil et al. 2006).Why does fatigue occur? The simple answer is that one or several of the physiological processes that enable the contractile proteins to generate a force become impaired. A more complete answer, however, also acknowledges that the site of impairment depends on the task being performed. This effect is known as the task dependency of muscle fatigue and is one of the principles to have emerged in this field over the last 100 years (Asmussen, 1979; Enoka & Stuart, 1992; Bigland-Ritchie et al. 1995). According to this principle, there is no single cause of muscle fatigue and the dominant mechanism is specific to those processes that are stressed during the fatiguing exercise (Cairns et al. 2005). This concept is analogous to the principle of specificity that characterizes the adaptations evoked by several weeks of physical training (Kraemer et al. 2002; Aagaard & Bangsbo, 2006).Due to the specificity of the impairments that occur during fatiguing contractions, there are no general answers to such questions as are old adults more fatigable than young adults, are women less fatigable than men, can the nervous system sustain an adequate activation of muscle during fatiguing contractions, and are there differences between muscles? The difficulty in answering such questions becomes obvious when the results obtained in selected studies are compared. First, consider the influence of age on fatigability. Baudry et al. (2007) measured the decline in torque when young (mean s.d.; 30.5 2.5 years) and old (77.2 1.4 years) men and women performed maximal shortening and lengthening contractions with the dorsiflexor muscles. The task was to perform five sets of 30 maximal contractions at a rate of one contraction every 3.5 s, and with each contraction comprising a 30 deg range of motion and the speed controlled by a motor at 50 deg s1. There was a 60 s pause between each set of contractions. The shortening and lengthening contractions were performed on separate days. The young adults were stronger than the old adults as indicated by a greater peak torque during a maximal isometric contraction (38.3 3.1 and 28.6 1.3 Nm, respectively). Nonetheless, the decline in peak torque during both tasks was greater for the old adults. Figure 1A shows the data for the lengthening contractions in which the final peak torque declined by 27.1% for the young adults and by 42.1% for the old adults. The decrease in peak torque after 150 maximal shortening contractions was 40.9% for the young adults and 50.2% for the old adults. Furthermore, the decline in peak torque during each set of 30 maximal contractions was greater for the old adults for both the shortening and lengthening contractions (Fig. 1A). The fatigue experienced by both groups of subjects was associated with changes in the control of excitationcontraction coupling by Ca^{2+} and, for the old adults only, impairment of neuromuscular propagation. These results indicate therefore that the old adults were more fatigable than the young adults when performing maximal shortening and lengthening contractions with the dorsiflexor muscles. A, average torque exerted by young and old adults during five sets of 30 maximal lengthening contractions with the dorsiflexor muscles. Each data point indicates the mean s.e.m. of five successive contractions for 16 subjects in each group. Adapted with permission from Baudry et al. (2007). B, each data point denotes the time to failure for the one young man and one old man who were matched for strength. The task was to sustain an isometric contraction with the elbow flexor muscles at 20% of maximum for as long as possible. The time to failure was longer for the older man of each pair. Adapted from Hunter et al. (2005).In contrast to these results with maximal anisometric contractions, Hunter et al. (2005) found that old men (71.3 2.9 years) could sustain a submaximal isometric contraction with the elbow flexor muscles for a longer duration than young men (21.5 4.4 years). The task was to sustain a net muscle torque that was 20% of the maximal voluntary contraction (MVC) torque for as long as possible. Each subject was required to match the exerted torque to a target that was displayed on a monitor. The task was terminated when the torque exerted by the subject declined by 10% of the target torque or greater for at least 5 s. Eight pairs of men were matched for MVC torque (young: 65.9 8.0 Nm; old: 65.4 8.7 Nm), which meant that they exerted a similar net muscle torque during the fatiguing contraction. The time that the old men could sustain the submaximal isometric contraction was longer (22.6 7.4 min) than that for the young men (13.0 5.2 min) as indicated by all data points lying above the line of identity in Fig. 1B. The decrease in MVC torque at task failure, which indicates the amount of fatigue that occurred during the fatiguing contraction, was similar for the two groups of men (31.4 10.6%). The briefer time to failure for the young men was associated with more rapid increases in the amplitude of the surface EMG, the frequency of bursts of activity in the EMG signal, the rating of perceived exertion, mean arterial pressure and heart rate. Hence, the rate at which fatigue developed during the submaximal isometric contraction was more gradual for the old men, which contrasts with the results reported by Baudry et al. (2007) for maximal anisometric contractions with the dorsiflexor muscles.The second example on the specificity of the impairments during fatiguing contractions involves the sex difference that is often observed during such tasks. Women are usually able to sustain a contraction for a longer duration, especially at lower contraction intensities (Hicks et al. 2001; Clark et al. 2005; Hunter et al. 2006) but not maximal contractions (Baudry et al. 2007). The most common explanations for this sex difference are the greater muscle mass activated by men and a lesser reliance on glycolytic metabolism by women. As men are typically stronger than women, they must activate a larger muscle mass to exert the same relative force (% maximum) as women, which will be accompanied by larger intramuscular pressures and a greater occlusion of blood flow (de Ruiter et al. 2007). The influence of strength on the duration that submaximal contractions could be sustained was examined by comparing the performance of men and women who were matched for strength. Subjects were matched for MVC torque, which meant that the target torque (20% of maximum) required the same absolute torque (men: 12.6 1.6 Nm; women: 12.8 1.6 Nm) by the elbow flexor muscles (Hunter et al. 2004a). There was no difference in the duration that the men and women could sustain the contraction (819 306 and 864 391 s, respectively) and the decline in MVC force (31 9%) at task failure (Fig. 2A). Accordingly, there were no differences between the men and women in the rates of increase for the amplitude of the surface EMG, the frequency of bursts of activity in the EMG signal, the rating of perceived exertion, mean arterial pressure or heart rate. Furthermore, Clark et al. (2005) found that the duration an isometric contraction could be sustained at 25% MVC force with the knee extensor muscles was similar for men (165 20 s) and women (180 20 s) only when blood flow was occluded (Fig. 2B), even though the peak force that the men could exert (905 34 N) was greater than that for the women (722 34 N). A, mean (+s.e.m.) time to task failure for sustained and intermittent isometric contractions performed by strength-matched men and women with the elbow flexor muscles. The target torque was 20% of MVC torque for the sustained contractions and 50% of MVC torque for the intermittent contractions. Reprinted with permission from Hunter et al. (2004b). B, mean (s.e.m.) time to task failure for sustained isometric contractions performed by men and women with the knee extensor muscles with the blood flow to the limb either occluded or not occluded. The men were stronger than the women in this study. The target was 25% of MVC torque. Data from Clark et al. (2005).Nonetheless, strength cannot explain all of the sex differences in fatigability. For example, women can perform a greater number of intermittent contractions than men even when the two groups are matched for strength. Hunter et al. (2004b) matched the MVC torque of the elbow flexors for men (64.8 9.2 Nm) and women (62.2 7.9 Nm) and found that the women (1408 1133 s) could perform the intermittent contraction (6 s contraction, 4 s rest) to a target of 50% MVC torque for a longer duration than the men (513 194 s) (Fig. 2A). Despite the contractions requiring the same absolute torque, the rate of decline in MVC torque was greater for the men (3.0 Nm min1) compared with the women (0.97 Nm min1). In contrast, the decrease in MVC torque for the dorsiflexor muscles after performing 24 MVCs in 4 min was similar for men (79 4%) and women (78 4%) when blood flow was occluded, but the decline in MVC torque differed (men: 34 3%; women: 21 4%) when blood flow was not occluded (Russ & Kent-Braun, 2003). As the supply and demand of ATP is altered during the occlusion of blood flow (Lanza et al. 2006), the lower rate of fatigue development observed in women during intermittent contractions may be attributable to a sex difference in the relative contributions of the metabolic pathways that supply ATP during fatiguing contractions (Russ et al. 2005).The third example on the specificity of the impairments during fatiguing contractions concerns the potential contribution to the decrease in force by a reduction in the activation of muscle by the nervous system. If activation is inadequate, a few electrical stimuli delivered to the motor nerve during an MVC will evoke additional force from the muscle (Merton, 1954). The deficit in activation is quantified by normalizing the amplitude of the force increment to the force that can be elicited by the same stimulus when the muscle is at rest; the ratio is expressed as an index of voluntary activation (Hales & Gandevia, 1988; Kent-Braun & Le Blanc, 1996). Voluntary activation (% maximum) during brief MVCs varies among individuals, across days, and between trials (Allen et al. 1995), and among muscles (Belanger & McComas, 1981; McKenzie et al. 1992; Klass et al. 2007). A decrease in voluntary activation, as can occur during long-lasting contractions, can contribute to the development of fatigue.The magnitude of the decline in voluntary activation during fatiguing contractions ranges from minimal to substantial. When subjects sustained an MVC with the elbow flexor muscles for 3 min, Gandevia et al. (1996) found that the 74.1 8.6% decline in force was accompanied by only a 10% decrease in voluntary activation of biceps brachii. In contrast, Babault et al. (2006) observed a more pronounced decrement in voluntary activation when subjects performed sets of maximal isometric and shortening contractions with the knee extensor muscles. Three sets of maximal shortening contractions reduced MVC torque by 59.0 8.1% and voluntary activation by 26.5 2.1%, whereas three matched isometric contractions decreased MVC torque by 57.9 8.6% and voluntary activation by 35.7 3.8%. The time course of the decrease differed in the two tasks (Fig. 3). Similarly, Sgaard et al. (2006) reported that a submaximal isometric contraction (25% MVC torque) sustained with the elbow flexor muscles for 43 min decreased MVC torque by 41.4 13.2% and voluntary activation measured in response to stimulation of the motor nerve from 98.0 5.1% prior to the fatiguing contraction to 71.9 38.9% (27% reduction) at the end of task. Smith et al. (2007) estimated that two-thirds of the 28% decrease in MVC torque after a 70 min isometric contraction sustained at 5% MVC torque with the elbow flexor muscles was due to a decline in voluntary activation. These studies suggest that some of the fatigue experienced during repetitive maximal contractions and long-duration submaximal contractions can be caused by an inadequate activation of muscle by the nervous system. Furthermore, the insufficiency of the activation can be exacerbated by such conditions as a decline in the levels of blood glucose and exercise in hot environments (Nybo & Nielsen, 2001; Nybo, 2003; Todd et al. 2005). A, decrease in MVC torque after three sets of maximal shortening contractions (60 deg s1 over a 75 deg range of motion) and three matched maximal isometric contractions with the knee extensor muscles. The two tasks were performed on separate days and the isometric contractions were sustained to produce the same relative decrease in MVC torque. The data correspond to mean +s.e.m. for the shortening contractions and isometric contractions for the first, second and third sets of shortening contractions and the three isometric contractions. Adapted with permission from Babault et al. (2006). B, corresponding decreases in voluntary activation as assessed with paired stimuli (10 ms apart) during selected maximal contractions. Initial values for voluntary activation were 88.3 3.0% for the session in which the shortening contractions were performed and 89.4 3.1% in the session for the isometric contractions. Data from Babault et al. (2006).The final example on the specificity of the impairments examines some of the differences that have been observed across muscles. Since the early observations on the differences in the contractile properties of red and white muscle and the classification of motor units based on fatigability (Buller et al. 1960; Edstrm & Kugelberg, 1968; Burke et al. 1973), the endurance capacity of muscle, at least in experimental animals with electrically evoked contractions, can depend on the proportion of its muscle fibre types (Cairns et al. 2005). When similar stimulation protocols are applied to motor units in human muscle, however, it is not possible to identify comparable groups of motor units based on differences in contractile properties, including fatigability (Bigland-Ritchie et al. 1998). Furthermore, human muscles are comprised of a mixture of fibre types that blurs potential differences in fatigability attributable to the proportions of the various fibre types in a muscle, and all voluntary actions involve multiple muscles.Nonetheless, there are other differences between muscles that can influence the adjustments during fatiguing contractions, such as the afferent feedback delivered by group IIIIV afferents that are activated by the metabolic by-products of the muscle contraction. One strategy that has been used to identify a contribution by group IIIIV afferents to the development of fatigue is to compare the recovery of muscle function when blood flow is normal and when it is impeded. When blood flow is occluded, the metabolites that accumulate in the fatigued muscle will continue to provide a stimulus that sustains the discharge of group IIIIV afferents (Hayes et al. 2006). With this approach, Bigland-Ritchie et al. (1986a) found that the depression in discharge rate of motor units in biceps brachii after a sustained MVC did not recover during the 3 min when blood flow was occluded but did recover to control values within 3 min when blood flow was not occluded. Such results and those of Duchateau & Hainaut (1993) suggest that a peripheral reflex mediated by group IIIIV afferents from the fatigued muscle contributed to the decrease in discharge rate during this type of exercise. In contrast, others found that feedback delivered by afferents that are sensitive to an occlusion of blood flow did not contribute to the decrease in voluntary activation after a 2 min MVC with the elbow flexor muscles (Taylor et al. 2000; Butler et al. 2003).Subsequent studies, however, have demonstrated that the contribution of group IIIIV afferents to the development of fatigue can vary across muscles. Martin et al. (2006) examined the contribution of feedback by group IIIIV afferents to the fatigue experienced during sustained 2 min MVCs with the elbow flexor and extensor muscles. The protocols involved comparing the amplitude of potentials evoked in muscle with transmastoid stimulation of the corticospinal tract to assess the excitability of motor neurones during MVCs and during recovery when blood flow to the muscles was occluded and when it was intact. The influence of these afferents on motor neurone excitability was estimated by comparing the amplitude of the evoked potentials in the presence and absence of ischaemia that altered the feedback delivered by group IIIIV afferents. When the fatiguing contraction was performed with the triceps brachii muscle, the amplitude of the evoked response decreased during the MVC and remained depressed while blood flow was occluded, but recovered within 15 s with the restoration of blood flow. The amplitude of the potentials evoked in triceps brachii also decreased after the fatiguing contraction was performed with the elbow flexor muscles. In contrast, the amplitude of the potentials evoked in biceps brachii increased after a fatiguing contraction with triceps brachii. These results indicate that group IIIIV afferents depressed the excitability of triceps brachii motor neurones, but facilitated those that innervate biceps brachii. Given the known differences in central connections between muscles that can exert different actions (Pierrot-Deseilligny & Burke, 2005), it seems likely that fatiguing contractions can involve other differences in adjustments between motor neurone pools.These four examples underscore the specificity of the impairments that contribute to the development of muscle fatigue and thereby make it impossible to identify a single causal mechanism for muscle fatigue. Furthermore, the extensive literature on the specificity of the impairments demonstrates that it is possible to impair any of the physiological processes involved in the activation of the contractile proteins and the force that they can produce. However, relatively little is known about the functional relevance of these impairments and the conditions that can trigger a change in the prevailing mechanisms.How does fatigue influence muscle function? By definition, a muscle begins to experience fatigue as soon as its maximal force or power capacity starts to decline. When the task involves sustaining a maximal contraction, the decline in performance parallels the increase in fatigue. When the task requires a submaximal contraction, however, the onset of fatigue is probably not associated with the termination of the task. As most activities of daily living involve submaximal forces, the onset of fatigue may not limit the ability of an individual to perform a task and, furthermore, task failure may not be caused by fatigue of the principal muscles involved in the task.The association between task failure and muscle fatigue has been studied in protocols that required subjects to inspire against various resistive loads. The force capacity of the inspiratory muscles can be estimated from the maximal inspiratory pressure an individual can generate, which McKenzie et al. (1997) measured as subjects inspired for 20 min against resistive loads that ranged from 35 to 90% of maximal inspiratory pressure. Task failure occurred when subjects came off the mouthpiece or when they could not reach the target pressure throughout inspiration for three consecutive breaths. All subjects experienced task failure before 20 min with the high loads (7590%). Task failure was associated with the development of hypercapnia and dyspnoea (hunger for air) and not either muscle fatigue or an impairment of voluntary activation. Gorman et al. (1999) used a similar protocol with a load that was 80% of maximal inspiratory pressure and defined task failure as an intolerable discomfort of breathing. The duration that breathing could be sustained was related to the accumulation of CO_2 and not fatigue of the inspiratory muscles. Similarly, although Rohrbach et al. (2003) could detect fatigue in the diaphragm and rib cage muscles when subjects inspired against a load that was 67% of maximal inspiratory pressure, the duration that breathing could be sustained was not related to the amount of fatigue in these muscles, which suggested that such factors as the accumulation of CO_2 and a decrease in arterial oxygen saturation probably caused task failure.Task failure has also been studied in limb muscles. One approach has been to compare the performance of two similar tasks and to identify the adjustments that limit the duration of the more difficult task (Hunter et al. 2004c; Maluf & Enoka, 2005). In one task, referred to as the force task, the limb was attached to a restraint and the subject was required to maintain a constant force for as long as possible while viewing force feedback on a monitor. The other task, referred to as the position task, required the subject to support an inertial load that was equivalent to the force exerted during the force task, and to maintain a constant joint angle for as long as possible while viewing position feedback on a monitor. Although both tasks involve sustained isometric contractions and required the same net muscle torque for each subject, stretch reflexes evoked during such tasks suggest that the neural strategy underlying the same motor output differs (Doemges & Rack, 1992a,b; Dietz et al. 1994). Furthermore, the recruitment thresholds of motor units in biceps brachii differ for tasks that involve maintaining a position or exerting a force (Tax et al. 1989).Maluf et al. (2005) compared the performance of the first dorsal interosseus muscle on the force and position tasks when the hand was placed in the position shown in Fig. 4A. The extended index finger exerted an abduction force (20% of maximum) against a force transducer for the force task and supported an inertial load while maintaining a constant angle at the metacarpophalangeal joint for the position task. The force task was terminated when the subject was unable to maintain the abduction force within 1.5% of the target force, whereas the position task was ended when the finger could not be held within 10 deg of the target angle. The fluctuations in the abduction force and the joint angle during the two tasks are shown in Fig. 4B. The duration that the position task could be sustained (593 212 s) was substantially less than that for the force task (983 328 s). The briefer duration for the position task was associated with a more rapid increase in the amplitude of the surface EMG for first dorsal interosseus (Fig. 4B and C). Due to minimal EMG activity in the antagonist muscle (second palmar interosseus), the greater rate of increase in EMG amplitude for the agonist suggested a more rapid recruitment of the motor unit pool during the position task, even though the net muscle torque was similar for the two tasks. To examine this possibility, Maluf et al. (2005) compared the two tasks at a target force that corresponded to the upper limit of motor unit recruitment for first dorsal interosseus (60% MVC force). Consistent with the explanation that the position task involved a more rapid recruitment of motor units when the target force was 20% of MVC force, there was no difference in the time to failure for the force (93 41 s) and position tasks (86 31 s) when all motor units were recruited at the onset of each contraction. A, the two tasks were performed with the first interosseus muscle with the hand in the posture shown. The index finger pushed up against a rigid restraint for the force task and supported an inertial load for the position task. B, representative data for the position (top) and force (bottom) tasks as performed by one individual. For each task, the data comprise the EMG for the antagonist (second palmar interosseus) and agonist (first dorsal interosseus) muscles, and either the joint angle (position task) or the abduction force (force task). Adapted with permission from Maluf et al. (2005). C, group data (n = 20) for EMG amplitude for the agonist and antagonist muscles during the force and position tasks. EMG amplitude was calculated for each 1% interval of the time to failure and averaged across subjects. 95% confidence intervals are shown for the first, middle and last data point of each condition. Adapted from Maluf et al. (2005).Motor unit recordings confirmed this conclusion. Mottram et al. (2005) recorded the discharge of the same motor unit (n = 32) when subjects performed the two tasks with the same load (20% MVC force) for 161 96 s. The contractions with the elbow flexor muscles were not performed to task failure so that it was possible to record the discharge of the same motor unit in a single experimental session. The two tasks were performed in a randomised order and sufficient rest was given so that there was no statistical difference in MVC force at the start of each task. The average discharge rate (13.1 pps) and coefficient of variation for interspike interval (22.8%) were similar at the beginning of the two contractions, but discharge rate decreased more and the coefficient of variation for discharge times increased more during the position task (Fig. 5). The greater increase in the coefficient of variation during the position task suggests a more rapid rise in synaptic noise during this task (Stein et al. 2005). Despite the decline in discharge rate, the synaptic input to the motor neurone pool increased during the two contractions and recruited additional units. The subcutaneous recordings detected the recruitment of the same 26 units during the two tasks, but another 6 units that were recruited only during the force task and another 20 units that were activated only during the position tasks. The enhanced recruitment during the position task was accompanied by more rapid increases in both the perceived exertion and mean arterial pressure. Even though each subject exerted the same net muscle torque during the two tasks, the motor output was sustained by different motor unit activity. A, mean s.e.m. discharge rate at the beginning, middle and end of the force and position tasks for the 32 motor units. The decrease in mean discharge rate was significantly greater during the position task (13.110.6 pps) than during the force task (13.312.0 pps). B, the coefficient of variation for interspike interval did not change from the beginning (23.1 7.8%) to the end (22.4 8.6%) of the force task, whereas it increased significantly during the position task (22.4 10.4 to 26.7 9.2%). Data from Mottram et al. (2005).Insight into the rate-limiting mechanisms can be obtained by comparing performances across individuals and with practice of the two tasks. The duration that the force and position tasks can be sustained at the same relative intensity varies widely among individuals. For example, the time to failure for the elbow flexor muscles when performing the position task while supporting a load equivalent to 20% MVC force (17.3 3.7 Nm) ranged from 176 to 1411 s (567 353 s) for 20 subjects (Rudroff et al. 2007b). Based on a multiple regression analysis, 78% of the variability in the time to failure could be explained by three variables: the frequency of EMG bursts for the long head of biceps brachii during the first 20% of the contraction, the amplitude of the EMG bursts for brachioradialis during the first 20% of the contraction and the target torque. Individuals with longer times to failure had greater EMG burst frequencies for the long head of biceps brachii, lower EMG burst amplitudes for brachioradialis and lower target torques. As bursts of activity in the surface EMG correspond to the transient recruitment of higher threshold motor units, individuals with longer times to failure exerted a lower muscle force and transiently recruited higher threshold motor units more frequently in one primary muscle (long head of biceps brachii) and intermittently recruited units with smaller amplitude action potentials in the surface EMG of another primary muscle (brachioradialis).Consistent with this result, Hunter & Enoka (2003) previously found that the capacity to alter the bursts of EMG activity enabled some subjects to prolong the time to failure with practice. Subjects performed the force task with the elbow flexor muscles at a target force of 20% MVC force on three occasions with at least 2 weeks between each performance. A cluster analysis classified the subjects into two groups: those who increased time to failure (responders) and those who did not (non-responders). The responders increased time to failure from session 1 to session 3 by > 15% (60 28%; range: 25115%), whereas the non-responders did not experience any change in the time to failure across sessions (3 11%; range: 1215%). MVC force was similar for the responders (289 117 N) and non-responders (290 114 N) and decreased by a similar amount (31.3 16.6 and 35.9 18.1%, respectively) at task failure. The most significant adaptation associated with the prolongation of the time to failure was a delay in the onset of the bursts of activity in the surface EMG for the elbow flexor muscles. Thus, some individuals were able to improve performance by delaying the transient recruitment of higher threshold motor units. As the net muscle torque was the same across the three sessions, varying the number of activated motor units influences the relative force contributed by the involved muscle fibres. The increase in the time to failure exhibited by the responders therefore indicates that task failure in session 1 was not attributable to a decline in the force capacity of the activated muscle fibres, but rather was a consequence of the strategy used to perform the task.Although activity in postural muscles can influence the duration a task can be sustained (Le Bozec & Bouisset, 2004), few studies have examined the influence of synergist and accessory muscles on the performance of fatiguing contractions (Caffier et al. 1992; Sacco et al. 1997). When the force and position tasks were performed with the elbow flexor muscles with the arm in the position shown in Fig. 6A, the ratio of the times to failure for the two tasks at the same relative intensity (20% MVC force) differed from that observed for the hand muscle. The time to failure for the position task (702 582 s) was 51 26% of the duration for the force task (1402 728 s) for the elbow flexor muscles, compared with 63 28% for first dorsal interosseus. Subjects reported that performance of the position task with the elbow flexor muscles when the arm was abducted from the trunk by about 0.4 rad (Fig. 6A) imposed an additional demand on shoulder muscles that was not experienced during the force task. The lesser ratio for the elbow flexor muscles with the arm in this posture indicates the significant influence that postural activity can have on limiting task performance. Accordingly, Rudroff et al. (2007a) found that the rate of increase in the amplitude of the EMG activity for selected shoulder and trunk muscles was much greater during the position task compared with the force task (Fig. 6B). A, the two tasks were performed with the elbow joint at a right angle and the upper arm abducted from the trunk by about 0.4 rad (23 deg). B, the average rectified EMG (mean s.e.m.) for infraspinatus (circles), supraspinatus (squares) and teres minor (triangles) during the force and position tasks. The target force was 20% MVC force. The EMGs were normalized to the peak values observed during the MVCs and averaged over 30 s epochs at 20% intervals during each task. Adapted with permission from Rudroff et al. (2007a). C, rectified EMG averaged over 30 s epochs at 25% intervals for the duration of the position task and at the end of the force task across the biceps brachii, brachialis and brachioradialis muscles during the two tasks. The target force was 15% MVC force. The EMGs were normalized to the peak values observed during the MVCs. Data from Hunter et al. (2002).The relative times to failure for the force and position tasks when performed with the elbow flexor muscles are also influenced by the orientation of the forearm. When the arm was rotated forward so that the upper arm was horizontal and not abducted from the trunk, there was no difference in the time to failure for the force (476 246 s) and position tasks (468 270 s) when the forearm was in a neutral position, whereas the position task (477 276 s) was briefer than the force task (609 250 s) when the forearm was supinated (Rudroff et al. 2005, 2007a). Despite the influence of limb posture on the relative times to failure for the two tasks, the rate of increase in the amplitude of the EMG activity was similar for the two tasks in both postures (Fig. 6C). In contrast to the more rapid increase in EMG amplitude during the position task for first dorsal interosseus (Fig. 4B), the comparable rates of increase in EMG amplitude for the elbow flexor muscles suggest that the briefer times to failure for the position task are due to an inability to increase the EMG as much as during the force task (Fig. 6C). As the activity of motor units in biceps brachii does differ when the arm posture facilitates a briefer duration for the position task (Mottram et al. 2005), another mechanism appears to constrain the increase in EMG activity during the position task. A likely candidate is the reflex connections between the elbow flexor muscles. Although these muscles all contribute to the flexor torque, they produce opposing actions about the pronationsupination axis of the forearm (Zhang et al. 1998) and there are inhibitory pathways between some of them (Naito et al. 1996, 1998).The strategy used by the nervous system to perform the force and position tasks probably involves heightened activation of the stretch reflex pathway during the position task. The amplitude of the stretch reflex, for example, is enhanced when a limb acts against a compliant load compared with a rigid restraint (Akazawa et al. 1983; De Serres et al. 2002). An increase in the stretch reflex response suggests that the task involves greater net excitation from spinal and supraspinal sources, which results in the earlier recruitment of the motor unit pool and thereby premature task failure (Maluf et al. 2005; Mottram et al. 2005). Although the more rapid recruitment of the motor unit pool can be explained by an increase in the descending drive to the motor neurone pool (Lscher et al. 1996), the greater decrease in mean discharge rate and increase in discharge variability during the position task suggests the involvement of other mechanisms also (Mottram et al. 2005). As these effects were observed in the same motor unit during the two tasks, the different adjustments can probably be attributed to variation in peripheral synaptic input to the motor neurones rather than to changes in the intrinsic properties of the motor neurones.Accordingly, the amplitude of the Hoffmann reflex in biceps brachii decreased sooner during the position task compared with the force task, whereas the amplitude of the compound muscle potential evoked in biceps brachii by transcranial magnetic stimulation increased at a similar rate during the two tasks (Duchateau et al. 2007). These observations suggest that the difference in the synaptic input received by the motor neurone pool probably involved presynaptic rather than postsynaptic mechanisms (Maluf et al. 2007). This interpretation is consistent with the observation that the application of muscle vibration during a sustained fatiguing contraction, an intervention that enhances the excitatory input from Ia afferents to the motor neurone pool, can transiently restore the discharge rate of motor units (Bongiovanni & Hagbarth, 1990; Griffin et al. 2001). Taken together, these data suggest that the duration of the position task may be limited by spinal mechanisms, presumably due to a reduction in the peripheral excitatory input to the motor neurone pool.The study of task failure affords a strategy to identify the functional significance of physiological adjustments that occur during fatiguing contractions. The rate-limiting adjustments can be identified by comparing the performance of one group of subjects on two similar tasks, two groups of subjects sustaining the same task, and one group of subjects performing the same task before and after an intervention. Although relatively few studies have adopted this approach, it appears capable of relating the specificity of the impairments that occur during fatiguing contractions to many of the tasks performed during activities of daily living. These studies underscore the significant contributions of synergist, antagonist and postural muscle activity in limiting the performance of many different physical activities.

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Emg during fatigue. Can an emg make you tired. Emg too early. Will emg.