

Surface EMG as a Fatigue Indicator During FES-induced Isometric Muscle Contractions

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Summary: The electromyogram (EMG) signal has potential as an indicator of stimulated muscle fatigue in applications of functional electrical stimulation (FES). In particular, it could be used to detect near lower limb collapse due to the associated muscle fatigue in FES-aided standing systems and thereby prevent falling. Surface EMG measurement, however, is hampered by stimulation artifact during FES. Modified surface stimulation and EMG detection equipment were designed and built to minimize this artifact and to permit detection of the electrical signal generated by the muscle during contraction. Artifact reduction techniques included shorting stimulator output leads between stimulus pulses and limiting and blanking slew rate in the EMG processing stage. Isometric fatigue experiments were performed by stimulating the quadriceps muscle of 20 able-bodied (a total of 125 trials) and three spinal cord injured (18 trials) subjects. Fatigue-tracking performance indicators were derived from the root-mean-square (RMS) of the EMG amplitude and from the median frequency (MF) of the EMG power spectral content. The results demonstrate that reliable fatigue tracking indicators for practical FES applications will be difficult to obtain, but that amplitude-based measures in spinal cord injured subjects show promise. © 1997 Elsevier Science Ltd. All rights reserved.

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INTRODUCTION

The electromyogram (EMG) signal is the manifestation of the electrical activity produced by actively contracting motor units. EMG monitoring is widely used in biomechanics and movement control

research to determine how the central nervous system (CNS) controls muscular contraction to produce motion, and in clinical practice to diagnose the state of neural or muscle malfunction. More recently, the EMG signal evoked by electrical stimulation has been used to provide information on muscle performance with possible application in functional electrical stimulation (FES) systems which restore function to paralysed limbs by artificial activation of motor neurons and the muscles they innervate.

One of the simplest applications of FES is for paraplegic standing. Individuals with complete paralysis of the lower limbs due to spinal cord injury

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(SCI) can stand by hyper-extending the hips to assume a 'C posture', stimulating the quadriceps muscles to lock the knee joints in extension, and using a hand support such as a walker or parallel bars to stabilize overall body motion caused by free motion at the ankles and trunk. FES-aided standing has been hypothesized to result in significant medical and psychological benefits to individuals with SCI. Yet, without sensory feedback from the paralyzed muscles, the user does not have a sense of the fatigue state of his or her muscles and may not be able to sit before contractile failure occurs. This is of particular importance because muscles fatigue far more rapidly when contracted via electrical stimulation than when voluntarily contracted. The mechanisms for premature fatigue are not well understood, but they are believed to be caused by the inverse size-order recruitment of axons resulting from activation with external stimulation^{9,16,24}, and the unnaturally high rate of motor unit activation and accompanying rapid transition to ischemic conditions that result from the synchronous activation of axons with FES as contrasted with the slower, asynchronous activation that occurs when a motor unit is excited by the CNS^{3,5,26}.

It is tempting to use EMG as an indicator of muscle state because it can be obtained non-invasively and it reflects the contractile activity of the underlying muscle. Our interest has thus been in developing methods for measuring surface EMG during surface electrical stimulation of muscle, and understanding its utility as an indicator or predictor of muscle fatigue. Our ultimate goal is to use this information to design a fatigue monitor for paraplegics standing with FES.

Our investigation had two components. First, we designed a stimulator and EMG processing equipment to minimize the stimulus artifact and extract clean EMG signals. Because EMG is recorded from the muscles being stimulated and because surface stimulation currents are orders of magnitude larger than those produced by motor unit action potentials, the stimulation pulse artifact normally overwhelms the EMG signal at the recording electrode during the pulse.

Second, we conducted experiments in several able-bodied and a few SCI human subjects where processed EMG was compared to muscle force during isometric contractions of the quadriceps muscle. The objective was to determine if either amplitude or frequency properties of EMG were correlated to the decline in force as the stimulated muscle

fatigued, because it is well known that both the amplitude and frequency content of the EMG change with fatigue during voluntary and electrically evoked contractions^{4,18}.

Others have studied the properties of EMG evoked through electrical stimulation and its applications to FES. A comprehensive overview of properties, measurement methods and applications of electrically evoked EMG is provided in the review by Merletti et al.¹⁸, which supplements the tutorial material on CNS-evoked EMG in the book by Basmajian and DeLuca⁴. Graupe¹⁰ and Graupe and Kohn¹¹ have used EMG in FES-aided gait applications both as a command input signal where above-lesion EMG was detected and processed to provide inputs to the stepping controller, and as a stimulated muscle state detector where amplitude changes in EMG appeared to be correlated to fatigue. Solomonow et al.^{28,29} examined the utility of EMG as a feedback signal for closed-loop FES applications and found an approximately linear relation between EMG mean absolute value and stimulated muscle force when using nerve cuff stimulation and intramuscular wire EMG electrodes in acute animal model experiments. The linear relation was thought to result from the size-order recruitment stimulation scheme made possible through a special, high-frequency block method. Mizrahi et al.²⁰ found an approximately exponential relationship between EMG peak-to-peak amplitude and stimulated muscle force in experiments where constant amplitude stimulation was applied to the quadriceps muscle of paraplegic subjects. As with our study, their goal was to determine whether a consistent EMG-force relation is present during stimulated muscle fatigue.

One of the major obstacles to the use of EMG as a feedback signal in FES applications is obtaining stable, artifact-free recordings. When stimulating with surface electrodes and measuring the resulting EMG from the stimulated muscle, the stimulus artifact is generally one or more orders of magnitude larger than the EMG signal itself, and its effects can linger if the EMG amplifier saturates. Special stimulation and recording apparatus must therefore be used to achieve effective artifact reduction.

A variety of methods for reducing stimulus artifact have been described in the literature. As reviewed in McGill et al.¹⁵, the strength of the artifact is influenced by the relative placement of stimulating and recording electrodes, the location of the recording ground electrode, the dynamic charac-

teristics of the EMG amplifier input stage, and the properties of the stimulator output stage. McGill et al.¹⁵ and Kornfield et al.¹⁴ recommend aligning the EMG electrode along equipotential lines from the stimulus. Placing a large ground electrode between stimulus and recording electrode helps^{14,15,21}, as does the use of biphasic stimulation pulses²¹. A common approach for eliminating amplifier saturation is to blank the EMG during the stimulus artifact using a sample-and-hold stage^{1,8,25}, a process that can be improved upon if an estimate of the mean value of the EMG is substituted into the artifact period¹². Alternatively, in software the pure artifact can be estimated and subtracted from the recorded signal¹⁵. A circuit that improves upon simple blanking is presented by Minzly et al.¹⁹, where the input stage of the EMG amplifier is briefly switched from the electrodes to ground via CMOS switches upon detection of a stimulus pulse.

If implanted stimulators are used, the resulting artifact is smaller and simpler to remove. In an animal model, Solomonow et al.²⁷ were able to derive acceptable EMG recordings using nerve cuff stimulating electrodes and intramuscular fine wire EMG electrodes. The detected artifact was reduced by applying an eight-pole low-pass Chebyshev filter to the EMG signal with a cut-off frequency of 450 Hz. Simple filtering also appears to be effective for surface stimulation recording of EMG if a voltage-controlled stimulator output stage is used, because the low output impedance of the stimulator effectively shorts the stimulating electrodes between stimulus pulses, which in turn prevents long artifact transients due to charge build-up in the tissue capacitance^{13,15}. Graupe¹⁰ was able to reduce artifacts using a voltage-controlled stimulator with EMG signal blanking during the pulse, followed by a 300 Hz, low-pass filter. However, it is advantageous to use a current-controlled stimulator for FES applications because the muscle activation response tends to be more reliable.

METHODS

Stimulator

The stimulator and EMG detector were based upon the design of Knaflitz and Merletti¹³, who improved considerably the hybrid current/voltage-controlled stimulator proposed by Del Pozo and Delgado²². Our stimulator is current-controlled with a transformer-coupled output to provide isolation

and output shorting between pulses to reduce artifact time (Figure 1). Low-power, voltage-controlled stimulus pulses were converted to high-power current-controlled pulses in a voltage-to-current amplifier (V-I) stage and then turned into high-voltage (yet still current-controlled) pulses through a step-up transformer before being passed to the stimulation electrodes. Between stimulus pulses, the electrodes were shorted together using an isolated, high-voltage, solid-state relay (International Rectifier model PVA3354). The short started on the trailing edge of the stimulus pulse (plus the 100 μ s switching time for the relay) and ended just before the next pulse, well beyond the end of the evoked EMG. The stimulation pulses were fixed at 300 μ s in width with an interpulse-interval of 33 ms. Stimulation intensity was controlled by modulating pulse amplitude within the range of 0 to 150 mA.

EMG Electrode and Amplifier

Figure 2 contains a block diagram of the EMG amplifier. The key EMG processing stages are: pickup, buffering, differential amplification, slew rate limiting, high-pass filtering, variable gain amplification, blanking, low-pass filtering and adjustable amplification. In the figure, the number within each block is the part number of the integrated circuit that carried out the function of the block. A custom EMG electrode was built with two contacts formed from 1 mm \times 1 mm \times 1 cm-long silver bars which were placed 1 cm apart. The contacts were connected to a pair of unity gain buffer amplifiers (potted in the same module as the contacts) so that the EMG cable lead could be driven with a low-output impedance source. The electrode module had no

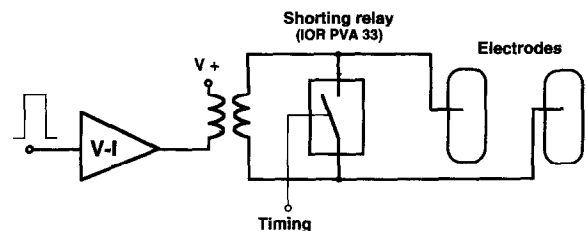


FIG. 1. Block diagram of stimulator used to suppress the stimulus artifact. Low-power voltage controlled pulses with the appropriate timing are converted to high-power current controlled pulses in the V-I stage. A step-up transformer converts the pulses to low current, high voltage for delivery to the surface stimulation electrodes. Between stimulus pulses, the output is shorted using an isolated, high-voltage relay. Shorting the output reduces the stimulus artifact.

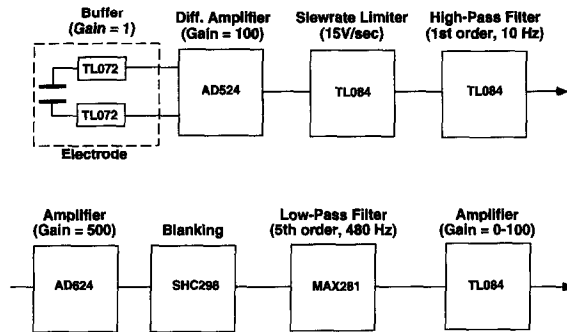


FIG. 2. Block diagram of the EMG amplifier. The number within each block refers to the integrated circuit part number which forms the basis for the circuit in that block. The components inside the block labelled 'Electrode' are mounted in a small epoxy block with integral pickup contacts which is separated from the remainder of the circuit by a 4 ft-long electrode cable. The final output of the amplifier is sampled at 2000 Hz by the computer.

amplification or filtering components which could ring or saturate in response to the stimulus artifact.

The signal was subsequently passed through a differential amplifier with a gain of 100 and then to a slew rate limiter which clamped the slope of fast rising signals to 15 V/s, preventing a fast-rising artifact from either causing ringing in the following high-pass filter or saturating the adjustable gain amplifier. The single pole high-pass filter had a cut-off of 10 Hz to eliminate low-frequency drift in the signal. The amplifier could be set to a gain of either 1000 or 500, although the lower gain was used most often.

A blanking stage based on a sample-and-hold circuit was triggered by timing signals generated by the stimulator to hold the EMG signal at a fixed level during the stimulus artifact. The position and length of the unblank window could be adjusted manually via front panel controls to exclude the artifact and include the desired EMG M wave. During stimulation, this resulted in an approximately 50 ms-long period to capture the M wave with minimal artifact. A five-pole Bessel low-pass filter followed the blanking stage, based upon an integrated four-pole switched capacitor filter block. The cut-off frequency was set to 480 Hz. The final analogue stage was an amplifier whose gain could be adjusted between 1 and 100. The resulting signal was sampled at 2000 Hz with a 12-bit A/D converter.

Knee Torque Measurements

To measure isometric quadriceps torque about the knee joint, subjects were seated on an elevated chair with the thigh immobilized by a wide strap. The ankle was fixed to a load cell with a range of ± 100 lb (Model LCD-100, Omega Engineering, Stamford, CT) underneath the chair and positioned so that the knee joint was held in approximately 90° of flexion. The SCI subjects were tested in their own wheelchairs, which were rolled onto a small platform allowing the ankle to be fixed to the same load cell with the foot clearing the ground. In both cases, the load cell signal was low-pass filtered with a cut-off frequency of 500 Hz, amplified and sampled at 2000 Hz with a 12-bit A/D converter.

Electrode Placement

Carbon rubber, 3 in. \times 5 in., self-stick stimulation electrodes (Model 86905250, Empi Co., St Paul, MN, U.S.A.) were placed in the standard position on the anterior surface of the thigh to elicit contractions of the quadriceps muscle². The EMG electrode was placed between and at least 2 in. from the two stimulating electrodes and oriented parallel to the muscle fibre. No conductive gel was required. For some tests (see description of the protocol below), the EMG electrode was placed distal to the two stimulation electrodes and near the patella in an attempt to maximize the delay time between the stimulus artifact and the M wave. The ground reference for the EMG electrode (a carbon rubber strip with conductive gel) was attached with velcro straps to the contralateral leg below the knee. Several locations for the ground were tried, including the ipsilateral leg below the knee and various versions of grounding shields surrounding the EMG electrode.

Subjects

Stimulation experiments were performed on the quadriceps muscles of 20 able-bodied and three spinal cord injured subjects to determine whether EMG could be used to predict isometric muscle fatigue. All SCI subjects had been participating in a regular programme of FES muscle strengthening as part of a related research programme in FES-aided standing and gait. The strengthening regime involved on-off stimulation of the quadriceps for approximately 1 h per day, five days per week. SCI

subject no. one was 3 yr post-injury, injury level T6 and had been in the strengthening programme for 15 months. SCI subject no. two was 25 yr post-injury, injury level C5/6 and had been in the programme 18 months. SCI subject no. three was 4 yr post-injury, injury level T10 and had been in the programme eight months.

Protocol

After an explanation and demonstration of the stimulator and experimental protocol, each subject was given 5–10 min to familiarize him or herself with the sensation of electrical muscle stimulation. Most able-bodied subjects found the stimulation uncomfortable, but not painful. The stimulation electrodes were then repositioned such that the highest contractile knee torque was obtained for a stimulation level just above threshold, a position which was assumed to be closest to the motor point of the muscle.

Each able-bodied subject was requested to increase the stimulation current manually to elicit a contraction while the leg was isometrically constrained to find his or her maximum comfort level. This setting was named the 'high' level of stimulation. The threshold current for just measurable contractile knee torque was named a 'low' level of stimulation, and the average of the high and low settings was named the 'medium' level. The high level of stimulation for the SCI subjects was set at 150 mA, the maximum for the stimulator, while the low and medium levels were determined as for able-bodied subjects.

Each fatigue experiment consisted of four 60-s stimulation trials with 5–15 min of rest between each trial. With the subject seated and his or her leg restrained by a cuff about the ankle, stimulation amplitude was ramped from zero to the trial level (low, medium or high) over 1 s, and then held constant for the remaining 59 s, thus inducing fatigue. For three of the four trials—one at each of the high, medium and low stimulation levels—the EMG electrode was placed between the stimulation electrodes. For the fourth trial, the EMG electrode was moved to the distal position (described above) and the test run at the medium stimulation level.

For 10 of the 20 able-bodied subjects, a second fatigue experiment was performed on the opposite leg. For one subject, an extra series of five trials was also performed. The data was analysed based

on a total of 125 trials for the able-bodied subjects and 18 trials for the SCI subjects.

Signal Processing and Data Analysis

Every 2 s, a set of eight EMG M-wave epochs were recorded for a total of 30 EMG sets during the 60 s of muscle stimulation. These were spike-triggered averaged to derive one smoothed, representative M wave for each 2 s of stimulation. The force data was also averaged over each 2-s interval. The RMS amplitude of the averaged EMG M wave was found by subtracting the mean, summing the square of each data point, dividing by the number of samples and then taking the square root. The frequency content of the M wave was found by subtracting the mean, applying a Welch smoothing window, zero padding the record to 512 samples, computing the Fast Fourier Transform (FFT) and squaring the resulting magnitude to derive a power spectrum estimate. The half-power median frequency (MF) was found by first summing the power across all frequency points and dividing by two. Next, the power was summed from the left and from the right to converge on the frequency point where both running sums equalled the result from the first step. This was the median frequency point. All processing algorithms were derived from those found in Press et al.²³.

Four performance indices were used to determine whether the EMG amplitude or average frequency content tracked the decline in average knee torque as the muscle fatigued during 60 s of stimulation. Before forming the indices, EMG RMS, EMG MF and the torque were normalized so that over the trial the minimum value for each was 0.0 and the maximum 1.0. After normalizing, the RMS difference between EMG RMS and torque and between EMG MF and torque were computed (for the 30 data points in the trial) and multiplied by 100 to form the TrackRMS and TrackMF indices. A low number for either index would mean that the respective EMG measure closely followed the changes in torque. The intent of the other two indices was to see whether EMG RMS or EMG MF generally had the same trend as torque over time. For these, the slope of EMG RMS and MF and the torque was computed by averaging the second to fifth data points and the last four data points to get initial and final points from which slopes could be calculated. The difference between the slope of the torque and EMG RMS was the

SlopeRMS index, and the difference between the slope of the torque and EMG MF was the SlopeMF index. Small numbers in these indices indicated that the appropriate EMG measure had the same trend as torque, whereas large numbers indicated that the EMG measure went in one direction whilst the torque went in the other.

RESULTS

Artifact Reduction

Figure 3 is an EMG recording taken at a stimulus level too low to elicit a detectable M wave and thus shows artifact alone. The solid line is the artifact when the stimulator was not shorted between pulses whereas the dashed line was with the shorting active. Shorting reduced the artifact time from approximately 25 ms to 7 ms. Figure 4 shows a typical EMG recording as various stages of detector processing were added. In (a) there was no blanking, stimulator shorting or slew rate limiting. In (b) slew rate limiting was added; in (c) stimulator shorting was added; and in (d) blanking was added, resulting in a relatively clean M-wave signal without excessive artifact contamination. Typically, the remaining input-referred artifact was less than 10 mV peak-to-peak. Note that in (d), the displayed M wave is monopolar because its first phase overlapped the residual artifact and was zeroed by the blanking circuit. In other setups with more separation between the artifact and the M wave, the full bipolar M wave was revealed. In all cases, the M wave we used for analysis was artifact-free, even if it meant clipping its first phase as was the case for (d).

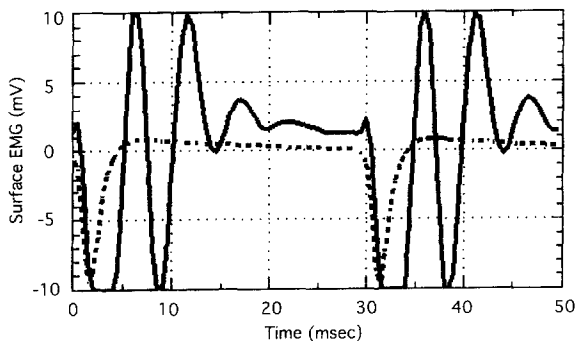


FIG. 3. Signal recorded by the EMG amplifier when only a stimulus artifact was present, because the stimulus was too weak (28 ma) to produce a detectable M wave for this subject. The solid line shows the artifact when the stimulator was not shorted between stimulus pulses, whereas the dotted line shows an artifact greatly reduced in duration when the stimulator shorting circuit was active. Two pulses are shown.

The placement of the EMG electrode had a large effect on the recorded artifact. The proximal placement (between the two stimulation electrodes) resulted in an artifact that was larger in amplitude and longer in duration than the distal placement. Although the M-wave signal strength at the distal site was of reasonable magnitude, the electrode was not recording from the bulk of the muscle fibres. For all ground electrode locations and configurations, there was no significant difference in either noise rejection or artifact reduction.

Fatigue Tracking

Figure 5 illustrates how the M wave changed with time during a typical fatiguing contraction, and also shows a clear delineation between M wave and stimulus artifact. The three superimposed traces of trains of three successive, unblanked M waves were recorded at 0, 15 and 45 s after the start of continuous stimulation. They show that as time increased, the RMS EMG amplitude declined and the shape broadened. The broadening shape correlated with a shift in the frequency spectrum of the M wave, and thus for this record the MF decreased with time. The shape and amplitude of the artifact, the downward plunge of the signal during the first 5 ms, did not change with time.

Figure 6 demonstrates the range of experimental results we obtained. The top three rows are typical trials from three different able-bodied subjects, whereas the bottom row shows a trial from one of the SCI subjects. The plots in the left column show normalized torque and EMG MF versus time, whereas those on the right show normalized torque and RMS EMG for the same trial. The values of the four performance indices for these trials are listed in the caption.

Table 1 and Table 2 display the mean and standard deviation of the four performance indices for the 125 able-bodied subject and 18 SCI subject trials. The values of the indices shown in Figure 6 can be used to relate the numbers visually in these tables to actual performance in typical trials. For the able-bodied subject experiments, the means for both of the slope indices were small, but those for the tracking indicators were not. For the SCI subjects, the mean TrackRMS index was relatively small, as were the two slope indices, but the TrackMF index was not. Also, the standard deviations of the indices were smaller for the SCI subject trials than for the trials with able-bodied subjects

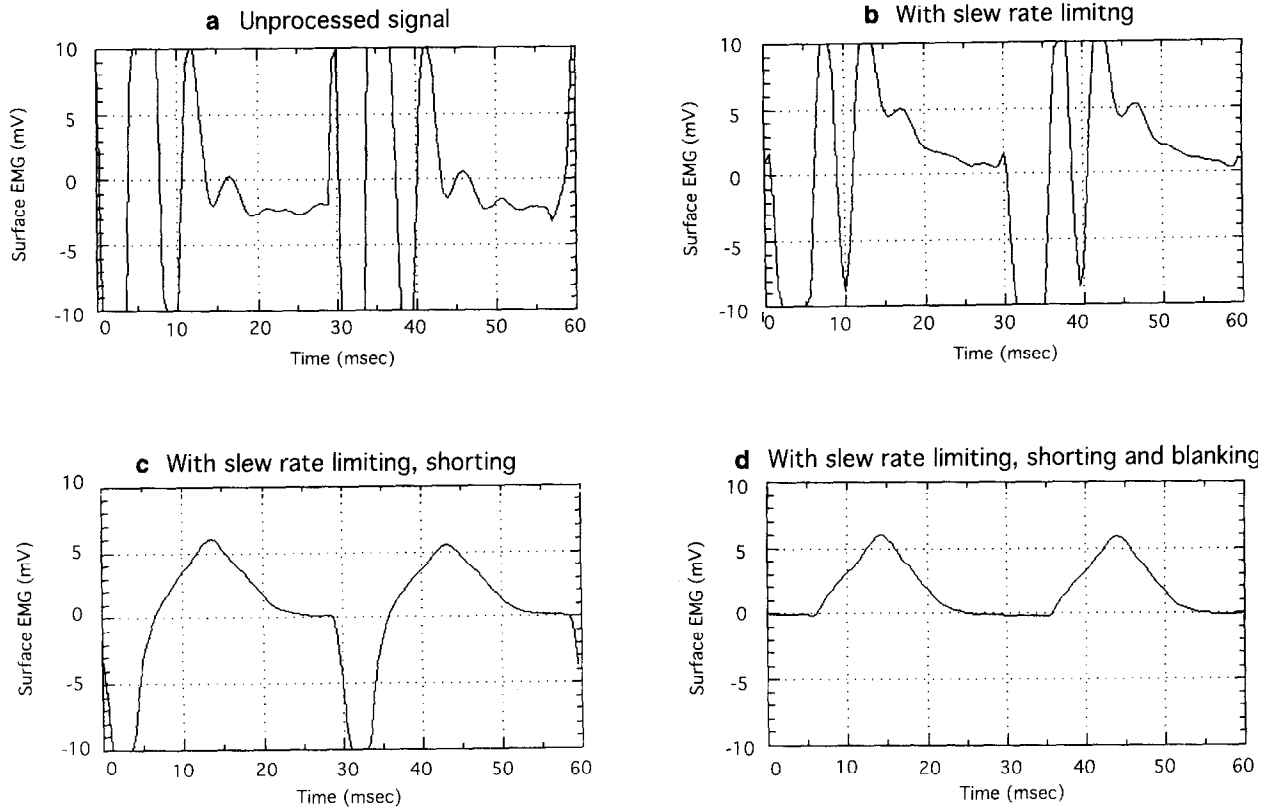


FIG. 4. Typical EMG recording as various stages of the EMG amplifier were activated. **a**, Only amplification and filtering and no stimulator shorting; **b**, adds in slew rate limiting; **c**, adds stimulator shorting; **d**, adds blanking to complete the fully processed signal and shows a clean M-wave signal. For this particular setup, the initial downward phase of the M wave overlapped the tail end of the artifact and was therefore clipped by the blanking circuit.

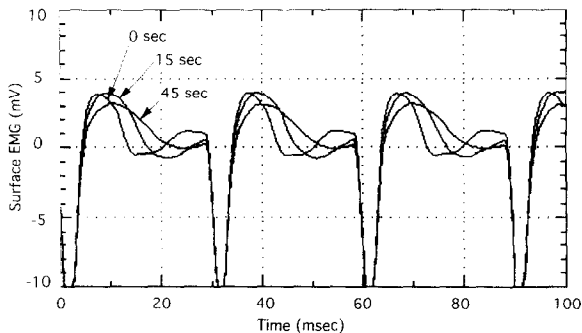


FIG. 5. Overlapped EMG records from a single, 60-s fatigue experiment. The traces were taken at 0, 15 and 45 s into the contraction and each trace shows a train of three successive M waves. As time increases, the M waves decrease in amplitude and broaden in shape. The large downward spike is the unblanked stimulus artifact which does not change with time.

indicating that there was better repeatability in the SCI subjects. By using an F-distribution statistical test to compare the variances between the two populations, it was determined that the variances were significantly higher ($\alpha=0.05$) for the able-bodied subjects for all indices except TrackMF.

DISCUSSION

Fatigue Tracking Experiments

The EMG parameters did indeed change with time, but the performance indicator results showed that for able-bodied subjects in this particular experiment, TrackRMS and TrackMF were unreliable indicators for predicting or tracking fatigue. Furthermore, although the mean values for SlopeRMS and SlopeMF indicators were small (implying good ability to predict torque trends), these results can be misleading. For example, consider the right plot in the second row of Figure 6. Although the slope of the EMG RMS did not track the slope of the torque accurately, the overall slope for both as defined by their endpoints is about the same, resulting in a low SlopeRMS value. Also, the high standard deviation for the slope indices means that there were many with large positive and large negative values. An example can be seen in the top left plot of Figure 6 where the torque rose whereas EMG MF fell during the trial. Furthermore, one should remember

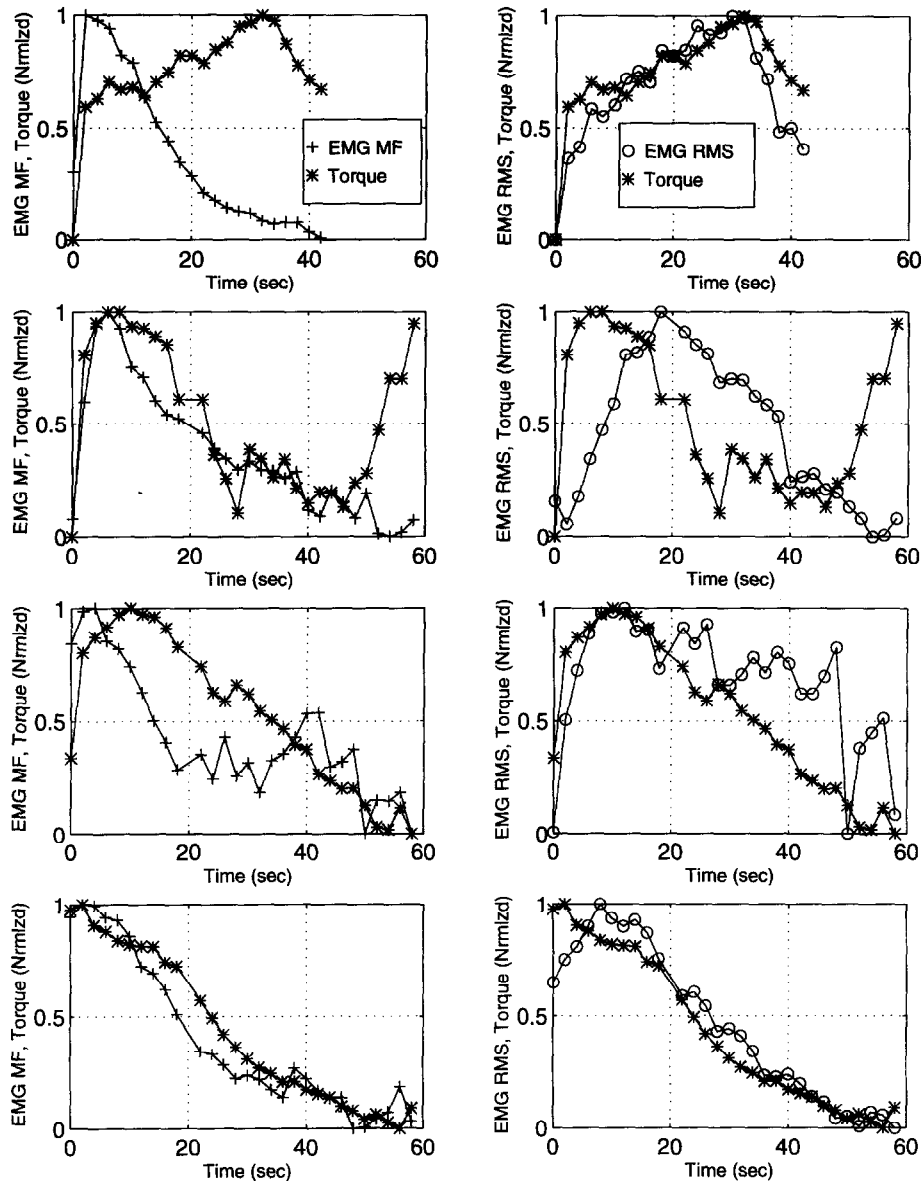


FIG. 6. A sampling of typical results from the fatigue experiments. Each row contains the data from a single trial with the left plot showing EMG median frequency (+ markers) and torque (* markers) as a function of time, and the right showing EMG RMS amplitude (unfilled o markers) and torque as a function of time. The first three rows come from experiments using able-bodied subjects, whereas the bottom row is an experiment from a paraplegic subject. Values of the performance indices (see text) for each trial follow. First row: TrackMF=60.3, TrackRMS=13.5, SlopeMF=45.1, SlopeRMS=2.5. Second row: TrackMF=29.1, TrackRMS=44.7, SlopeMF=20.8, SlopeRMS=-0.5. Third row: TrackMF=28.2, TrackRMS=28.5, SlopeMF=-1.9, SlopeRMS=-14.9. Fourth row: TrackMF=11.0, TrackRMS=11.5, SlopeMF=-0.3, SlopeRMS=-0.7.

that the slope indices cannot be computed in real time because they are based upon knowledge of the endpoint values.

There were several reasons for the higher variability seen in the able-bodied subject data. There may have been contamination due to voluntary contractions, which are difficult to control during stimulation despite the training in how to relax during

stimulation contractions that our subjects received. Although the normalizing and pooling of all trials was deliberate to see if simple indicators had practical application, normalizing can lead to other problems. For example, all records were normalized to range between zero and 1.0 over the 60 s, even if torque and EMG measures hardly changed. For these records, the indices would be very sensitive to small

TABLE 1. Mean and standard deviation of the four fatigue tracking performance indices (defined in the text) pooled over all 125 experiment trials for the 20 able-bodied subjects

	Mean	Std dev
TrackRMS	30.7	12.7
TrackMF	38.3	14.3
SlopeRMS	-6.7	19.5
SlopeMF	-0.4	21.6

TABLE 2. Mean and standard deviation of the four fatigue tracking performance indices (defined in the text) pooled over all 18 trials with the three spinal cord injured subjects

	Mean	Std dev
TrackRMS	14.4	7.6
TrackMF	24.5	11.0
SlopeRMS	-4.7	6.8
SlopeMF	-8.3	13.0

changes in torque or EMG. In addition, normalizing is not a possibility for indices that are to be computed and utilized in real time. Furthermore, for most able-bodied subjects, the torque levels elicited through stimulation were a small percentage of their maximum voluntary contraction even at their highest stimulation level. With more motor units recruited, as we were able to do with the SCI subjects, the EMG fatigue indicators might show less variability. Nevertheless, the ability to predict fatigue at lower contraction levels is still important, because for many SCI subjects the levels of stimulus-induced contraction required for stable FES-standing can be quite low.

For the SCI subjects, the variances of all indices were smaller than those for the able-bodied subjects. At least two factors probably contributed to this: first, the assurance that no voluntary contractions could contaminate the records, and second, the well-trained response to FES of the SCI subjects' muscles given their participation in a long-term stimulated muscle strengthening programme. Furthermore, TrackRMS was a relatively well-behaved indicator of fatigue for the SCI subjects. EMG amplitude information may thus still be useful in practical FES applications to monitor fatigue.

Other EMG measures could be used. For example, Solomonow et al.²⁸ found that a mean absolute value (MAV) measure of EMG amplitude provided

a better fit to force data than RMS or peak value when using implanted stimulation and wire EMG electrodes in an animal model. Mizrahi et al.²⁰ had reasonable success fitting EMG peak-to-peak amplitude to force with an exponential model in fatiguing, isometric-stimulated contractions of the quadriceps muscle in paraplegic subjects. In preliminary experiments, we found no significant difference between RMS, MAV and peak measures, but we did find that the peak measure was more susceptible to noise spikes^{6,7}. Also, it is known that the RMS measure is a somewhat better representation of muscle state than MAV, and does not require one to define the peak^{4,18}. Likewise, mean rather than median frequency could be used as a spectral measure, but others have determined that for sustained voluntary contractions, median frequency is more appropriate for detecting subtle shifts due to fatigue^{17,18}. It remains to be seen, however, whether this is also the case for FES-induced contractions.

EMG Detection

Although shorting the stimulation electrodes and slew rate limiting and blanking in the EMG processing circuit were successful in greatly reducing the stimulus artifact, reliable detection of EMG continued to be a problem. This was particularly the case when the EMG electrode was positioned between the stimulating electrodes and over the midsection of the quadriceps muscle, so that a large number of active muscle fibres could be sampled. In this configuration, the detection site was necessarily close to the stimulation electrodes, and there was often overlap between the residual artifact and the evoked M wave. This may have been exacerbated by the design of our stimulator output stage, because the inductance in the output coupling transformer and the capacitance in the skin and tissue could result in a slowly decaying resonance. A capacitively coupled output stage would eliminate this behaviour.

In the cases where there was overlap, the blanking circuit clipped the initial phase of the M wave (e.g. Figure 4) and may have affected the calculated MF and RMS values. However, even for these cases, both MF and RMS indicators were able to capture changes in M-wave shape and amplitude due to fatigue and could thus be used as a within-subject fatigue indicator. One reason we chose RMS over a peak-to-peak amplitude measure was because the clipped M waves often did not have distinct positive and negative peaks.

One could place the EMG electrode distal to the stimulation site to increase the delay time between the stimulus artifact and M wave, but for muscles such as the quadriceps the electrode would then be sampling very few muscle fibres. There is thus an inherent trade-off between obtaining clean EMG recordings and measuring a signal that is representative of the activity of the whole muscle.

The results do not bring out the preparation and tuning required in each experiment before a clean EMG could be recorded. The EMG electrode had to be positioned properly and adjusted frequently to obtain a strong signal that was free of any motion artifact caused by the muscle changing shape as it contracted under the electrode. Likewise, it took some effort to determine the optimal blanking window that eliminated the remaining artifact without clipping the M wave. Although the time and effort required for setup did not affect the outcome of these experiments, it could present a serious barrier to using EMG as a control signal in practical FES applications.

CONCLUSIONS

The objective of this work was to determine if EMG measures could be used as an indicator of fatigue in electrically stimulated muscle. To detect EMG in the presence of stimulation, it was first necessary to eliminate the stimulus artifact that would otherwise overwhelm the EMG signal when the recording and stimulation electrodes are in close proximity. The shorting stimulator combined with an EMG amplifier that contained integral slew rate limiting and blanking successfully brought the artifact down to a level where EMG M waves could be detected. Nevertheless, clean EMG recordings were difficult to obtain and may thus limit the usefulness of EMG in practical FES applications.

In able-bodied subjects, none of the four performance indices calculated could be considered for use as a reliable fatigue indicator, although it is possible that with further processing to screen out specific trials, some useful information could be garnered. In SCI subjects, the potential is greater for a reliable fatigue indicator based on EMG signals. For these tests, amplitude EMG measures appear to have more promise as indicators of muscle state than frequency-based measures.

Before surface EMG can be considered for practical application in FES-aided standing, further work must be done to develop simple, reliable methods

for obtaining clean M-wave signals. Also, amplitude-based measures with more sophisticated processing to screen out bad data points must be developed. Implanted stimulators and EMG detection equipment may ultimately prove to be more suitable if feedback signals based on reliable measures of muscle state are required for FES applications.

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