Evoked EMG Can Be Used as a Fatigue Sensor of Paralyzed Muscle During Stimulation Patterns Corresponding to Neural Prosthesis Operation

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Abstract—*The focus of this paper is a new look at fatigue phenomena, and characterization of the relationship between the variation of the Evoked EMG (EEMG) and the variation of muscle force during prolonged electrical stimulation of paralyzed muscle using stimulation patterns corresponding to neural prosthesis operation. Experiments were conducted on paraplegic subjects with the lower limb vastus-lateralis muscle stimulated via percutaneous intramuscular electrodes under isometric conditions. An "artifact balancing" technique was used for recording the EEMG and overcoming the stimulus artifact. To characterize the changes of the muscle behavior during prolonged periodic ramp-hold stimulation, we fit hyperbolic functions to the EEMG and the torque. It is observed that functions' dead-zone regions increase with time, indicating decreasing excitability and contractibility. The changes in the dead-zone region of the torque responses closely track the changes in the corresponding EEMG responses with a delay. In addition, the time constant of the EEMG responses closely matches the changes in the torque responses. In this work, we also developed a calibration procedure that enables us to predict the torque from EEMG during prolonged ramp-hold stimulation with an average mean prediction error of about* 6%.

Keywords: Functional Neuromuscular Stimulation, Muscle Fatigue, Time-Frequency Representation.

1. INTRODUCTION

An understanding of fatigue phenomena in electrically stimulated muscle is still illusive and an important pre-requisite for the development of improved "neural prostheses" using Functional Neuromuscular Stimulation (FNS). Many investigators have attempted to quantify the muscle fatigue during voluntary and electrically elicited contraction [1]. Some have used the spectral and time-domain variables of EMG signal as indices of muscle fatigue [2-5]. Other researchers have modeled the decrease of force or torque over time by an exponential [6] or hyperbolic tangent [7], and used regression-based parameters as muscle fatigue indices. In spite of the body of work in the literature regarding fatigue phenomena during voluntary and electrically induced muscle contraction, the problem of fatigue detection in real-time has not been studied. The time-domain and spectral variables of the EEMG exhibit some variation during sustained electrical stimulation, but these variations are not repeatable. The relationships between changes in the variables of the EEMG and measured decrease of muscle force have not been adequately characterized.

In this work, we present new aspects of the fatigue process when the lower limb vastus-lateralis muscle in *paraplegic* subjects is stimulated via percutaneous *intramuscular* electrodes to provide selective stimulation. We will show that the EEMG signals contain substantial information about the fatigue condition of the muscle, if properly collected, processed and analyzed.

2. EXPERIMENTAL PROCEDURE

Experiments were conducted on two complete-level-T7 spinal cord injury paraplegics. Percutaneous intramuscular electrodes were implanted near the motor points of the major lower limbs as described in [8] During the experiments reported here, only the lower limb vastus lateralis muscle was stimulated, by activating the corresponding intramuscular electrode. The muscle was stimulated using pulse-width modulation at a constant frequency (20 Hz) and constant amplitude (10 mA), under isometric conditions. The knee of the test leg was fixed securely in 30° of flexion (where full extension is 0°).

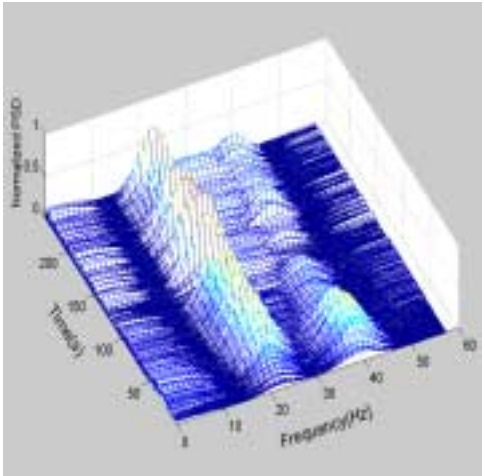
Isometric knee torque was measured using a Cybex II dynamometer. Measured values of the knee torque were low-pass filtered (cut-off frequency 100 Hz), and sampled at 1200 Hz. EEMG data were collected by a differential amplifier with a common mode rejection ratio of 120 dB and bandwidth of 250 kHz and then sampled at a rate of 1200 Hz.

3. EXPONENTIAL TIME-FREQUENCY DISTRIBUTION OF EEMG DURING PROLONGED ELECTRICAL STIMULATION

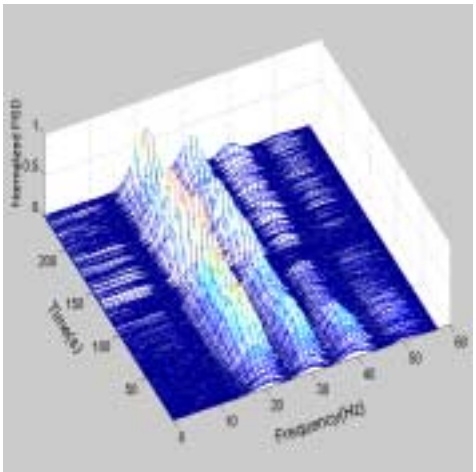
Analysis using the Exponential Distribution (ED) of Choi and Williams [9] overcomes many of the limitations of the Short-Time Fourier Transform, providing high resolution in both time and frequency while reducing the cross component amplitudes [9]. The ED for a discrete-time signal is defined as follows [9]:

$$ED(n,\omega) = 2\sum_{\tau=-\infty}^{\infty} e^{-j2\omega\tau}\left[\sum_{\mu=-\infty}^{\infty}\sqrt{\frac{\sigma}{4\pi\tau^2}}\cdot\times \exp\left(-\frac{\sigma(\mu-n)^2}{4\tau^2}\right)\cdot x(\mu+\tau)\cdot x^*(\mu-\tau)\right]$$

where $*$ represents the complex conjugate. The cross terms can be controlled by the parameter σ. A small value of σ reduces the effects of cross terms. On the other hand, the frequency resolution also depends on σ.



(a)



(b)

Fig. 1: Exponential time-frequency representation of the EEMG during sustained electrical stimulation of paralyzed muscle where the value of σ has been set to 0.1 (a) and 1.0 (b).

A large value of σ will leads to have a sharp autoterm resolution. Thus, the parameter σ trades off autoterm resolution for cross-term suppression.

Fig. 1(a) shows *the Exponential time-frequency distribution* (ED) of the EEMG during prolonged constant electrical stimulation of paralyzed muscle. where a 120-point rectangular window has been used and the value of σ has been set to 0.1. The most interesting result is that the power of EEMG is concentrated in the frequency of the stimulation signal and its first harmonic. In addition, as muscle becomes more fatigued, frequency components appear which are not the harmonics of stimulation frequency. **Fig. 1(b)** shows the same information as in **Fig. 1(a)** when the parameter σ has the value 1.0. It is clearly observed that a peak component appears at $f = (f_1 + f_2)/2$ due to interference of the signal components with increasing values of σ. As expected, the distribution oscillates in the direction of the time axis. The amplitude of oscillation depends on σ and the signal structure.

4. Fatigue Estimation During Prolonged Electrical Stimulation

Fig. 2(a) and **(b)** shows the measured EEMG and the knee torque during prolonged constant electrical stimulation of paralyzed muscle. We will associate *excitation fatigue* with changes in the envelope of the measured EEMG, and *contraction fatigue* with changes in the measured torque. The greatest discrepancy between the measured EEMG and measured torque is during potentiation. The measured EEMG changes track the knee torque reduction well during fatigued and maximally fatigued conditions. The variability of the EEMG is more than that of the torque, and that the contraction process behaves as low pass filtering of the EEMG.

In order to extract the DC gain of the EEMG, and of the torque, and thus to observe the relationship between *excitation* and *contraction* fatigue, we filter both signals by a *first order elliptic lowpass filter* [10] with a cutoff frequency of $\omega_c = 1$ Hz, and then smooth the filtered values by a *moving average filter* with order 10. **Fig. 2(b)** shows the processed values of EEMG and knee torque. Note that the variations of DC gain of the EEMG match the torque.

The *correlation coefficient* was used to measure the linear correlation between the excitation fatigue (measured EEMG) and contraction fatigue (measured torque). It was found that the average of the *correlation coefficient* for different day experiments has the value 0.9610.

The results show that the processed EEMG would be able to track the muscle force with mean prediction error of 6%.

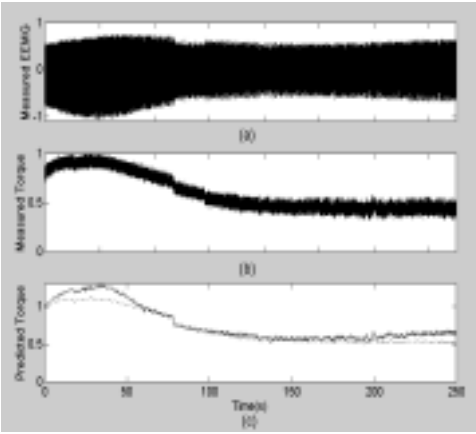


Fig 2. Torque prediction during prolonged constant electrical stimulation of paralyzed muscle using the measured EEMG: measured Evoked EMG (a), measured torque (b), predicted torque (c).

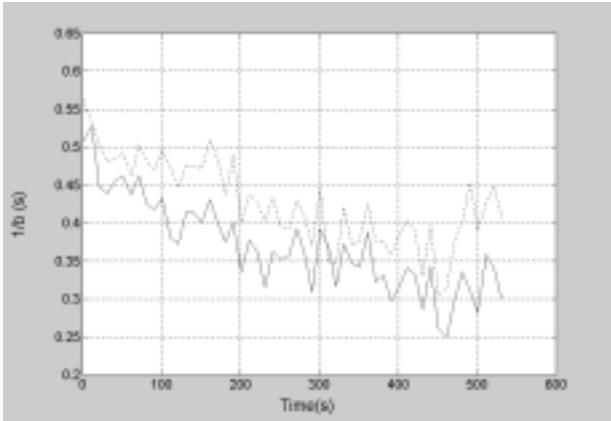
5. Prediction of Incipient Fatigue During Ramp-Hold Stimulation

Fig. 4(a) shows the measured knee torque and the lower envelope of EEMG during a typical periodic ramp-and-hold stimulation (period=10 s, ramp=4 s, hold=6 s). It is observed that that the EEMG does not track the long time variations of the torque during this ramp-hold stimulation pattern. In order to characterize changes of the muscle behavior during prolonged periodic ramp-hold electrical stimulation, we fit hyperbolic functions with four parameters to the EEMG and the knee torque which are measured during each period of ramp-hold stimulation.

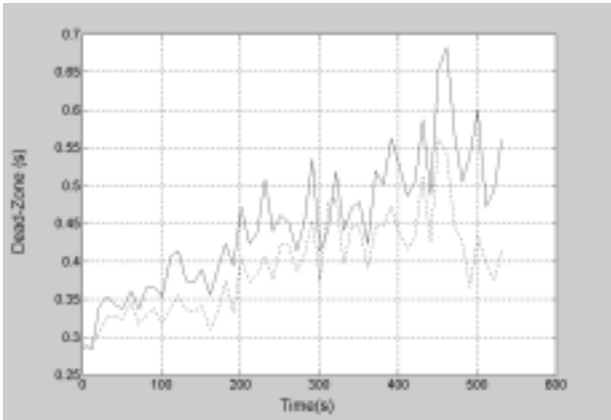
$$f_n(t) = a_n \,\mathbf{tanh}(b_n t - c_n) + d_n$$

where $f_n(t)$ is the *n*th muscle response (EEMG or knee torque) to the ramp-hold stimulation. The reciprocal of b_n represents the time constant, the product $a_n \cdot b_n$ denotes the slope and describes how fast the torque or EEMG increases. The c_n represents the dead-zone region and explains the level of excitability and contractibility. Finally d_n term is used to fit any offset in the output. For estimating the parameters, we used the Levenberg-Marquard method [11].

Fig. 3(a)-(c) show the evolution of the dead-zone region, time-constant, and slope, respectively. We observe that the c_n increases with time, indicating increasing the dead-zone regions, and therefore, decreasing the excitability and contractibility. Changes in the dead-zone region of the torque responses appear to closely track the changes in the EEMG, but with a delay. It is observed that the time constant of the EEMG responses closely track the changes in the measured torque.



(a)



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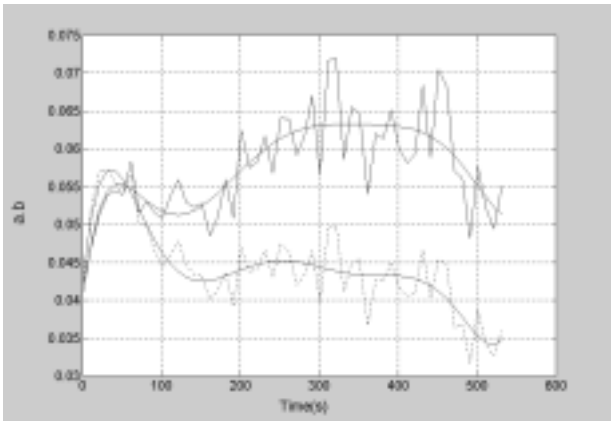

(c)

Fig. 3. Parameters of the fitted curves to the measured EEMG (solid) and to the knee torque (dot): (a) time constant; (b) dead-zone; (c) slope.

The slope of the ramp response does not change during prolonged ramp-hold stimulation. These observations suggest that the torque decreases proportionally to decreases in contractibility. To estimate the torque response using the EEMG, we define the calibration factor

$$\alpha(n) = \frac{c(1)}{c(n)} \qquad n = 1,\ 2,\ \ldots$$

where $c(n)$ is the nth dead-zone region of the ramp response of the excitation process (EEMG). The multiplication of the filtered values of EEMG by the calibration factor is an estimate of the torque.

The effectiveness of this calibration procedure is illustrated in **Fig. 4(a)** and **(b)** which shows the torque prediction during 500 seconds without calibration and with calibration, respectively. The improvement provided by this calibration is quite evident. In this figure, the calibration factor is computed from fitted curves to EEMG, and then multiplied by the filtered values of EEMG.

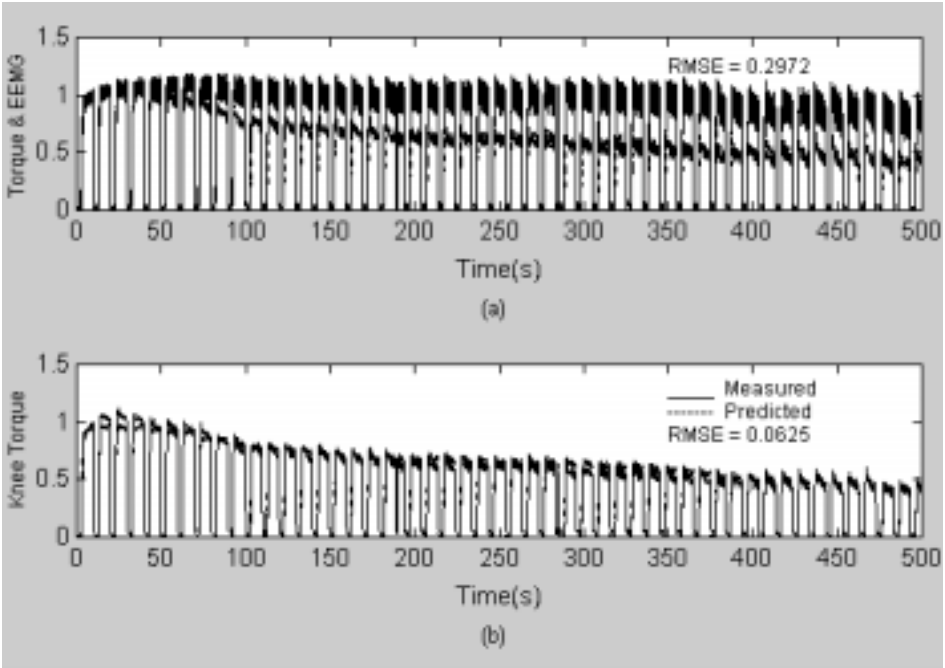

Fig. 4. (a) The measured of EEMG and the knee torque during prolonged ramp-and-hold stimulation of paralyzed muscle. (b) The measured (solid) and predicted (dot) knee torque using the calibration procedure.

6. DISCUSSION

In this paper, we have investigated the fatigue process during prolonged electrical stimulation of paralyzed muscle. There is a high correlation between excitation fatigue and contraction fatigue. Moreover, the muscle torque can be predicted from measured values of EEMG during sustained constant stimulation input. However the EEMG does not match the long time variations of the torque during ramp-hold stimulation, and the EEMG does not decrease as much as torque during fatigued state. One possible suggestion for this observation is that the recovery time of excitation process and contraction process is not the same.

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