Information on Ankle Angle from Intramuscular EMG Signals during Development of Muscle Fatigue in an Open-Loop Functional Electrical Stimulation System in Rats

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Abstract. Functional Electrical Stimulation (FES) is one method available for rehabilitation of spinal cord injured subjects. Although FES is used in the clinic today, reliable and robust feedback for a closed-loop system is limited.

The objective was to examine if intramuscular electromyographic (iEMG) recordings (of tibialis anterior and gastrocnemius medialis) can provide reliable information of functional movement (i.e. ankle angle) during development of fatigue.

Four longitudinal intrafascicular electrodes (LIFEs) were implanted in two fascicles of the sciatic nerve in three adult Sprague-Dawley rats. Open-loop FES was applied to produce rhythmic ankle movement. The FES stimulation pulse widths and amplitudes were determined for the individual rats based on the strength duration curve. Each frequency (30, 40, 50, 60 and 70 Hz) was applied to perform 100 step cycles followed by a 15 min rest period. Kinematic information on the ankle angle and iEMG were recorded simultaneously.

The results showed that the ankle angle and the iEMG amplitude decreased when the muscles fatigued. A correlation between the ankle angle and iEMG was present, which indicates that iEMG information can be used as feedback for a closed-loop system. The correlation was higher at higher stimulation frequencies (>0.76 at stimulation frequencies above 40 Hz).

1 Introduction

Injury to the spinal cord may cause permanent loss of voluntary motor function and sensation below the level of the lesion. Functional electrical stimulation (FES)

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is a technique that has been used for many years for the rehabilitation of subjects with spinal cord injury. [1], [2], [3]

The aim of the FES is to electrically activate the paralyzed muscles in a controlled way to restore motor function. FES can be applied to the subject using an open-loop (feedforward) or a closed-loop (feedback) control strategy. [1], [2], [3]

FES is commonly applied in the clinic in an open-loop mode that operates with fixed stimulation parameters. A clear advantage is that the paradigm is simple and easy to use. However, prolonged stimulation leads to muscle fatigue. An appropriate closed-loop stimulation strategy could alleviate this problem. However, closed-loop FES systems are dependent on feedback from the part of the body that is controlled, and the availability of sensors and signals to provide a reliable feedback signal from the controlled limb or organ is therefore essential. [1], [2], [3]

One source of feedback can be achieved by recording kinematic data that provides information on the joint angles. Kinematics would reveal when fatigue occurs. [1], [4] However, the recording of kinematic information requires specialized equipment and is a highly time consuming procedure, which is not suitable for daily use in the clinic or at home. Therefore it will be important to have access to another source of information. Information on movement and muscle fatigue can also be obtained through electromyography (EMG) recordings. The use of EMG recordings would be relatively easy to implement in the clinic or in a portable system since it is cheap, quick to setup and is used routinely today. Surface EMG has the advantage of being easy to record but suffers from cross talk and the daily need to don and doff the electrodes. An alternative signal source is the use of intramuscular EMG (iEMG), which is a more invasive technique. This may help overcome some of the drawback associated with surface EMG. Previous studies show that it is possible to extract information from the iEMG related to the force during movement [5].

There is today limited knowledge on whether information on joint angles may be extracted from iEMG during normal movement and how muscle fatigue may influence this.

The objective of this study was therefore to examine if information extracted from iEMG recordings can provide reliable information on a functional movement and during development of muscle fatigue in a rat model.

To investigate the objective, FES using longitudinal intrafascicular electrodes (LIFEs) was used to produce a cyclic movement of the hindlimb of the rat while recording iEMG. The FES was applied in an open-loop mode to induce muscle fatigue over time. Kinematic data was also recorded as a reliable measure of the movement.

2 Methods

Data was obtained from three adult healthy male Sprague-Dawley rats (294-615 g). The experimental procedures were approved by Florida International University Institutional Animal Care and Use Committee.

2.1 Animal Preparation

After induction of anesthesia with isoflurane gas (5 %), a single injection of Sodium Pentobarbital (40 mg/kg ip) was given. Anesthesia was maintained with isoflurane (0.5-2.0 %), throughout the experiment. The level of anesthesia was assessed with toe pinch, and observation of eye blink and the respiration rate. To prevent dehydration regular subcutaneous injections of isotonic saline in the dorsal cavity were administered.

The left sciatic nerve was exposed and four single channel LIFEs were inserted into the fascicles innervating the Tibialis Anterior (TA) and Gastrocnemius Medialis (GM) muscles. The TA and GM are the main muscles involved in the movement of the ankle and can be activated from one nerve. To verify that the electrodes were placed correctly inside the fascicles, electrical stimulation was applied while observing TA and GM muscle twitch. The LIFEs were sutured to the epineurium and the incision was closed.

Table 1 Applied LIFE stimulation parameters

	Amplitude (μA)	Pulse width (μs)
Rat 1: GM	50	30
Rat 1: TA	30	30
Rat 2: GM	50	60
Rat 2: TA	250	100
Rat 3: GM	30	40
Rat 3: TA	20	30

2.2 Experimental Setup and Data Acquisition

The rat was placed in a prone position on an elevated platform so that the hindlimbs were hanging freely.

3-D kinematic data was recorded by a Peak Motus System (Peak Performance Technologies, Inc. Centennial, CO) by placing cone shaped three-dimensional reflective markers on the hip, knee, ankle and toe. The system included two infrared cameras focused on the rat at an oblique angle of approximately 45° each. The kinematic data was sampled at 60 Hz.

To record differential iEMG, two stainless steel fine wire electrodes were inserted with an average interelectrode distance of 3.5 mm in the TA and GM muscles. The reference electrode was placed under the skin at the back of the rat. The iEMG data was amplified (A-M systems Model 1700, gain = 100) filtered (band pass: filter 100 Hz -10 kHz, notch filter at 60 Hz), sampled (10 kHz, NI USB-6259, National Instruments, USA) and saved in a PC using a custom LabView routine.

To determine the stimulation pulse width and amplitude a strength duration curve was first established by consecutively stimulating the four implanted LIFEs with different pulse widths (30, 40, 50, 60, 70, 80, 90, 100, 150, 200, 250 and 300 μs)

while increasing the stimulation pulse amplitudes until a muscle twitch was seen. A pulse width and 1.5 times the amplitude at rheobase were selected for stimulating the fascicles (see Table 1).

To determine the stimulation frequencies the muscle contraction was visually observed. The stimulation frequency was chosen such that it provided a fused contraction, which was later confirmed from kinematic data.

The open-loop stimulation was applied to the fascicles innervating TA and GM muscles alternately to produce a rhythmic movement. One hundred step cycles were performed at each stimulation frequency (30, 40, 50, 60 and 70 Hz) followed by a 15 min period of rest. Kinematic data from ankle movement and corresponding iEMG were recorded simultaneously.

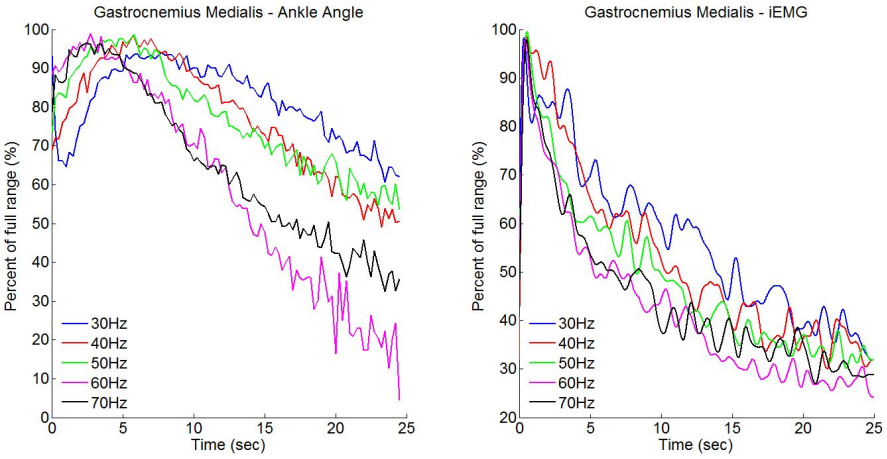

Fig. 1 Shows the range of movement of the ankle angle and iEMG amplitude envelope over time for the GM muscle. The maximal range of ankle angle and iEMG amplitude envelope was normalized to 100 %. A gradual decrease in the ankle angle and iEMG was observed for the GM over time.

2.3 Data Analysis

The kinematic markers were identified using the Peak Motus software. The data were digitized and filtered with a fourth-order band-pass Butterworth filter [6]. To find the range of movement for both the GM and TA individually the maximum range of the movement (maximal extension to maximal flexion) was calculated for each cycle.

To obtain the iEMG amplitude envelope to investigate when the amplitude decreased, the iEMG amplitude was full wave rectified and low-pass filtered (3^{rd} order low-pass Butterworth filter with 0.5 Hz cut off frequency). The maximal range of ankle angle and the iEMG amplitude envelope were normalized to 100 % and an average of the normalized data obtained for the three rats. These averaged data were used for the rest of the data analysis. Fatigue was defined as a decrease

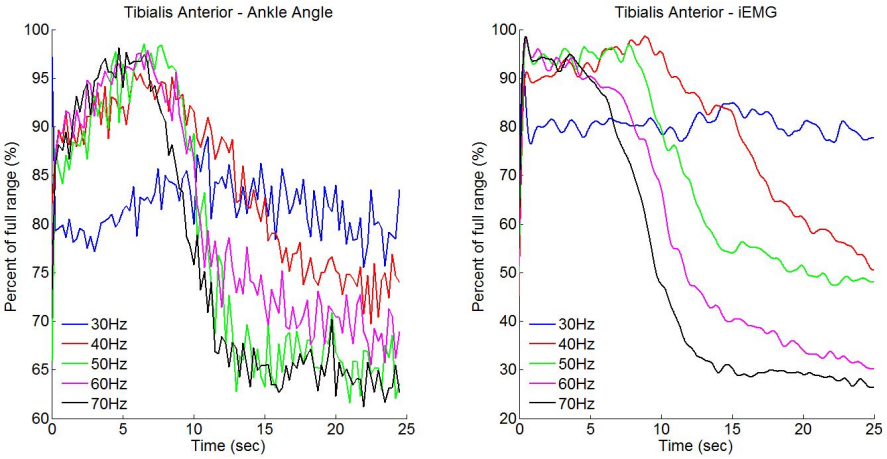

Fig. 2 Shows the range of movement of the ankle angle and the iEMG amplitude envelope over time for the TA. The maximum range for ankle angle and the iEMG amplitude envelope were normalized to 100 %. A correlation between the two signals can be observed. The decrease in the iEMG and ankle angle can be observed for the TA after 10-15 s indicating presence of fatigue.

in the ankle angle and the iEMG amplitude envelope. A correlation coefficient was calculated between the iEMG and ankle angle for both muscles for each frequency.

3 Results

The movement produced by the GM muscle for all the stimulation frequencies was in the range of 5-100 % of the maximum ankle angle, see Fig 1. The ankle angle increased during the first step cycles (approximately 10 s). After this it decreased during the rest of the step cycles. The response after 60 Hz decreased more rapidly than the other frequencies. The iEMG amplitude decreased continuously and rapidly for approximately the first 25 s (40-60 %). After this the rate of the decrease was less for the rest of stimulation (30-40 %).

Table 2 Correlations coefficients between ankle angle and iEMG for GM and TA for the different LIFE stimulations

	GM	TA
30 Hz	0.36	0.22
40 Hz	0.63	0.95
50 Hz	0.77	0.96
60 Hz	0.85	0.96
70 Hz	0.85	0.96

The movement produced by the TA muscle for all the stimulation frequencies was in the range of 65-100 % of the maximum ankle angle, see Fig 2. The ankle angle increased for the first 10 s. After this there was a decrease from 10-25 s thereafter a plateau was reached for the rest of the step cycles (65-85 %). The iEMG amplitude was found to be stable for the first 10 s. After this the 50, 60 and 70 Hz response rapidly decreased from 10-30 s (30-60 %), while the 30 Hz response produced no changes for the rest of the step cycles (80 %). The 40 Hz response also had a different tendency where it decreased less compared to 50, 60 and 70 Hz response (85 %). After this a plateau was reached for the 50, 60 and 70 Hz response (30-50 %). Here the 40 Hz response still decreased and did not reach a plateau (60 %). The ankle angle and the iEMG amplitude demonstrated similar response.

To quantify the degree of correlation between the ankle angle and the iEMG, the correlation coefficient between the ankle angle and the iEMG data were calculated (see Table 2). In the case of the GM muscle, it was observed that the higher the stimulation frequency that was applied, the higher the correlation observed (mean and standard deviation of 0.76 +/- 0.1 for 40 HZ – 70 Hz frequencies). In the case of the TA muscle the same tendency was observed, i.e. coefficients for the TA was high for the 40, 50, 60 and 70 Hz (0.95 +/- 0.1). This indicated that there was a good correlation between the ankle angle and the iEMG amplitude except when applying 30 Hz stimulation.

4 Discussion

In the current study the ankle angle was measured with kinematic data. This was compared to the iEMG amplitude to examine the correlation between these. The results showed that there was a correlation between the ankle angle and the iEMG. The correlation was higher for the TA than the GM.

4.1 Comparison of results with Other Studies

Previous studies from E. A. T. De Laat et al., S. G. Boe et al., J. R. Potvin et al. revealed that the relation between muscle force and amplitude is present and that the variable used for this was root mean square amplitude. Here they were looking at the linear force and root mean square amplitude using surface electrodes. [7], [8], [9]

4.2 Methodological Considerations

With the use of an animal model instead of a human model the physiological influences are not the same. When a human walks normally there is a force applied to the leg due to maintaining balance and standing upright. In this experiment, this was not taken into account since the leg of the rats was hanging freely and no external force was applied.

The markers were placed by visual inspection of the animal's anatomical structure. Placement of the markers may therefore have varied slightly from animal to animal. During the offline digitization of the kinematic video data it was possible that some degree of error was present in the marker identification because of indistinct images. Especially the toe marker was difficult to distinguish in the video during the extension phase of the movement.

During the stimulation the frequency was changed from 30-70 Hz. The stimulation was done in that same order during all of the experiments. It is not possible to judge if a particular stimulation frequency caused some cumulative influence on the next stimulation sequence. This could be solved by randomization.

A factor that may have had an influence on the results is potentiation. This occurs during continuous stimulation and also has a tendency to happen in fast twitch fibers, and causes a positive staircase phenomenon [10]. This could likely explain that some of the kinematic data had a tendency to not reach a maximum of 100 % just after the stimulation onset. Here the maximum movement range was not reached until approximately 10 s after the onset of the stimulation.

5 Conclusion

In the present study it was investigated if information extracted from iEMG could provide reliable information on a functional movement (i.e. the ankle angle) during development of muscle fatigue. A higher degree of correlation was found between iEMG and ankle angle when stimulation frequencies above 40 Hz were applied to produce muscle contractions. Also, the results indicate that TA may be a more reliable source of feedback than the GM since the TA had higher correlation coefficient than the GM (GM: average of 0.76 +/- 0.01, TA: 0.96 +/- 0.01).

Further research should focus on developing an animal model where the correlation coefficient would be higher. In a future perspective the improvement would be beneficial for the clinical rehabilitation with the FES for subjects with spinal cord injury.

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